Center of mass-based posturography for free living environment applications

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ABSTRACT

Background: Postural assessment is crucial as risk of falling is a major health problem for the elderly. The most widely used devices are force and balance plates, while center of pressure is the most studied parameter as measure of neuromuscular imbalances of the body sway. In out-of-laboratory conditions, where the use of plates is unattainable, the center of mass can serve as an alternative. This work proposes a center of mass-based posturographic measurement for free living applications.

Methods: Ten healthy and ten Parkinson's disease individuals (age = 26.1 ± 1.5, 70.4 ± 6.2 years, body mass index = 21.7 ± 2.2, 27.6 ± 2.8 kg/m^2, respectively) participated in the study. A stereophotogrammetric system and a force plate were used to acquire the center of pressure and the 5th lumbar vertebra displacements during the Romberg test. The center of mass was estimated using anthropometric measures. Posturographic parameters were extracted from center of pressure, center of mass and 5th lumbar vertebra trajectories. Normalized root mean squared difference was used as metric to compare the trajectories; Spearman's correlation coefficient was computed among the posturographic parameters.

Findings: Low values of the metric indicated a good agreement between 5th lumbar vertebra trajectory and both center of pressure and center of mass trajectories. Statistically significant correlations were found among the postural variables.

Interpretation: A method to perform posturography tracking the movement of the 5th lumbar vertebra as an approximation of center of mass has been presented and validated. The method requires the solely kinematic tracking of one anatomical landmark with no need of plates for free living applications.

1. Introduction

Postural assessment is crucial as risk of falling is a major health problem for the elderly (Tisserand et al., 2016). Both natural aging and several disorders, such as Parkinson's disease (Schoneburg et al., 2013), Alzheimer's disease (Leandri et al., 2009) or ankylosing spondylitis (Sawacha et al., 2012), increase significantly noise and uncertainty in postural control (Błaszczyk, 2016). Balance maintenance is a complex motor skill that involves the integration in the central nervous system of information originating from the vestibular, visual, and proprioceptive systems, which trigger eye and spinal reflexes (Fukunaga et al., 2014; Horak, 1997). Deficits in the central nervous system (Chong et al., 1999; Schoneburg et al., 2013), somatosensory and vestibular loss (Horak et al., 1990), reduced visual capabilities (Day et al., 1993; Guerraz and

Bronstein, 2008), musculoskeletal disorders (Bataller-Cervero et al., 2020), and pain-related etiologies (Kendall et al., 2018; Sung and Maxwell, 2017) have been linked with particular aberrations of the postural sway during instrumented tests.

The most widely used device to measure balance degradation and to test postural functions is the force plate (Winter et al., 1998; Wojtara et al., 2014), mainly through the recording during quiet standing of the trajectory of the center of pressure (CoP) (Koltermann et al., 2018; Palmieri et al., 2002). CoP tracks the point where the ground reaction forces resultant under the feet applies and it is considered to reflect in part the motor mechanisms that ensure balance, *i.e.*, the maintenance of the projection of the center of mass (CoM) inside the base of support; CoM is the parameter that regulates the body sway in the three-dimensional space (Michalak et al., 2019; Paillard and Noé, 2015).


Thus, static posturography on a force platform is a convenient tool to determine the risk of falling, especially for elderly people who might display limitations in performing functional tests, due to psychomotor disorders (Quijoux et al., 2021). However, in some cases there is the need to perform these tests in out-of-laboratory conditions to assess how different environments and medium (*i.e.*, water buoyancy (Palamara et al., 2017) or anti-gravity conditions (Birgani et al., 2016)) influence postural strategies. Two-dimensional CoM trajectory could be accurately derived knowing the CoP two-dimensional position and assuming that muscles adopt a spring-like behavior which causes the CoP to move in phase with the CoM as the body sways (Winter et al., 1998). However, this application would be far beyond from being applicable in free living conditions as it requires the use of a force plate for CoP measurement. Nevertheless, direct measurement of CoM is commonly derived by anthropometric measurements (Tisserand et al., 2016), through the use of stereophotogrammetric systems (Pavei et al., 2017; Winter et al., 1998). However, these instruments are expensive, time-demanding, their spatial operating range depends on the number of cameras available, their set up is not easily transferable from one place to another, and they require a skilled operator supervision, thus preventing their application in an everyday monitoring program in ecological settings (Yang and Pai, 2014). CoM estimation though inertial measurement units have recently gained a lot of attention in order to overtake the cumbersomeness and expensiveness of optoelectronic systems (Germanotta et al., 2021; Pavei et al., 2017), however, recent publications reported a lack of accuracy for some of these devices in real-life situations (Fasel et al., 2017; Pavei et al., 2020) and a lack of information on their sensitivity in comparison with force plates (Ghislieri et al., 2019). Despite those limitations, a recent review highlighted how there is an extensive literature evaluating balance through the adoption of wearable sensors (Ghislieri et al., 2019). The authors discussed that this might be due to its advantages in clinics, where it is thought to be more reliable within a wide range of pathologies (Giggins et al., 2014; Lin et al., 2021; Mancini et al., 2012) rather than clinical scores, which might be biased.

On the other hand, CoM has been observed as an imaginary point located in the lower part of the spinal cord, approximately at 55% of the body height (Saunders et al., 1953). Therefore, the purpose of the present work is to evaluate how CoM approximation with a single body anatomical landmark, placed on the 5th lumbar vertebrae (L5), could be used as an alternative to perform posturographic assessments. For the best of authors' knowledge, no literature reports results on posturographic assessment performed on three-dimensional CoM trajectory, thus correlation analysis was adopted to verify the agreement of L5-based postural variables with the one extracted considering CoP trajectory. The main hypothesis of the study is that posturographic assessment based on a simplified measurement of CoM would be informative and applicable in free living applications.

2. Methods

2.1. Subjects

For the purpose of the study, two different groups of people were considered: a convenient sample of 10 young healthy adults (YHA), as there is a consensus that the highest level of robustness in postural control is shown in young healthy subjects (Błaszczyk, 2016), and a group of 10 people with Parkinson's disease (PPD), as disrupted motor control in those people is largely documented in the literature (Baston et al., 2016; Horak et al., 1990; Schoneburg et al., 2013). For the convenience sample of ten HYA (males/females = 5/5, age = 26.1 ± 1.5 years, body mass index = 21.7 ± 2.2 kg/m^2) who participated in the study the protocol was approved by the Local Ethic Committee (CE/PROG 98/20–15/12/2020). Subjects gave written informed consent to the experimental protocol. YHA inclusion criteria were: participants free of musculoskeletal, neurologic, cardiopulmonary, and other systemic
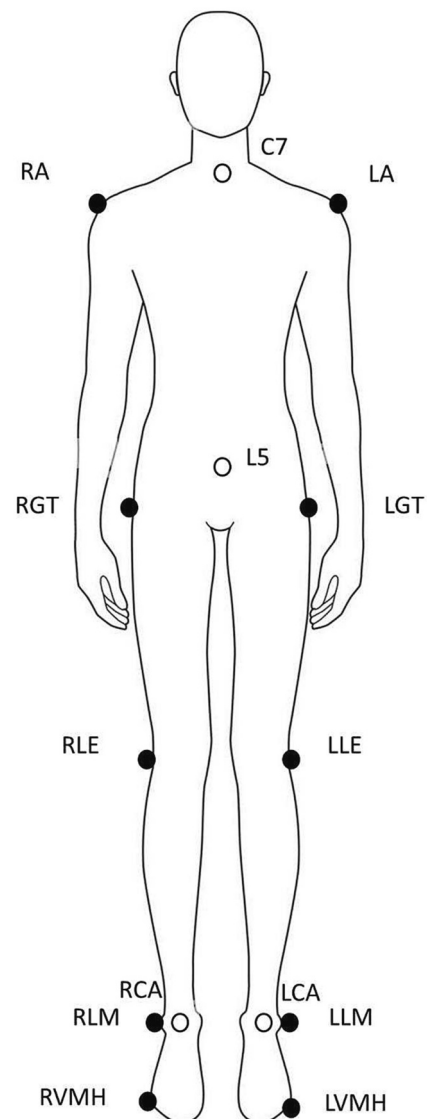
Fig. 1. Subset of markers considered for CoM computation. Black full dots represent the markers that are visible from the frontal view, white dots represent the markers placed on the subject's back. RA = right acromion, LA = left acromion, C7 = 7th cervical vertebra, L5 = 5th lumbar vertebra, RGT = right great trochanter, LGT = left great trochanter, RLE = right lateral epicondyle, LLE = left lateral epicondyle, RLM = right lateral malleolus, LLM = left lateral malleolus, RCA = right calcaneus, LCA = left calcaneus, RVMH = right 5th metatarsal head, LVMH = left 5th metatarsal head.

disorders, and able to walk independently. They were well informed about the experimental procedures and the purpose of the study.

The ten PPD (males/females = 5/5, age = 70.4 ± 6.2 years, body mass index = 27.6 ± 2.8 kg/m^2) were eligible for inclusion if they consented to participation, had PD diagnosed according to the current criteria: Hoehn and Yahr stage <3 on levodopa, and no history of falls in the past. Exclusion criteria were the presence of important freezing of gait affecting the recordings, dyskinesias and peripheral neuropathy, presence of co-morbidities preventing mobility (orthopedic diseases) or safe exercise (including major medical conditions such as malignancies), history of deep brain stimulation surgery or other conditions affecting stability (*e.g.*, poor visual acuity or vestibular dysfunction), Hoehn and Yahr stage >3 on levodopa, and inability to travel to the physiotherapy venues.

Table 1

Normalized root mean squared distance for the comparison between CoP, CoM and L5 trajectories both in mediolateral and anteroposterior directions. Results are reported for each research subject and as the average ± one times the standard deviation.

	Young healthy adults											
	Eyes open						Eyes close					
Subjects	CoM − L5		CoM − CoP		L5 − CoP		CoM − L5		CoM − CoP		L5 − CoP	
	ML	AP	ML	AP	ML	AP	ML	AP	ML	AP	ML	AP
HYA 1	11.02	8.52	7.01	4.37	4.92	3.92	14.69	11.83	8.93	10.83	5.76	5.64
HYA 2	10.41	8.75	7.06	5.23	7.92	5.65	12.46	11.76	7.41	8.39	9.19	5.13
HYA 3	7.61	11.97	5.31	9.49	5.25	5.51	6.47	8.02	6.51	11.92	5.96	7.27
HYA 4	17.30	14.79	9.78	6.79	8.28	5.66	14.57	18.85	22.89	18.63	8.19	6.38
HYA 5	14.27	7.65	8.28	4.42	7.56	5.63	23.95	36.79	7.77	13.94	16.82	20.06
HYA 6	8.16	12.99	4.64	6.39	5.15	4.68	6.59	12.36	6.55	8.30	6.30	5.21
HYA 7	10.41	17.31	8.04	8.03	7.04	6.81	14.53	7.41	8.02	6.44	6.56	5.29
HYA 8	10.75	25.66	4.59	11.99	8.46	4.85	17.65	23.62	13.60	11.03	13.99	6.24
HYA 9	7.76	35.83	6.75	14.99	5.99	4.58	7.49	16.45	6.19	12.16	5.39	7.72
HYA 10	15.07	8.49	8.66	7.07	6.14	7.48	13.82	9.88	8.50	9.01	6.68	5.49
Mean ± Standard Deviation	11.28 ± 3.28	15.19 ± 9.09	7.01 ± 1.75	7.88 ± 3.42	6.67 ± 1.35	5.48 ± 1.06	13.22 ± 5.41	15.69 ± 8.95	9.64 ± 5.12	11.07 ± 3.46	8.49 ± 3.88	7.44 ± 4.52

	People with Parkinson's Disease											
	Eyes open						Eyes close					
Subjects	CoM − L5		CoM − CoP		L5 − CoP		CoM − L5		CoM − CoP		L5 − CoP	
	ML	AP	ML	AP	ML	AP	ML	AP	ML	AP	ML	AP
PPD 1	4.59	11.54	4.48	7.86	3.74	5.05	11.03	10.41	7.90	8.04	5.67	5.23
PPD 2	10.78	17.92	10.38	12.71	6.88	7.30	8.67	31.45	9.75	14.10	8.39	5.81
PPD 3	21.78	13.09	12.59	10.87	4.49	6.47	–	–	–	–	–	–
PPD 4	14.92	7.80	10.11	7.45	5.21	6.10	13.51	28.62	8.14	16.09	4.31	8.20
PPD 5	18.05	20.62	10.99	14.19	4.90	8.95	13.42	27.18	9.12	16.51	5.06	5.86
PPD 6	12.47	11.29	7.27	5.84	3.28	7.62	20.94	8.78	8.43	5.39	5.19	4.76
PPD 7	13.22	9.33	5.99	6.51	5.48	4.96	16.39	19.34	8.91	14.74	7.54	8.30
PPD 8	10.99	18.60	8.08	23.92	4.66	7.14	16.31	25.17	10.51	14.06	5.27	6.39
PPD 9	10.21	6.66	9.04	6.73	7.05	6.73	23.89	9.83	9.93	12.98	8.06	8.92
PPD 10	15.67	15.74	9.93	11.39	7.29	7.21	7.80	15.94	8.68	11.14	7.14	7.29
Mean ± Standard Deviation	13.27 ± 4.72	13.26 ± 4.79	8.88 ± 2.45	10.75 ± 5.45	5.29 ± 1.39	6.75 ± 1.19	14.67 ± 5.36	19.64 ± 8.01	9.04 ± 0.87	12.56 ± 3.73	6.29 ± 1.49	6.75 ± 1.48

2.2. Experimental setup

The acquisitions of HYA's kinematic and kinetic data were performed at BioMovLab (Department of Information Engineering, University of Padova, Italy). A 6-cameras optoelectronic system (Smart E, 60–120 Hz, BTS Bioengineering S.p.A., Garbagnate Milanese (MI), Italy) captured the three-dimensional trajectories (frame rate = 60 Hz) according to the modified IOR-gait protocol as in (Sawacha et al., 2009). A force plate (FP4060, 960 Hz, Bertec Corporation, Columbus, Ohio, Canada) synchronously acquired the CoP displacement while the Romberg test was performed. The acquisitions of PPD's kinematic and kinetic data were performed at the Rehabilitation Facility, GVDR, Padova, Italy, throughout an 8-cameras optoelectronic system (Smart D, 300 Hz, BTS Bioengineering S.p.A., Garbagnate Milanese (MI), Italy) and a synchronized force plate (BTS-p-6000, 1200 Hz, BTS Bioengineering S.p.A., Garbagnate Milanese (MI), Italy). The same marker set as the one adopted for the YHA was applied. Moreover, the same examiner used the same system and the same protocol to test the subjects (Mancini et al., 2012).

Romberg test is commonly used as a tool to detect the presence of proprioceptive deficits (Fitzpatrick and McCloskey, 1994; Romanato et al., 2021; Sawacha et al., 2013) and consists in performing upright posture evaluation in both closed and open eyes conditions (Lanska and Goetz, 2000). Both the cohorts of participants performed the same posturography tasks. They were instructed to stand still on the force plate, with their feet placed to maintain the heels together and assure a 30 degrees angle between the right and left toes, and to relax the arms along the body. Subsequently, they were asked to maintain the upright standing position for 60 s with their eyes open (EO) and for 60 s with
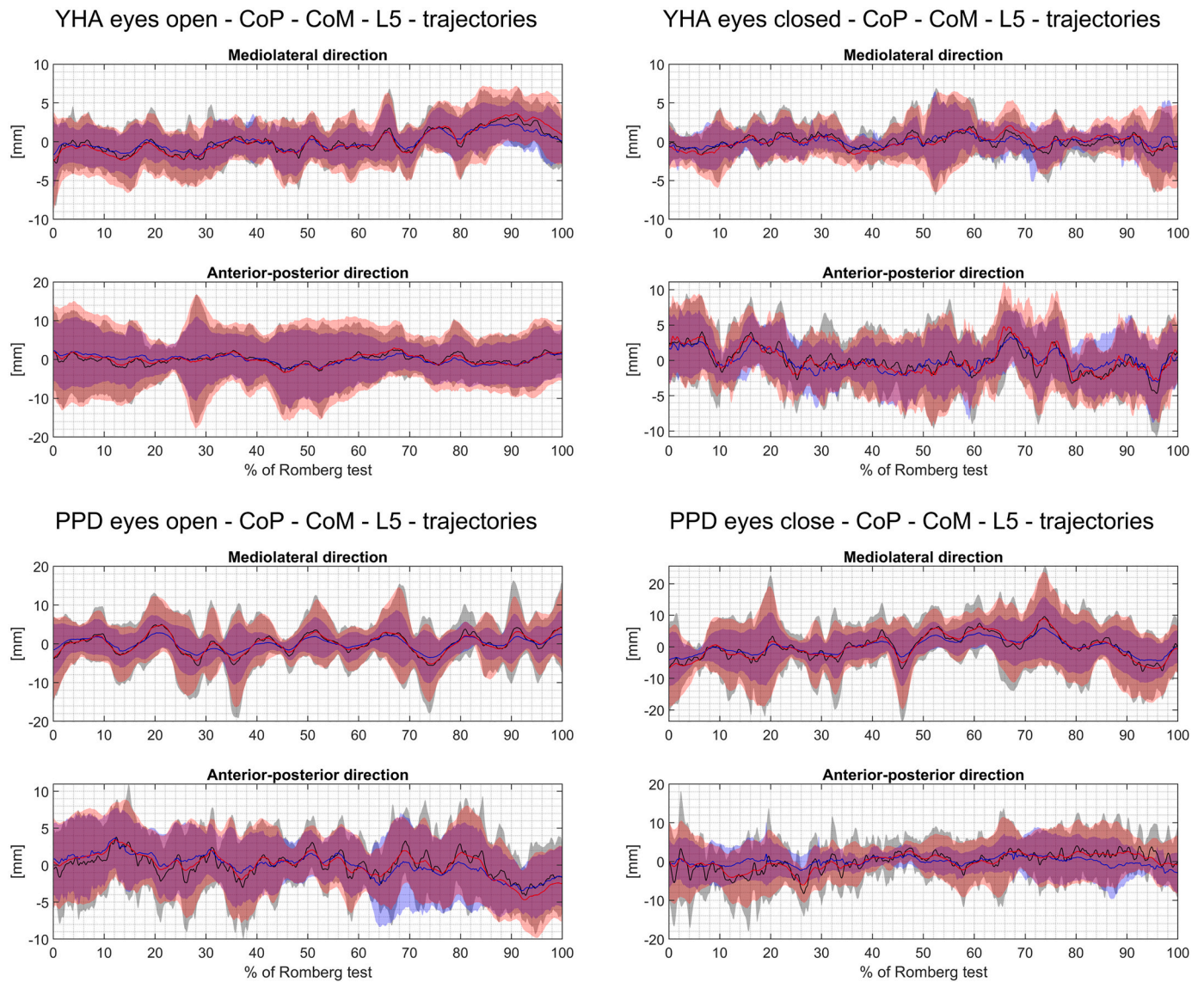
their eyes closed (EC) (Sawacha et al., 2013).

2.3. Data analysis

The kinematic model used to estimate the body CoM displacement consisted of a 7-segments body: feet, shanks, thighs, trunk with both head and arms modeled with 12 markers (Fig. 1) according to Dempster tables (Dempster, 1955). The L5 three-dimensional trajectory was adopted as further approximation of the body CoM. The CoP trajectory was registered during the Romberg test from the force plate.

The signals were lowpass filtered with a 5th order Butterworth filter and cut-off frequency of 7 Hz, and down sampled, where necessary, reducing the frequency to 60 samples/s. From the signals the following posturography parameters were computed:

- Ellipse, *i.e.*, area, containing 95% of the CoP (or CoM, or L5) data point
- Sway area, *i.e.*, a measure of the area included in CoP (or CoM, or L5) displacement per unit of time (mm^2/s)
- Root Mean Square (RMS) distance, *i.e.*, the mean squared value distance from the CoP (or CoM, or L5) time series center (mm/s)
- CoP (or CoM, or L5) path, *i.e.*, the total length of the CoP (or CoM, or L5) trajectory
- CoP (or CoM, or L5) path in both in the anterior-posterior (AP) and in the medio-lateral (ML) directions, *i.e.*, the sum of the distances between consecutive points in the AP and ML directions
- CoP (or CoM, or L5) mean velocity
- CoP (or CoM, or L5) mean velocity in both the AP and in the ML directions.

YHA eyes open - CoP - CoM - L5 - trajectories

YHA eyes closed - CoP - CoM - L5 - trajectories

PPD eyes open - CoP - CoM - L5 - trajectories

PPD eyes close - CoP - CoM - L5 - trajectories

Fig. 2. Mean trajectory of the CoP (in black), CoM (in blue) and L5 (in red) in percentage of the Romberg test. Standard deviation (± 1) clouds were reported accordingly. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Matlab (version R2021b, MathWorks, USA) was adopted for the data processing.

2.4. Statistics

Statistical analysis was performed using Matlab (version R2019b, MathWorks, USA). Normalized root mean squared difference (NRMSD) was used to compare the differences between the CoM and L5 trajectories in the mediolateral and anteroposterior directions. Normalization was performed with respect to the range of the measured values. It should be mentioned that, based on the theoretical model proposed by Winter and colleagues (Winter et al., 1998), were it is assumed that during upright posture muscles act like springs and cause the CoP to move in phase with the CoM, the CoP trajectory should have been retrieved from both the CoM and L5 trajectories, applying the following formulae (eq.1,2):

$$CoP\_CoM = CoM - \frac{I}{Wd}C\ddot{o}M \qquad \left(1\right)$$

$$CoP\_L5 = L5 - \frac{I}{Wd}\ddot{L}5 \qquad \left(2\right)$$

where *CoP_CoM* is CoP derived from CoM adopting the Winter theoretical model, *CoP_L5* is CoP derived from L5 bony landmarks displacement adopting the Winter theoretical model, *I* is the inertia of the body around the ankle moment, *W* is the weight of the body, and *h* is the height of the CoM (or L5) above the ankle joint. By considering the study of (Hasan et al., 1996) who demonstrated how CoP_CoM and CoP_L5 time series are highly correlated, we compared the trajectories experimentally measured rather than their projections on the CoP plane (*i.e.*, the CoP_CoM and the CoP_L5). This leaded to the following comparisons:

- CoM–L5
- CoM–CoP
- L5–CoP

Then, Spearman's correlation coefficient (after that normality assumption failed (Shapiro-Wilk Test)) was used to investigate either the

Table 2

Coefficient of determinations for the posturographic variables for both the conditions (closed eyes and open eyes) in both the populations. Statistically significant correlations are reported as * = p < 0.05.

Young healthy adults – eyes open

Variables	CoP			CoM			L5			ρ	
	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	CoM - CoP	L5 - CoP
Ellipse (mm²)	432.23	279.76	847.63	249.42	185.90	554.04	520.68	342.16	1001.49	0.94*	0.94*
Sway Area (mm²/s)	0.35	0.19	0.51	0.13	0.10	0.32	0.22	0.13	0.42	0.94*	0.99*
RMS	7.97	3.26	9.26	6.25	3.09	8.85	9.09	3.94	11.32	0.96*	0.99*
Path (mm)	387.91	93.71	273.47	178.39	76.02	266.09	214.79	76.57	288.85	0.64*	0.96*
Path ML (mm)	196.61	55.07	134.51	93.57	51.27	172.59	105.28	38.98	135.78	0.72*	0.85*
Path AP (mm)	291.87	68.76	213.51	131.37	47.95	165.24	169.49	61.86	236.58	0.61*	0.90*
Mean velocity (mm/s)	9.71	2.37	7.02	4.48	1.93	6.78	5.38	1.95	7.35	0.64*	0.92*
Mean velocity ML (mm/s)	4.92	1.39	3.36	2.35	1.30	4.39	2.64	0.99	3.46	0.72*	0.85*
Mean velocity AP (mm/s)	7.31	1.75	5.48	3.30	1.21	4.21	4.25	1.58	6.02	0.61*	0.90*

Young healthy adults – eyes close

Variables	CoP			CoM			L5			ρ	
	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	CoM - CoP	L5 – CoP
Ellipse (mm²)	251.01	195.53	645.78	155.99	155.09	485.36	278.27	256.83	852.78	0.75*	0.98*
Sway Area (mm²/s)	0.26	0.15	0.48	0.13	0.17	0.53	0.19	0.17	0.53	0.72*	0.88*
RMS	5.59	1.54	5.38	4.37	1.67	5.39	5.79	1.94	6.72	0.72*	0.84*
Path (mm)	447.28	116.17	316.41	255.54	210.15	701.68	291.37	207.39	726.03	0.50	0.75*
Path ML (mm)	196.89	56.48	169.37	125.26	133.50	444.02	110.49	39.32	124.65	0.67*	0.84*
Path AP (mm)	360.06	98.72	266.44	198.26	149.54	501.28	249.88	202.89	700.23	0.62*	0.88*
Mean velocity (mm/s)	11.18	2.90	7.91	6.39	5.25	17.53	7.28	5.18	18.14	0.50	0.75*
Mean velocity ML (mm/s)	4.92	1.41	4.23	3.13	3.34	11.09	2.76	0.98	3.12	0.67*	0.84*
Mean velocity AP (mm/s)	8.99	2.47	6.66	4.95	3.74	12.53	6.24	5.07	17.49	0.62*	0.88*

People with Parkinson's Disease – eyes open

Variables	CoP			CoM			L5			ρ	
	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	CoM - CoP	L5 - CoP
Ellipse (mm²)	525.71	364.41	1288.81	256.03	155.02	464.56	469.50	294.76	1049.74	0.87*	0.78*
Sway Area (mm²/s)	0.54	0.42	1.41	0.16	0.12	0.39	0.25	0.19	0.69	0.42	0.78*
RMS	8.13	3.98	14.14	5.08	2.45	7.71	7.73	3.66	13.18	0.83*	0.76*
Path (mm)	579.95	253.67	723.02	228.28	121.91	412.62	300.06	183.51	629.74	0.52	0.79*
Path ML (mm)	367.32	217.02	690.13	142.68	110.53	373.68	223.25	190.58	640.38	0.75*	0.85*
Path AP (mm)	368.50	134.71	396.17	143.28	60.79	185.54	150.96	33.94	105.79	0.55	0.67*
Mean velocity (mm/s)	14.49	6.33	18.07	5.07	3.05	10.31	7.49	4.59	15.74	0.52	0.79*
Mean velocity ML (mm/s)	9.21	5.42	17.25	3.57	2.76	9.34	5.58	4.76	16.00	0.75*	0.85*
Mean velocity AP (mm/s)	1.56	3.37	9.90	3.58	1.52	4.64	3.77	0.85	2.64	0.55	0.67*

People with Parkinson's Disease – eyes close

Variables	CoP			CoM			L5			ρ	
	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	CoM - CoP	L5 - CoP
Ellipse (mm²)	1129.63	1039.84	2835.88	393.59	322.35	897.79	895.01	699.19	1834.34	0.98*	0.97*
Sway Area (mm²/s)	1.31	1.29	3.95	0.24	0.17	0.49	0.45	0.35	0.95	0.93*	0.93*
RMS	11.02	5.99	15.59	6.79	3.69	10.34	10.39	5.49	14.79	0.97*	0.95*
Path (mm)	885.41	495.28	1296.62	295.68	137.88	458.88	368.91	207.58	655.64	0.83*	0.80*

Table 2 (*continued*)

People with Parkinson's Disease – eyes close											
Variables	CoP			CoM			L5			ρ	
	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	Mean	Standard deviation	Max-Min range	CoM - CoP	L5 - CoP
Path ML (mm)	511.19	305.32	855.75	180.25	128.35	424.68	262.97	200.72	635.06	0.92*	0.87*
Path AP (mm)	607.45	312.99	833.19	190.00	69.71	196.04	200.30	59.90	179.90	0.68*	0.73*
Mean velocity (mm/s)	22.17	11.44	32.40	7.40	3.44	11.47	9.23	5.18	16.38	0.83*	0.80*
Mean velocity ML (mm/s)	12.80	7.61	21.38	4.51	3.20	10.61	6.58	5.01	15.87	0.92*	0.87*
Mean velocity AP (mm/s)	15.21	7.79	20.82	4.76	1.74	4.89	5.01	1.59	4.49	0.68*	0.73*
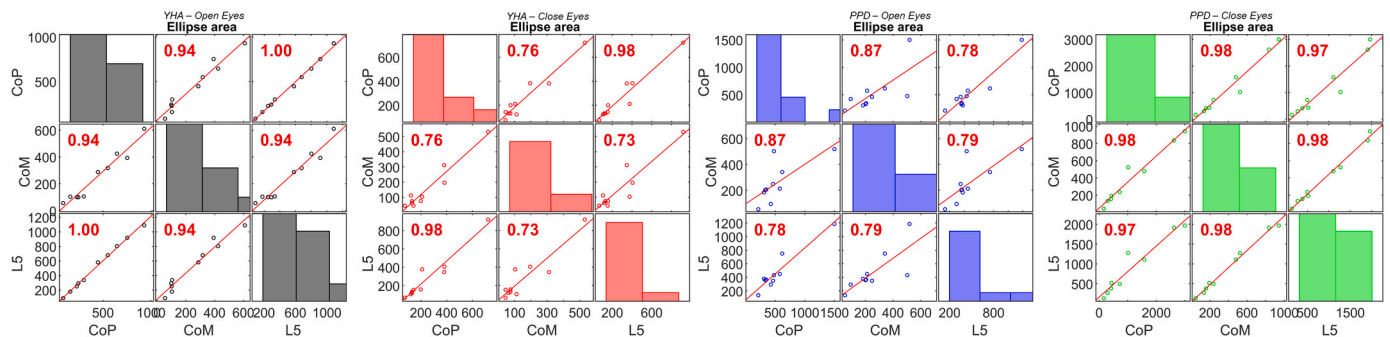
Fig. 3. Ellipse area data distribution for the four conditions (*i.e.*, YHA – open eyes in black, YHA – close eyes in red, PPD – open eyes in blue, PPD – closed eyes in green). Each subplot represents the correlation plot between the postural variable obtained by the CoP, CoM or L5 measurements. Spearman's correlation coefficient is reported within each box, red font indicates statistically significant correlation ($p < 0.05$)). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

correlation between the posturography results obtained through the CoP analysis and the CoM analysis, either the correlation between the posturography results obtained through the CoP analysis and the L5 analysis. Correlation results were interpreted as follows: 0.0–0.2 little if any; 0.2–0.4 weak; 0.4–0.7 moderate; 0.7–1.0 strong (Germanotta et al., 2021). Statistical significance was set at $p < 0.05$.

A *post hoc* power analysis (G*Power, v. 3.1.9.4, Universitat Kiel, Germany (*Sample size determination and power analysis using the G*Power software - PMC*, 2023)) was performed (statistical test: correlation – point biserial model) to determine the statistical power in both the populations.

3. Results

All the data have been processed successfully, except for only one member of the PPD group whose kinematic and kinetic measures for the eyes close condition were corrupted. Table 1 reported the NRMSD for the comparison between CoP, CoM and L5 trajectories both in mediolateral and anteroposterior directions. Time series of the three measures trajectories in percentage of the task were reported in Fig. 2.

R^2 of the postural variables were reported in Table 2 for both the conditions in both the populations.

As an example, were reported the data distribution (obtained by means of a customization of the **corrplot** Matlab function) of one postural variable among all the different conditions (*i.e.*, YHA – open eyes, YHA – close eyes, PPD – open eyes, PPD – closed eyes) (Fig. 3), as well as the data distribution of all the extracted postural variables for one condition (*i.e.,* YHA – open eyes) (Fig. 4). All results are reported in the supplementary material file (Supplementary material A, Figure SM1, SM2 and SM3).

4. Discussion

A simplified measurement of CoM, as its approximation with L5 anatomical landmark, was employed to perform postural assessment in a group of 10 young healthy adults and 10 people with Parkinson's disease. The proposed method was compared with postural assessments either considering a direct measurement of the CoM based on the adopted kinematic model (Fig. 1) or the CoP measured *via* force plate. Two-dimensional trajectories highlighted similar patterns within the three measurements (Fig. 2), and low values of NMRSD (Table 1). It is worth to be mentioned that highest values of NMRSD were found both in anteroposterior direction, where kinematics models are less accurate as small displacements are more sensitive to tracking errors (Gutierrez-Farewik et al., 2006), and when comparing L5 kinematic model with the respect to the CoM kinematic model, in agreement with what reported in dynamic situations by Pavei and colleagues (Pavei et al., 2017). Those values were considered to be acceptable for the aim of our work as, to provide an alternative measurement of the CoM able to faithfully reproduce its tracking over space, was beyond our purposes.

The author's main hypothesis was verified by the correlation analysis carried out on the postural variables. In fact, high correlations between the proposed method and the gold standard have been found for each considered variable, for each condition and in both the research participants' groups (Table 2). These results could suggest to perform posturographic assessment based on the chosen anatomical landmark in absence of force plates. Current literature has focused on the translation of clinical biomechanical assessments in out-of-the-lab conditions in order to retrieve meaningful information in environments linked with everyday activities (Ghislieri et al., 2019; Mancini et al., 2012; Yang and Pai, 2014). Moreover, the interest to evaluate how different environments and mediums influence postural strategies has arisen. Being able to understand how water buoyancy or anti-gravity environments impact on postural strategies would be beneficial for target rehabilitation
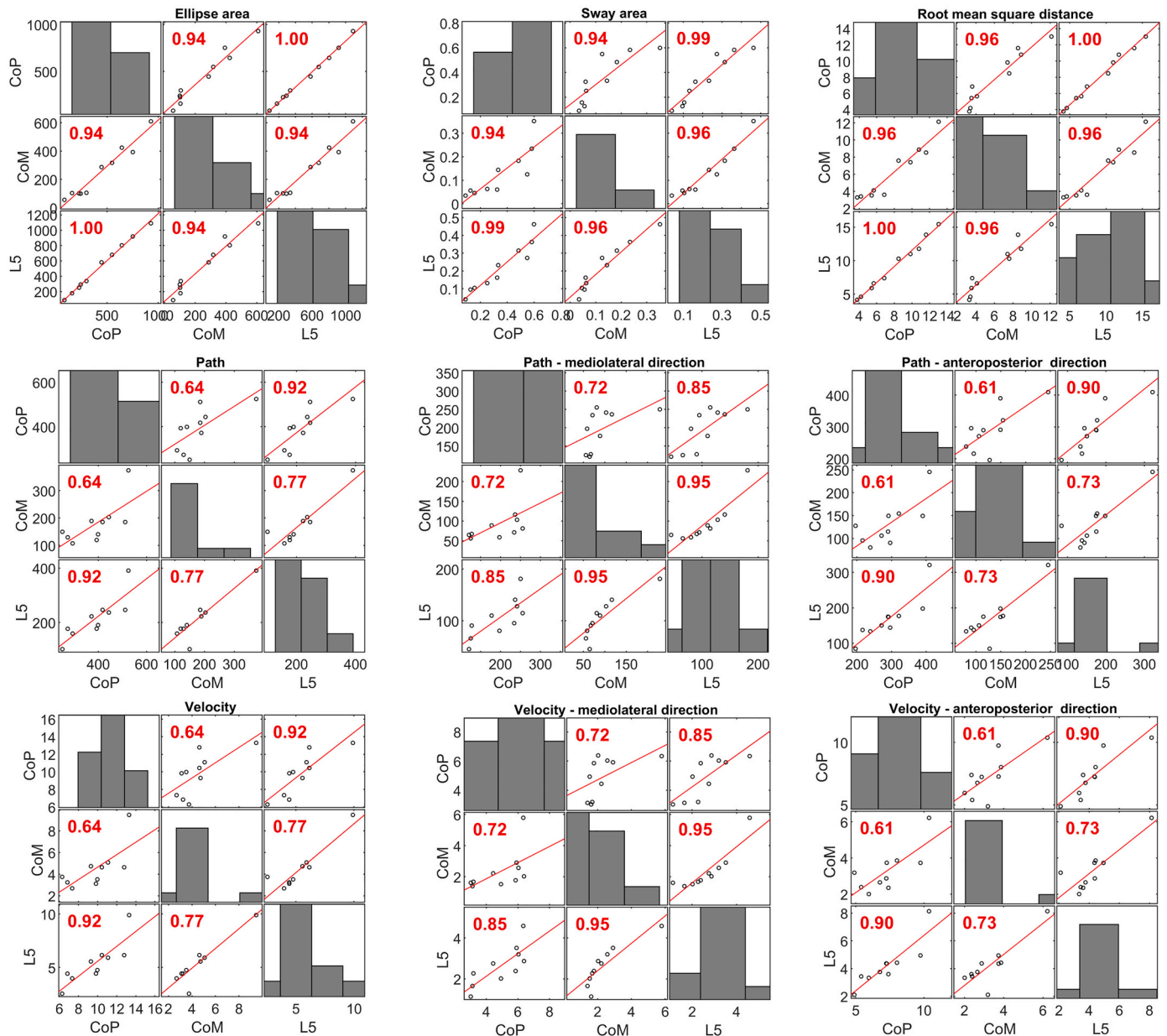
Fig. 4. Postural variables data distribution for one specific condition (*i.e.*, YHA – open eyes). Each subplot represents the correlation plot between the postural variable obtained by the CoP, CoM or L5 measurements. Spearman's correlation coefficient is reported within each box, red font indicates statistically significant correlation (p < 0.05). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

treatments for those people suffering from balance impairments caused by neurological or neurodegenerative disorders (Birgani et al., 2016; Palamara et al., 2017). As the use of a force plate is not easy to adopt in underwater conditions (*e.g.*, in rehabilitation pools), being able to track the CoM sway from a single anatomical point would be not only informative, but also a viable solution. In fact, video-based approaches, as the one proposed in (Sawacha et al., 2021), could be effortlessly implemented with waterproof commercial cameras. In addition to this, the method could be employed for on-field biomechanical evaluations in athletes aiming at the return on field after structural injuries (Guiotto et al., 2021).

Limitations of the current work must be acknowledged. First, no *a priori* power analysis was conducted to determine if the sample size was adequate to address the authors' research question. However, the *post hoc* power analysis indicated a statistical power above 0.90 in both the populations. Another critical aspect might be the fact that the data of the

two groups have been collected in different laboratories settings. Nevertheless, intergroup comparison was not the purpose of our study, while high correlations obtained in both settings and in both the assessed groups suggest the robustness of the proposed approach. In fact, measures of balance were correlated with the reference either when considering a group of YHA, where motor control in quiet standing is considered to be robust, or when performing postural assessment in those people where balance is considered to be highly compromised. Finally, the gold standard for posturography assessment remains CoP measurement *via* force plates. However, to consider the CoM, rather than the CoP, in posturography could be valuable from a clinical perspective. In fact, the CoM is thought to be the variable by which the central nervous system controls the postural body sway (Paillard and Noé, 2015), so that the associated theories should be based on the motion of this variable (Corazza and Andriacchi, 2009).

To the authors' knowledge, this is the first paper reporting reliability

of a posturographic assessment adopting a kinematic tracking of the CoM from a single body landmark. Future research should be focused on verifying whether the proposed method is suitable to assess intergroup differences in balance in cross-sectional studies' designs, or to be adopted in free living conditions (*i.e.,* rehabilitative pools, sport, and medical centers).

5. Conclusion

In conclusion a method to perform posturography tracking the movement of L5 as an approximation of CoM has been presented and assessed. The method requires the solely kinematic tracking of one anatomical landmark with no need of plates which might be useful for assessment in out-of-lab conditions. Results reported low levels of NMRDS across the different testing conditions and the different populations, which indicated an acceptable tracking of the L5 trajectory with the respect of both the CoP and the CoM. Furthermore, statistically significant correlations were found among the extracted postural variables, suggesting that a reliable balance assessment would be obtained with such an approximation. The method would be easily accessible in those free-living conditions where the surrounding environment or the different mediums would not consent to the use of a force plate. This approach could be in line with current research targeting mobile video motion analysis applications (Halilaj et al., 2021; Parks et al., 2019).

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2023.105950.

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