Journal of Human Kinetics volume xx/xxxx, x–x DOI: 10.5114/jhk/224588 x
Section I – Kinesiology

The Optimal Control of Human Standing Posture

by

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Human erect posture has turned postural stability control into a key determinant of human motor behavior. The process of postural control can be simplified to the maintenance of the body's center of gravity (COG) within the base of support (BOS). In such a biomechanical model, the central COG position within the BOS and minimized movement-related disturbances are symptomatic of optimal stability. The postural stability control and its optimization are shaped during ontogenetic development and achieve their optimum in adolescents. To assess the optimal postural stability, a group of 100 young (50 females and 50 males), healthy, and able-bodied subjects participated in the study. The assessment of their postural stability was based on trajectories of the COG oscillations during ten 60-s quiet standing trials with and without visual input. The observed COG oscillations remained within 10 mm and 2 mm in anteroposterior (AP) and mediolateral (ML) planes, respectively. The COG oscillations significantly correlated with individual body anthropometry and the size of the base of support. The findings of this novel analysis of postural sway can be used as a reference for diagnostic purposes, particularly in age-related changes of postural stability and neuromuscular disorders.

Keywords: *postural control; center of gravity; optimal stability; sway analysis*

Introduction

Human standing posture is inherently unstable and to remain stable it must be controlled online by the multimodal sensory-motor control system (Błaszczyk et al., 1994; Massion, 1998). The optimal control of the erect posture provides maximal resistance to environmental and endogenous interferences (Błaszczyk, 2016). This unique capability of human motor control is formed during ontogeny, reaching its maximum efficiency in adolescents (Błaszczyk and Fredyk, 2021; Błaszczyk and Beck, 2023). It can then be maintained at the maximal level for the young adolescent period, depending mainly on habitual motor activity. A sedentary lifestyle, motor disease, and aging processes lead to a progressive dysfunction of movement-posture interaction, leading to postural instability and culminating in accidental falls in older adults (Błaszczyk et al., 1997).

In quiet standing, postural control can be assessed based on the maintenance of the body's center of gravity (COG) over the base of support (BOS) (Błaszczyk et al., 1993; Masani et al., 2003; Mergner et al., 2003; Morasso et al., 2019). This process is approximated by the inverted pendulum model (Błaszczyk, 2016; Błaszczyk et al., 2020; Loram et al., 2001). In the single inverted pendulum model, the integration of proprioceptive, visual, and vestibular inputs form in the brain the body schema (Assaiante et al., 2014) that, in postural control, determines the ankle angle and thus the COG setpoint within the BOS (Masani et al., 2003; Merfeld et al., 1999; Proske, 2015; Winter et al., 2001). The toppling torque-driven single inverted pendulum model (SIP) approximates well the control of human standing posture (Boonstra etal., 2013; Gage et al., 2004; Kennedy et al., 2014; Lakie et al., 2003). The posture is maintained by ankle joint stiffness which compensates for the toppling torque (Amiri and Kearney, 2019; Hettich et al., 2014). Importantly, the passive stiffness contributes about 90% (Héroux et al., 2014; Loram and Lakie, 2002). During quiet stance, both lower limbs act as

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interacting single inverted pendulums with pivots in the ankle joints. In this case, the toppling torque is directed frontally, whereas the lateral torques are negligible. Due to the asymmetry of neuromuscular control for each limb, the chaotic breathing and hemodynamics-related COG perturbations, secondary, chaotic mediolateral displacements are observed (Błaszczyk and Klonowski, 2001).

It has been hypothesized that the stability control model of human erect posture is based on an optimal control strategy (Qu et al., 2007). In the SIP model, the human body is represented by the center of gravity supported by a simple, single-segment inverted pendulum (SIP). In young people with normal body structure, the length of the pendulum is roughly equal to half of their body height. Consequently, in the SIP model, neural control of the erect posture may be limited to compensation for changes in posture destabilizing torques mainly by a transient activation of the ankle stabilizing muscles, i.e., the triceps surae. In the maximally stable control, the COG position should be maintained in the middle of the BOS (Błaszczyk et al., 1994). Such positioning provides an equal probability of recovering balance in case of an unpredicted perturbation. During quiet standing, the COG position is continuously disturbed by breathing (Kuznetsov and Riley, 2012) and hemodynamics (Conforto et al., 2001; Nomura et al., 2013). The resultant quasi-periodic and asynchronous COG displacements induce small and random disturbing torques to the ankle joints (Loram et al., 2001). Optimal stability of the upright posture depends on an individual's capability to generate effective corrective forces. Consequently, we assumed that the optimal postural control must be characterized by the ability to maintain precisely the central COG position within the BOS, with minimal spontaneous oscillations. Finally, we assumed that the nervous system must provide robust control, which allows for maintaining stable posture and desired motor performance despite uncertainties and disturbances within the organism or its environment. To verify these hypotheses, we collected body sway trajectories during quiet stance in a large group of young, healthy, and able-bodied individuals. The recorded COP signals were then used to calculate the COG trajectories (Błaszczyk, 2008). This novel approach to postural control allows for the most reliable assessment of postural control (Błaszczyk et al., 2016). In our first attempt to address the postural control problem, we compared the characteristics of COG trajectories in both male and female subjects while standing with eyes open (EO) and eyes closed (EC).

Methods

Participants

A group of fifty young healthy female subjects (mean age = 20.7 ± 1.8 yrs; body height = 1.67 ± 0.06 m; body mass = 59.0 ± 7.1 kg) and a group of fifty healthy male subjects (mean age = 21.2 ± 2.0 yrs; body height = 1.80 ± 0.07 m; body mass = 77.0 ± 8.7 kg) participated in the study. All subjects gave their written informed consent before participation. The inclusion criteria were: age between 19 and 25 years and the body mass index (BMI) within the normal range (18.5–24.9). The exclusion criteria were balance disorders, lower extremity and vertebral column deformities or injuries, and uncorrected vision. Detailed anthropometric characteristics of the study groups are shown in Table 1.

This study was conducted following the principles of the Declaration of Helsinki, and approved by the Senate Ethics Committee of the Jerzy Kukuczka Academy of Physical Education in Katowice, Katowice, Poland (protocol code: 1/2011; approval date: 26 October 2011).

Measures

Mean anteroposterior (AP), mediolateral (ML), and the total COG velocity were computed by dividing the path lengths by the duration of the trial, i.e., 60 s. The mean velocity VCOG was used to define the magnitude (SVm) of the COG stability vector, while the azimuth (φ) of the vector was computed according to the formula (Błaszczyk, 2016):

$$\varphi = SVa = \arctan\frac{SAP}{SML} \quad (1)$$

where: *SAP*, *SML* were the anteroposterior and the mediolateral COG path length, respectively.

To get a deeper insight into postural control, we "centralized" COG trajectories to the origin of the Cartesian axes coordinate (de Oliveira, 2017). Such normalization allows for the analysis of the individual stability vector variables

separately in each quadrant, providing information on the directional symmetry of postural control (Figure 1).

The directional sway indices (DI) were computed according to the following formulas (Błaszczyk et al., 2014):

$$DIAP = \frac{SAP}{Stot} \quad \text{and} \quad DIML = \frac{SML}{Stot} \tag{2}$$

where: *Stot* was the total length of COG trajectory during the 60-s trial.

Ranges of COG AP and ML displacement during the trial were defined as:

$$RAP = AP_{max} - AP_{min} \; and \; RML = ML_{max} - ML_{min} \tag{3}$$

Based on the ranges and taking length (*LBOS*) and width (*WBOS*) of the BOS, the stability indices (SI) were calculated as follows:

$$SIAP = 1 - \frac{RAP}{LBOS} \quad \text{and} \quad SIML = 1 - \frac{RML}{WBOS} \tag{4}$$

Design and Procedures

Before starting the experiment, the outline of the participants' feet was taken while standing on the force plate in the habitual position. The foot envelope was used to determine the length and width of the base of the support area. During the experiment, participants were asked to stand still for 60 s on the force plate. Their center-of-pressure (COP) trajectories were recorded through the force plate (Type 9281C Kistler Group, Switzerland) with a sampling frequency of 40 Hz. Each of the participants completed five trials with eyes open (EO) and five with eyes closed (EC). The anteroposterior (AP) and mediolateral (ML) COG trajectories were retrieved then from the COP signals by low-pass filtering (Cheby 2 in Matlab) at 0.4 Hz (Błaszczyk, 2008). The filtering was preceded by normalization, i.e., by the removal of constant trends from the AP and ML trajectories.

Statistical Analysis

All data analyses were performed with Statistica v.10 (StatSoft, USA) software. To verify the normal distribution of the analyzed data, the W Shapiro-Wilk test was used. Two-way repeated-measures analysis of variance (ANOVA) followed by the least significant difference (LSD) post hoc test was applied to compare variables between both groups and both experimental conditions. The bivariate correlation analysis between COG sway variables and participants' anthropometry and BOS size was performed with the Pearson r-test.

Results

Characteristics of the COG Trajectory

We analyzed ranges (peak-to-peak amplitude) of the COG orbit around the setpoint while standing quietly with and without visual input. Under the EO condition, the range of ML oscillations reached the p-p value of 8.5 ± 3.1 mm, while during the EC tests, the mean value of the ML range increased by 22.3% ($p \leq 0.001$), reaching the value of 10.4 ± 3.1 mm. Greater p-p COG amplitudes were observed in the AP plane (32.7%, $p \leq 0.001$), reaching the values of 12.1 ± 3.2 mm and 15.7 ± 4.1 mm for the EO and EC conditions, respectively.

Based on the linear p-p ranges data and participants' anthropometric data, the angular ranges of COG displacements of the model were computed. The estimated angular range in the AP plane was 0.8 ± 0.2° and 1.1 ± 0.2° for EO and EC conditions, respectively. The estimated range of angular displacements in the ML plane was 0.5 ± 0.2° and 0.7 ± 0.2° for EO and EC conditions, respectively.

ML and AP COG Velocities

Changes in the mean COG velocity in the AP and ML planes were compared while standing quietly with EO and EC. The average speed of COG AP was 1.8 ± 0.5 mm/s in the female group and 1.9 ± 0.4 mm/s in the male group when tested with EO. The value of COG AP velocity increased to 2.1 ± 0.6 mm/s and 2.3 ± 0.6 mm/s when measured with EC in the female and male groups, respectively. The analysis of variance showed only a significant effect of visual input on the mean COG AP speed ($F_{1.98} = 109.1$, $p \leq 0.0001$).

Both factors (group and vision) differently affected the mean COG ML velocity. The mean velocity in the female group (EO) was 1.3 ± 0.4 mm/s and increased to1.6 ± 0.6 mm/s while tested with EC. The average COG ML velocity in the male group was 1.4 ± 0.4 mm/s and 1.9 ± 0.6 mm/s under

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EO and EC conditions, respectively. The ANOVA confirmed significant effects of sex ($F_{1.98} = 6.56$, $p \leq 0.02$) and vision ($F_{1.98} = 142.1$, $p \leq 0.0001$) on COG ML velocity. Also, the interaction of both factors was statistically significant ($F_{1.98} = 13.55$, $p \leq 0.0005$). The results of these analyses are summarized in Figure 2.

COG Stability Vector and Stability Index

Two components of the stability vector, the magnitude (SVm) equal to the mean COG velocity, and the vector azimuth (SVa = φ), determine the magnitude of COG local disturbance. The mean SVm in the female group was 2.5 ± 0.6 mm/s (EO) and increased to 2.9 ± 0.8 mm/s (EC). The impact of visual input on SVm was more pronounced in male individuals, where mean COG velocity increased from 2.6 ± 0.6 mm/s (EO) to 3.4 ± 0.9 mm/s (EC). The ANOVA confirmed the significant effect of both factors, i.e., sex ($F_{1.98} = 173.2$, $p \leq 0.0001$) and vision ($F_{1.98} = 4.7$, $p \leq 0.05$), on the mean COG velocity. Also, the interaction of both factors reached the significance level ($F_{1.98} = 7.62$, $p \leq 0.01$).

The effects of sex and vision on the sway azimuth were examined next. In the group of women, the azimuth remained at the same level of 0.9 ± 0.1 rad, under both EO and EC conditions. In the male group, the COG azimuth was 0.9 ± 0.09 rad (EO) and increased to 0.96 ± 0.1 rad in testing with EC. The ANOVA showed a highly significant effect of vision ($F_{1.98} = 262.3$, $p \leq 0.0001$) in addition to a significant sex x vision interaction ($F_{1.98} = 13.4$, $p \leq 0.0005$). The results of these analyses are presented graphically in Figure 3.

Based on the stability radius concept (Bingham and Ting, 2013), the postural stability indices were computed to assess the optimal postural control. In the EO trials, the SIAP remained at the level of 0.95 ± 0.01 and decreased to 0.94 ± 0.01 under the EC condition, whilst the respective SIML values were 0.97 ± 0.01 (EO) and 0.96 ± 0.02 (EC).

COG Directional Indices (DIAP and DIML)

The ANOVA showed that the sex factor did not have a significant effect on DIAP ($p \leq 0.1$), while it differed significantly between EO and EC trials ($F_{1.98} = 5.65$, $p \leq 0.02$). Also, the sex x vision interaction was statistically significant ($F_{1.98} = 5.22$, $p \leq 0.03$). In the female group, DIAP stayed at the level of 0.73 ± 0.09 in the EO and EC tests, while in the male group, DIAP decreased from 0.72 ± 0.07 (EO) to 0.7 ± 0.07 (EC).

The results of ANOVA performed for DIML confirmed the significant effect of vision ($F_{1.98} = 5.58$, $p \leq 0.02$) and the significant interaction of the sex and vision factor ($F_{1.98} = 5.12$, $p \leq 0.03$). In the female group, the DIML was independent of vision, remaining at the level of 0.52 ± 0.1, while in the male group, DIML was 0.54 ± 0.08 and 0.57 ± 0.08 in EO and EC tests, respectively. These results are presented in Figure 4.

Correlation Analysis

The results of the correlation analysis are summarized in Table 2. Particularly in the female group, the COG output measures, including the COG sway vector azimuth, and the DIAP and DIML, were correlated with the participants' body height, BMI, and size of the BOS. Importantly, the significant correlations between body anthropometry and stability vector azimuth were specific only for the female participants, and no such correlations were found in the male group. The highest positive correlations (0.92 and 0.8) were found between the COG sway vector amplitude (VCOG) and VAP velocity, both under EO and EC conditions.

Discussion

In the present experiment, we applied a novel method of postural stability analysis based on the center of gravity (COG) trajectories. The majority of published, up-to-date papers use the center of pressure (COP) signals for this purpose, which renders the results unreliable (Błaszczyk and Beck, 2023; Prosperiniet al., 2013). In our search for the optimal control, we analyzed COG random oscillations while standing quietly in young adults known for the most stable posture, whose sensory and neuromuscular development had already been completed (Assaiante et al., 2014). The quality of optimal postural control was assessed using system output characteristics, i.e., COG trajectories while standing quietly with eyes open and closed. Two standardized sway measures were used to analyze the control: stability vector and directional indices. The results showed that in young individuals during the quiet stance, the COG habitual position was maintained in the middle of the BOS and due to endogenous disturbances, the COG oscillated slowly around it

within the peak-to-peak range, thus not exceeding 20 mm, which corresponded to ±1° range of angular displacement. The analyzed COG measures correlated significantly with body anthropometry and the size of their BOS.

Table 1. Anthropometric characteristics (mean ± S.D.) of the experimental groups.

Group	Female	Male	$p \leq$
Age (yrs)	20.7 ± 1.8	21.2 ± 2.0	NS
Body height (m)	1.67 ± 0.06	1.80 ± 0.07	0.05
SIP length (m)	0.96 ± 0.05	1.0 ± 0.04	NS
Body mass (kg)	59.0 ± 7.1	77.0 ± 8.7	0.05
BMI (kg/m^2)	21.0 ± 2.2	23.8 ± 2.5	0.05
BOS length (cm)	24.8 ± 1.1	27.2 ± 1.4	0.05
BOS width (cm)	24.2 ± 2.2	26.4 ± 1.3	0.05

SIP (single inverted pendulum) length = pelvic iliac crest height; BMI: body mass index; BOS: base of support, NS: statistically non-significant difference

Table 2. Values of the Pearson's linear correlation coefficient (r) between body anthropometry and the normalized COG sway parameters in the groups of young able-bodied subjects tested with eyes open and eyes closed.

Linear Correlation Coefficient (r)			VCOG Amplitude	VCOG Azimuth	DIAP	DIML
Body Height	F	EO	-	−0.36	−0.35	0.37
		EC	0.29	−0.33	−0.34	0.33
	M	EO	-	-	-	-
		EC	0.28	-	-	-
Base of Support Width	F	EO	-	0.43	0.43	−0.44
		EC	−0.25	0.46	0.45	−0.47
	M	EO	-	0.24	0.25	−0.25
		EC	-	0.23	0.23	−0.22
Base of Support Length	F	EO	0.20	−0.23	−0.23	0.23
		EC	0.26	−0.20	−0.21	0.20
	M	EO	0.24	-	-	-
		EC	0.26	-	-	-
Body Mass Index	F	EO	-	0.45	0.44	−0.44
		EC	-	0.42	0.42	−0.41
	M	EO	-	-	-	-
		EC	-	-	-	-

F: female; M: male; EO: eyes open; EC: eyes closed; VCOG amplitude and azimuth = sway vector parameters; DIAP: anteroposterior directional index; DIML: mediolateral directional index

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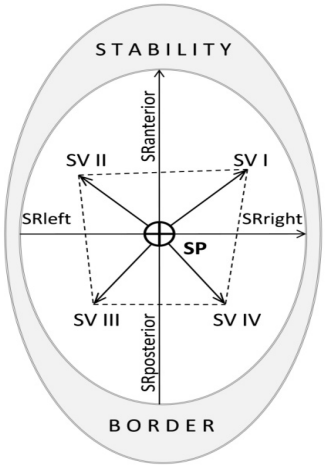


Figure 1. Stability vector amplitudes in young healthy male subjects standing quietly for 60 s with eyes open, presented in the Cartesian coordinate system. The set point of the COG controller is positioned in the origin of the coordinate system whereas the amplitudes of the COG sway (VC1–VC4) are shown as vectors in each quadrant. The vector amplitudes are equal to the mean COG velocity around the SC in the preferred directions marked by the vector azimuth. Please note sway slight asymmetry between anterior and posterior vectors while their vectorial sum is equal to zero, meaning that no drift of the setpoint in the control of standing posture in young healthy subjects. The distance between SC and the stability border (e.g., SR1 and SR2) represents the stability radius, and consequently, the VC-to-SR ratio might be a measure of postural stability.

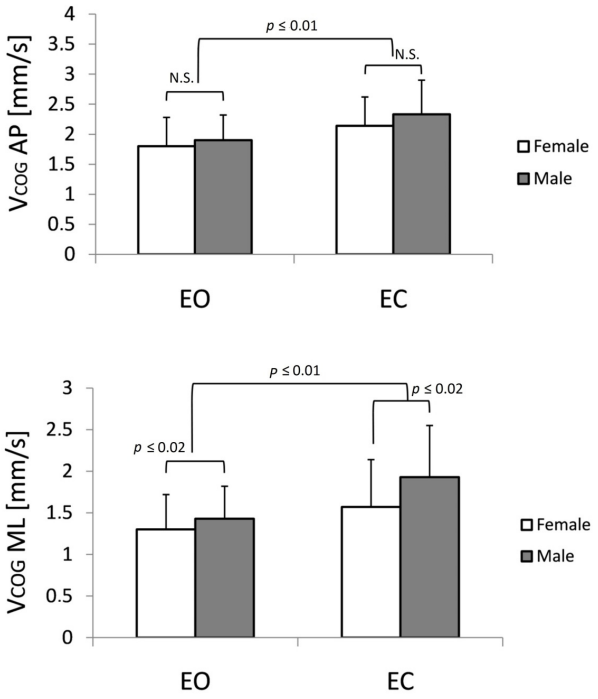


Figure 2. Mean COG velocities in anteroposterior (AP) and mediolateral (ML) directions in the male and female groups while tested with eyes open (EO) and eyes closed (EC).
N.S.: statistically non-significant

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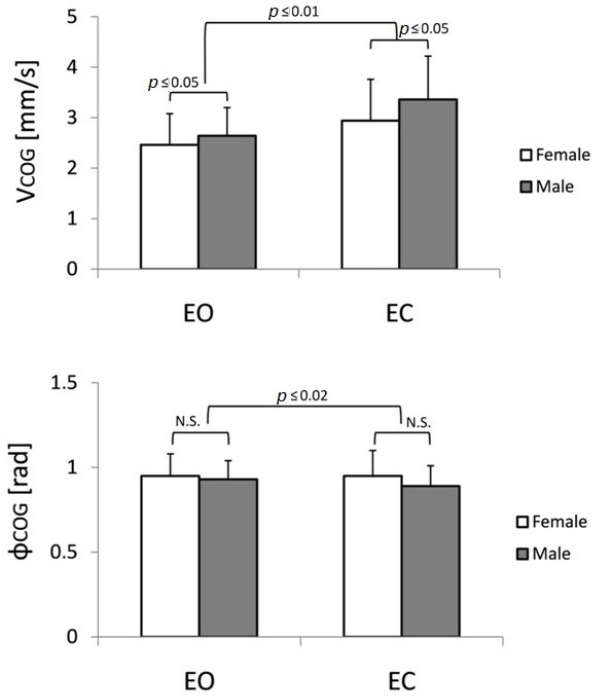


Figure 3. Mean values of COG sway vector parameters (vector amplitude = VCOG, vector azimuth = φCOG).
N.S.: statistically non-significant

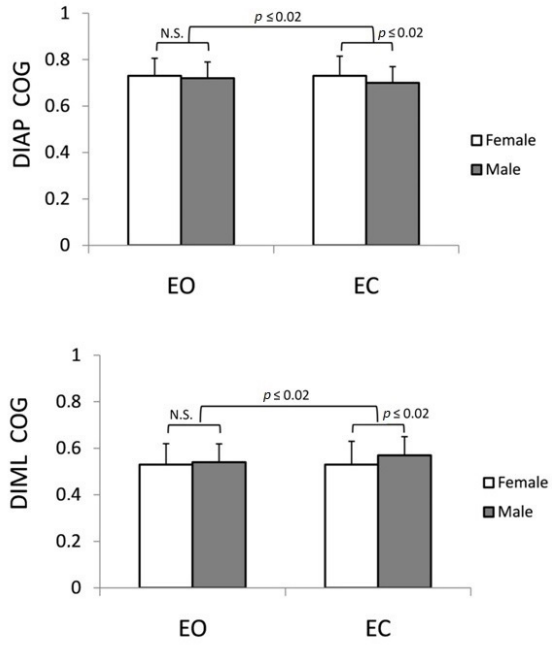


Figure 4. COG sway directional indices (anteroposterior = DIAP, mediolateral = DIML) in male and female subjects while standing still with eyes open (EO) and eyes closed (EC) for 60 s.
N.S.: statistically non-significant

The primary aim of this study was to assess characteristics of optimal postural stability control based on COG spontaneous oscillation characteristics. This allowed us to analyze the postural control process using the COG signal, which, in contrast to COP signals, is a poor output signal (Błaszczyk and Beck, 2023). Our results confirmed that during the quiet stance, the mean values and the COG peak-to-peak range of endogenous disturbances were very limited and negligible compared to the BOS size, which approximated the limits of postural stability. We estimated that in a quiet stance, the local COG oscillation range represented only 5% of the distance between the COG setpoint and the nearest stability border. Thus, these physiological COG oscillations around the set point did not compromise postural stability. Assuming that in young individuals the mean stability radius during the quiet stance is about 90 mm, and with local disturbance magnitude SV_m of 2.9–3.4 mm/s, we can estimate temporal stability limits ranging from 26 to 30 s. Within these temporal limits, the proprioceptive nervous system may efficiently detect and compensate for any setpoint disturbance. Since ankle proprioception may activate the ankle stabilizers within 100 ms, we can conclude that the temporal stability limits provide optimal control of standing posture.

The manipulation of the visual input is a commonly applied test for the assessment of postural control (Najafabadi et al., 2025). We noticed that, similarly to COP characteristics (Błaszczyk and Beck 2023), the exclusion of visual feedback resulted in a slight but statistically significant increase in the COG velocity. In both experimental groups, postural control without visual feedback was characterized by an unchanged COG position within the BOS, and only the COG sway range increased slightly. In the absence of vision, verticality is set based on vestibular input (Arntz et al., 2019; Dakin and Rosenberg, 2018), which adjusts accordingly the gain of proprioceptive feedback loops in the ankle joints (Merfeld et al., 1999; Proske, 2015).

The optimally stable erect posture may be characterized by the natural proportion between the AP and ML COG displacements, which can be assessed based on the DIAP and DIML, as well as the SV azimuth (Błaszczyk and Beck, 2023). Directional analysis of COG displacement during the quiet stance confirms that the movement is mainly in the AP plane. Analysis of the sway directional indices (DIAP and DIML) confirmed the well-known problem of bipedal stance control asymmetry (Anker et al., 2008; Błaszczyk et al., 2000; Dessery et al., 2011; Gutnik et al., 2008). Limb functional asymmetry and related COG displacement within the BOS may affect postural stability (Błaszczyk et al., 2000). A significant level of asymmetry in the quiet bipedal stance was documented in right-handed people (Gutnik et al., 2008). Importantly, with increasing weight-bearing asymmetry, the overall control of postural sway velocity increased mainly in the frontal plane with increasing weight-bearing asymmetry in favor of the more loaded leg (Anker et al., 2008). Consequently, during a double support stance, the main disturbances to the COG setpoint are limited to the sagittal plane which seems to trigger secondary oscillations in the frontal plane. The effect may be attributed to the difference in the ankle angle control due to limb preference and lateralization (Anker et al., 2008; Dakin and Rosenberg, 2018; Gutnik et al., 2008), which results in asynchronous activation of the left/right ankle stabilizing feedback loop.

There are only few studies that have systematically explored the relationship between biomechanical factors, such as body anthropometry and postural sway characteristics (Błaszczyk and Beck, 2023; Chiari et al., 2002; Hue et al., 2007; Ku et al., 2012). This knowledge, however, is indispensable to improving the reliability and diagnostic applicability of static posturography. We assumed that optimal postural control can be realized in the inverted pendulum model in a linear process and the validity of this hypothesis was verified based on the relationships of the body's COG random oscillations with individual anthropometric characteristics. Our previous studies on the effects of the BMI on postural sway showed a tendency towards greater COP sway in women compared to men (Błaszczyk and Beck, 2023). The present results documented a highly significant correlation between the mediolateral COG oscillations and the BMI under both EO and EC conditions, and the effect was solely observed in females. Additionally, a significant positive correlation between VCOG and body height was observed in both groups while standing with eyes closed. Positive significant

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correlations were also documented between the BOS width and the vector stability azimuth (φ) in both groups; however, the effect was more pronounced in females. This finding can be attributed to lower muscular forces in females, and ankle muscle strength is considered the most important factor of ankle stability and thus postural stability during an upright stance (Peng et al., 2024).

Conclusions

The results of the present research confirmed that in young, able-bodied individuals, during the quiet stance, the control of the COG position within the base of support and the range of random COG oscillations could be used as veridical indices of postural stability. Particularly, the steady COG set point without any drift during the trial strongly supported the notion of the most efficient postural control in young adults. The analysis of COG sway during the quiet stance seemed to be superior to the commonly used COP measurements. Its results well documented that in young healthy subjects, postural control is optimal and robust. Importantly, the characteristics of postural control are significantly dependent on gender and anthropomorphological factors, such as body height, the BMI, and the size of the BOS. Therefore, it is recommended to include the anthropometric measures for a reliable assessment of postural control using static posturography.

Author Contributions: Conceptualization: J.W.B.; methodology: J.W.B.; formal analysis: J.W.B. and M.B.; investigation: M.B.; data curation: J.W.B. and M.B.; writing—original draft preparation: J.W.B.; writing—review and editing: J.W.B.; supervision: J.W.B.; funding acquisition: M.B. Both authors have read and agreed to the published version of the manuscript.

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Funding information: This research was supported by statutory funds from the Jerzy Kukuczka Academy of Physical Education in Katowice.

Institutional Review Board Statement: This study was conducted following the principles of the Declaration of Helsinki, and approved by the Senate Ethics Committee of the Jerzy Kukuczka Academy of Physical Education in Katowice, Katowice, Poland (protocol code: 1/2011; approval date: 26 October 2011).

Informed Consent: Informed consent was obtained from all participants included in the study.

Conflict of interest statement: The authors declare no conflict of interest.

Acknowledgments: This research was a part of Monika Beck's Ph.D. thesis.

Received: 03 April 2025

Accepted: 15 June 2026

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