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*Title:* The trunk's contribution to postural control under challenging balance conditions

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**Youri Duchene:** Conceptualization, Methodology, Software, Formal analysis, Investigation, Writing - Original Draft, Visualization, Project administration. **Guillaume Mornieux:** Conceptualization, Methodology, Investigation, Writing - Original Draft, Visualization. **Arthur Petel:** Investigation, Writing – Reviewing and Editing. **Philippe Perrin:** Conceptualization, Writing – Reviewing and Editing, Resources. **Gérome Gauchard:** Conceptualization, Methodology, Writing - Original Draft, Resources, Supervision.

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**Abstract:**

Background: The double inverted pendulum model is imprecise when applied to studies of postural control. Although multijoint analyses have improved our understanding of how balance is maintained, the exact role of the trunk remains unclear.

Research questions: What is the trunk's contribution in postural control with respect to the other joints and how do trunk muscles control trunk kinematics?

Methods: Thirty-six healthy athletes (handball, karate, long jump) performed a highly challenging balance task while the ground support was dynamically tilted in the sagittal plane. The center of force (CoF) as well as lower limb joint angles and the trunk-pelvis angle were respectively measured with a force platform and inertial measurement units. The amplitude, sway path and standard deviation of the CoF and the joint angles were then calculated. Electromyography was used to record the activity of the rectus abdominis, external obliquus, and erector spinae muscles. Multiple linear regressions were computed to determine the joints' and muscles' contributions ( $\beta$ -coefficients) in predicting CoF variables and trunk kinematics, respectively.

Results: The linear combination of joint kinematic variables accounted for between 33% and 75% of the variance in the CoF. The ankle had the highest  $\beta$  and was a significant predictor of all CoF variables. The trunk yielded the second highest  $\beta$ -coefficient and was a significant predictor of the CoF sway path. Electromyography variables accounted for no more than 35% of the variance in the trunk kinematics, and erector spinae activity was the only significant predictor.

Significance: The trunk appears to be the second most important element during this specific postural task, in the magnitude of body sway in particular. But neuromuscular control of these trunk processes is difficult to characterize with surface electromyography only. The trunk should be taken into account when seeking to improve overall postural control (e.g. during training, rehabilitation).

**Keywords: Trunk control / Multijoint / Kinematics / Neuromuscular activity / Centre of force**

## 1. Introduction

Postural control is a complex process in which the vertical projection of the center of mass (CoM) must be kept within the base of support during quiet stance. Displacements of the center of foot force (CoF) are commonly computed to assess postural control [1]. Postural control has also been widely described in terms of joint kinematics [2,3]. Nashner & McCollum's double inverted pendulum is the classical model of postural control in the frontal and sagittal planes [4]. According to this model, postural control in the sagittal plane is first exerted around the ankle joint and then (if the postural challenge increases) by the hip joint [2,4].

However, it has also been shown that the knee joint influences postural control by lowering the CoM [5] and triggering compensatory postural strategies [6]; this challenges the classic paradigm of ankle-hip synergy [7]. Kilby et al.'s study of ankle, knee, hip and neck kinematics during balance tasks found that each of these joints had an active role in postural control. Moreover, an upper body strategy for maintaining balance during dynamic postural tasks has been described in the literature [8]. As the trunk is the most inertial segment of the body, it is not surprising that trunk control might be involved in postural control – especially during challenging postural tasks or dynamic situations [9]. Furthermore, the trunk might have an impact on balance by stabilizing the head and therefore the vestibular system [10]. For instance, it has been reported that postural performance during dynamic, one-leg stance was improved following specific neuromuscular training of the trunk in healthy individuals [11] and athletes [12]. It has also been reported that trunk sway during two-leg stance is greater for older adults [13,14] and patients with vestibular issues than in healthy people [15]. These findings emphasize the trunk's role in subjects known to have poor postural control. In the double inverted pendulum model, the hips and trunk are combined in a "hip strategy". However, Blackburn et al. [16] underlined the importance of distinguishing between the hips' and trunk's respective contributions to postural control; the researchers observed similar hip and trunk variances under several different postural control conditions and concluded that models of postural control should take

account of the trunk segment as well as the hip joint. Although the trunk clearly appears to have a role in postural control, its contribution (relative to lower limb joints) to this process has not been precisely determined.

With a view to understanding the trunk's involvement in postural control, the segment's neuromuscular activity has been investigated during external balance perturbations [2], a non-loaded, imposed posture [17] or a loaded, imposed posture [18]. To the best of our knowledge, Donath et al. are the only researchers to have studied the trunk's overall neuromuscular activity during an unrestrained, bipedal balance task [19]. They showed that the trunk's neuromuscular activity varies with the difficulty of the postural challenge and with age; the greater co-activation of trunk muscles in young adults (vs. older adults) was considered to be a means of stiffening the trunk and adjusting for postural imbalance not controlled by an ankle strategy alone. However, Donath et al. did not record the joint kinematics and so did not have a comprehensive vision of the influence of the trunk's neuromuscular activity on postural control in general and trunk kinematics in particular. We therefore decided to measure both neuromuscular and postural parameters, with a view to determining how neuromuscular control of the trunk is involved in postural control.

Hence, the main objectives of the present study were to (i) determine the trunk's contribution (relative to lower limb joints) to postural control during a challenging balance task and (ii) analyze the associated neuromuscular control processes. We hypothesized that in a challenging balance task, the trunk kinematic variables' contribution to CoF control would be at least as great as that of hip kinematic variables. Lastly, we explored the association between trunk muscle activation and trunk motion.

## **2. Materials and methods**

### *2.1. Participants*

Thirty-six healthy male adults participated in the study (mean (standard deviation (SD)) age: 23.3 (5.0); height: 1.83 (0.07) m; body weight = 79.4 (10.1) kg). This sample size was estimated based on a power analysis to achieve 80% power at an alpha criterion level of 0.05 during linear multiple regressions with four predictors analyses. All participants attended at least 3 training sessions a week and had over 10 years of experience in their respective sporting disciplines (handball, karate, and long jump). We chose these three sporting disciplines in order to study a broad range of postural control and footwork abilities. All participants reported that they had been free of lower limb or trunk injuries for the previous two months.

All participants gave their written informed consent prior to participation. The study was approved by the institutional review board (Comité de Protection des Personnes Sud Méditerranée III; reference: 2018.07.03 bis).

## *2.2. Material*

Participants stood on a dual platform that consisted of two connected footplates supported by four strain gauges mounted symmetrically on a supporting force platform (EquiTest®, Neurocom, Clackamas, OR, USA). Vertical force data were sampled at 100 Hz. Given that the experimental perturbations were applied in the sagittal plane, the CoF was calculated in the anteroposterior direction, as follows:  $CoF = \frac{((LF+RF)-(LR+RR))}{(LF+RF+LR+RR)} * 10.668$ , where left front (LF), right front (RF), left rear (LR) and right rear (RR) were the vertical forces and 10.668 (in cm) was the documented distance between the load cells and the center of the force platform.

Whole-body kinematics were measured using inertial measurement units (IMUs; CAPTIV, Tech Ergo Appliquées SARL, Vandoeuvre-lès-Nancy, France). The IMUs were placed on the left foot over the 3<sup>rd</sup> metatarsal, laterally on the middle of the left shank, laterally on the middle of the left thigh, between the posterior superior iliac spines, and on the 6<sup>th</sup> thoracic vertebrae. Before the static trial, the

magnetometer's reading was checked and set to zero with the participant in quiet stance. Data were collected with a sampling rate of 64 Hz and resampled at 1000 Hz. Quaternions were converted into angles by using the CAPTIV system's software. The system had a resolution of 0.1°.

Electromyographic activity (EMG) was recorded after the skin had been shaved and cleaned (baseline at rest:  $\leq 5 \mu\text{V}$ ). Wireless surface electrodes (Trigno™, Delsys, Natick, MA, USA) were attached parallel to the muscle fibers over the following muscles, bilaterally: the rectus abdominis (RA), the external obliquus (EOB) and the erector spinae (ES), in line with the SENIAM project's recommendations. RA electrodes were placed 2 cm laterally from the umbilicus, EO electrodes were placed at 15cm laterally from the umbilicus and ES electrodes were placed about 2 cm of the L1 spinous process.

### *2.3. Procedure*

Participants performed three different bipedal postural tasks on a force platform. They were asked to remain as still as possible with their knees slightly flexed and their arms at their sides, while looking straight ahead. The foot position was standardized. Firstly, the participants were asked to remain still for 40 seconds with their eyes open (EO) (the static trial). Secondly, two non-static, challenging balance tasks were performed under EO and then "eyes closed" (EC) conditions. The ground support (i.e. the force platform) was tilted in the sagittal plane by twice the estimated anteroposterior sway of the CoM [20] (Figure 1). For example, a 1° forward or backward sway in the CoM around the ankle resulted in a 2° platform tilt in the same direction. The angle between CoM and center of foot support is calculated by the Equitest® as follows:  $\theta = \arcsin (P_{\text{CoM}}/H_{\text{CoM}}) - 2.3$  where  $P_{\text{CoM}}$  is the distance between the center of force and the center of foot support,  $H_{\text{CoM}}$  is 55.27% of the person's height. The 2.3 degrees is the correction applied by the system to correct anterior position of the CoM. The participants performed a warm-up consisting in 10 trials with different degrees of postural control difficulty to get familiar to unusual perturbations. Each trial lasted 20 seconds with 30 seconds rest in-between. After the warm-

up, each participant performed two 20-second experimental trials (EO and then EC). Eight trials (7 during EC) were stopped and redone because subjects lost their balance and had to step off the platform.

#### *2.4. Data analysis*

The CoF's anterior-posterior maximum amplitude (CoF-AMP, in mm), sway path (CoF-SP, in mm) and SD (CoF-SD, in mm) were determined and provided respectively some information about postural accuracy, the magnitude of CoF displacement, and the variability of the postural control system [21,22]. We also calculated the AMP, SP and SD for the ankle, knee, hip and trunk-pelvis joints angles (in °). Data on kinetics and kinematics were low-pass filtered with a 15Hz fourth-order Butterworth filter.

EMG data were sampled at 1000 Hz and band-pass filtered from 10Hz to 500 Hz. For the RA and EOB signals, the heart's electrical activity was filtered out with a 30 Hz Butterworth high-pass filter. The root mean square (RMS) EMG amplitudes were computed and normalized against the value from the static trial. Lastly, the left and right normalized EMG-RMSs were averaged.

We analyzed the data recorded during the middle 15 seconds of each trial. Each dependent variable was averaged over the two trials. Data were analyzed using MATLAB software (version 9.2, MathWorks, Inc., Natick, MA, USA).

#### *2.5. Statistical analysis*

Means and SDs were calculated for all measured variables. To determine the respective contribution ( $\beta$ -weight coefficients) of joint kinematics and EMG-RMS in postural control and trunk control, we performed multiple linear regression analyses of (i) joint kinematic variables as predictors of CoF



variables, and (ii) EMG-RMSs as predictors of trunk kinematics. To check for multicollinearity, we examined the variance inflation factor (VIF) for each regression model. Normal distributions of quantitative data and residuals were checked with a quantile-quantile plot. Statistical analyses were performed with Statistica software (version 13, StatSoft, Inc., Tulsa, OK). The threshold for statistical significance was set to  $p < 0.05$ .

### **3. Results**

The values of CoF-AMP, CoF-SP, and CoF-SD were significantly higher in the EC condition than in the EO condition (Figure 2).

Likewise, joint kinematics were influenced by the postural task. For the ankle, knee, hip, and trunk, the respective AMP, SP and SD values were significantly higher in the EC condition than in the EO condition ( $p < 0.001$  for all; see Figure 3 for the trunk).

Furthermore, the trunk muscles were significantly more activated in the EC condition than in the EO condition ( $p < 0.001$ ). The EC - EO difference (in %, relative to the static condition) was greater than 50% for the EOB ( $108 \pm 55\%$  for EC vs.  $51 \pm 17\%$  for EO) and ES ( $181 \pm 93\%$  vs.  $115 \pm 46\%$ , respectively) and close to 100% for the RA ( $150 \pm 140\%$  vs.  $53 \pm 29\%$ , respectively).

#### *3.1. The trunk relative weight in postural control*

In all the regression models, the VIF was always below 5.0; we were therefore confident that multicollinearity was not a significant confounding factor. The results of the multiple linear regressions (i.e. joint kinematic variables as predictors of CoF variables) are summarized in Table 1. In the EO condition, the linear combination of joint kinematic variables accounted for between 60% and 75% of

the variance ( $R^2$ : 0.60–0.75;  $p < 0.001$ ) in the corresponding CoF variable. In the EC condition, the joint kinematic variables accounted for between 33% and 57% of the variance ( $R^2$ : 0.33–0.57;  $p < 0.05$ ).

In the EO condition,  $\beta$ -ankle was a significant predictor of all the CoF variables, and  $\beta$ -trunk was a significant predictor of CoF-SP ( $p \leq 0.05$ ).  $\beta$ -Ankle had the highest values (0.70 – 0.80), followed by  $\beta$ -trunk (0.08 – 0.27),  $\beta$ -knee (0.04 – 0.15), and  $\beta$ -hip (0.03 – 0.11). In the EC condition, only  $\beta$ -ankle was a significant predictor of CoF-SP and CoF-SD. Again,  $\beta$ -ankle had the highest values for each CoF variable, followed by  $\beta$ -hip (the second best predictor of CoF-AMP and CoF-SD) and  $\beta$ -trunk (the second best predictor of CoF-SP).

### *3.2. Trunk neuromuscular control*

In all the regression models of the EMG data, the VIF was always below 5.0. The EMG variables did not significantly account for trunk kinematics under the EO condition. However, the linear combination of the EMG variables accounted for between 21% and 35% of trunk kinematics variance in the EC condition ( $R^2$ : 0.21 – 0.35;  $p \leq 0.05$ ). For each kinematic variable, the ES had the highest  $\beta$ -coefficient (0.44 – 0.57), followed by the EOB (0.10 – 0.13) and the RA (-0.05 – -0.01).  $\beta$ -ES was the only significant predictor of trunk kinematic variables.

## **4. Discussion**

The present study's objectives were to determine the weight of trunk (relative to the lower limb joints) in postural control in athletes performing a challenging balance task and to analyze the neuromuscular control of the trunk. Our main findings were that (i) the trunk's contribution to postural control is at least as important as the hip's in the present task, and (ii) trunk motion during postural control seems to involve more the ES muscles than the surface abdominals.

Here, a challenging postural task was carried out on a tilting platform under EO and EC conditions. The greater difficulty under EC conditions (vs. EO) was evidenced by the significantly higher values of the CoF variables. Likewise, trunk motion and neuromuscular activity were higher under EC conditions. Our results confirm Gill et al.'s report [13] of greater trunk angular sway during more difficult postural tasks.

As the ankle had the highest  $\beta$ -coefficient of the four joints, our results confirm the well-known major role of the ankle in postural control [23,24]. In the EO condition, 61% of the variance in CoF-AMP (i.e. postural accuracy) was accounted for by the amplitude of the joint kinematics. The trunk had an important, corrective role ( $\beta$ -trunk: -0.18) in postural accuracy. In the EC condition, the joint kinematics only accounted for 33% of the variance in CoF-AMP. This low proportion might be due (at least in part) to the inclusion of extreme values in our calculation of the amplitude. Privation of vision might further increase the proportion of extreme values if the "safety zone" is explored so as to better estimate the CoM's position [25]. Moreover, the joints' outlier values might not be time-synchronized with those of the CoF, as a result of antiphase joint patterns. The CoF-SP (i.e. the magnitude of CoF displacement) corresponds to the ability to stand as still as possible (i.e. steadiness) [22]. In the EO condition, 75% of variance in the CoF-SP was accounted for by the amplitude of joint displacement. The ankle was the most important joint in this respect, and was followed by the trunk. The knee and hip joints made rather small contributions. Interestingly, the trunk was the second-largest contributor factor in the EC condition - demonstrating that body steadiness is explained (at least in part) by the amount of trunk displacement. Our results underline the importance of trunk steadiness and thus head-trunk coordination, which would provide an effective support for the vestibular system [10,26] and help to limit CoM accelerations [24]. CoF-SD was mostly accounted for by the ankle joint's SD. Surprisingly, the trunk did not appear to have an impact on CoF-SD. In view of the trunk's inertia, fast angular corrections through the ankle would certainly be more effective and would decrease the trunk's

contribution to regulation of CoF-SD. The predominance of ankle variance over trunk variance under EC conditions has already been described in the literature [25].

Overall, we found that the ankle was the main joint involved in posture control under challenging balance conditions. Although this was not unexpected (given the extensive literature [23]), our results in a platform tilt paradigm further confirm the importance of this joint in balance. It is noteworthy that the trunk was more important than the hip in controlling postural accuracy and steadiness; this highlights the likely value of performing separate kinematic analyses of the trunk and hip [16]. The knee joint contributed less than other joints, indicating that the primary strategy used by the knee to lower the CoM might not be essential under challenging balance conditions. Overall, a multijoint analysis is likely to be more valuable than conventional single or double inverted pendulum models for understanding postural control. This suggestion is in line with Hsu et al.'s report [27] that a six-joint model accounted for CoM variance better than the conventional inverted pendulum models did.

Although joint kinematics is the key factor for analyzing postural coordination, the present study's data on trunk neuromuscular activity provide a deeper understanding of the trunk's influence on postural control. In the EC condition, between 21% and 35% of the variance in trunk kinematic variables was accounted for by the three trunk muscles measured in the present study. The ES muscles made the largest contribution. With respect to anteroposterior body sway, it could be suggested that the ES may control trunk extension and flexion through concentric and eccentric contraction, respectively. In our task, the ES appeared to control the trunk's anteroposterior motion and the RA and EOB appeared to help stiffen the spine; this hypothesis might explain (at least in part) the low proportion of trunk motion variance accounted for by these muscles, and is in agreement with Stanton & Kawchuk's suggestion [28] that the RA and EOB help to stabilize the spine. Interestingly, EMG activity for the abdominal muscles was lower during the EO condition than during the static trial. As ES are possibly the main muscles controlling anterior-posterior trunk tilt, RA and EOB might decrease their activity to reduce spine stiffness and therefore allow faster and more accurate trunk displacements during a challenging

balance task. Furthermore, the analysis of “local” muscles (such as the transversus abdominis and the internal obliquus) might provide a more accurate view of trunk neuromuscular activity during postural control [19]. Lastly, the low percentage of variance accounted for by muscle activity might be due to inter-individual variability in muscle coordination and the intensity of trunk muscle contraction needed to stiffen the spine [29]. Difficulty in explaining trunk kinematics with EMG measurement has been reported for other tasks (such as changes of direction) in which neuromuscular activity was not associated with the trunk angle at initial foot-ground contact [30]. Trunk muscle activity did not account for trunk motion in the EO condition. This could be explained by limited trunk motion with respect to the pelvis with the eyes open (e.g. AMP=2°). Indeed, the perturbations caused by platform movements induce displacements of the trunk; these displacements might be regulated by the trunk muscles themselves or by lower leg joints action that stabilize the trunk. The lack of correlation between kinematics and muscle activation observed in the present study suggests that trunk motion is controlled by the lower limb joints and that trunk muscle activation stiffens the spine and helps to maintain balance [19].

We acknowledge several limitations in this study. First of all, different balance tasks induce different skills in order to maintain balance [31]. Therefore, the present results cannot be generalized to each challenging balance task and every healthy or active population. However, this study helps to underline the importance of the trunk in a challenging situation, confirming literature on other tasks [16]. Then, all trunk and pelvis muscular activity could not be taken into account, as only main surface muscles connecting thorax or spine to the pelvis were measured. Moreover, the modelling of the trunk as a rigid segment might not accurately reflect its kinematics [32], blurring the links between muscle activations and spine kinematics. Although the Captiv IMU system has been validated in the literature [33], the use of this methodology might reduce the accuracy of joint angles measurements compared to gold standard motion capture systems.

## **5. Conclusion**

When compared with the double inverted pendulum model, a multijoint analysis provided a better understanding of postural control. We found that during a highly challenging balance task, the trunk was the second most important segment or joint (after the ankle) in the control of postural accuracy and steadiness. Although the trunk muscles' EMG activity accounted for a relatively small proportion of the variance in trunk motion, our present results demonstrated that the ES seems to have an important role in controlling trunk anteroposterior motion, while the RA and EOB might tend to stiffen the spine.

### *Conflict of interest*

The authors declare that they have no conflict of interest.

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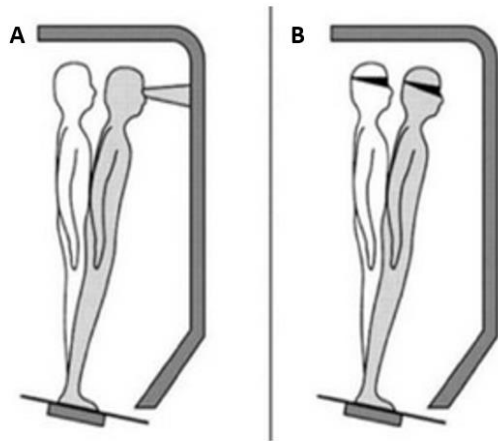


Figure 1: Representation of the balance task with the tilting platform in (A) the “eyes open” (EO) and (B) eyes closed (EC) conditions (EquiTest®, Neurocom, Clackamas, OR, USA).

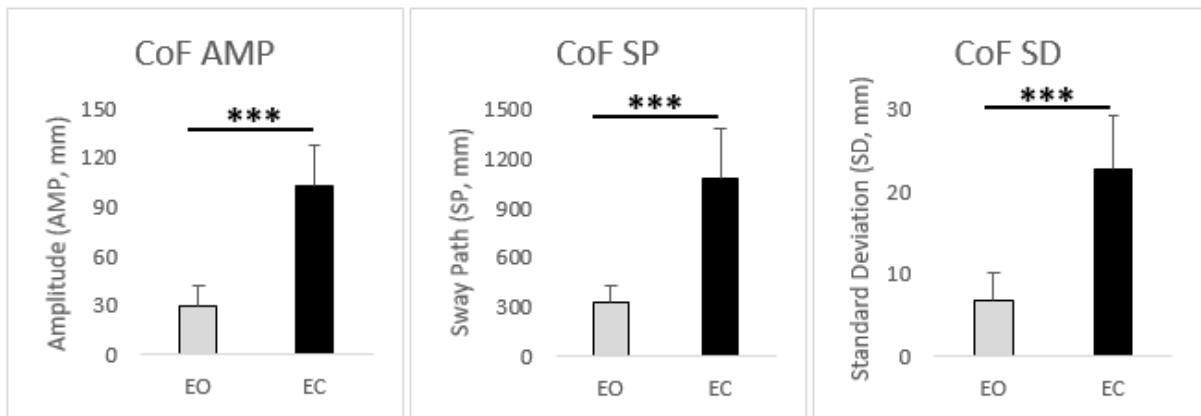


Figure 2: CoF amplitude (AMP), sway path (SP) and standard deviation (SD) under “eyes open” (EO) and “eyes closed” (EC) conditions.

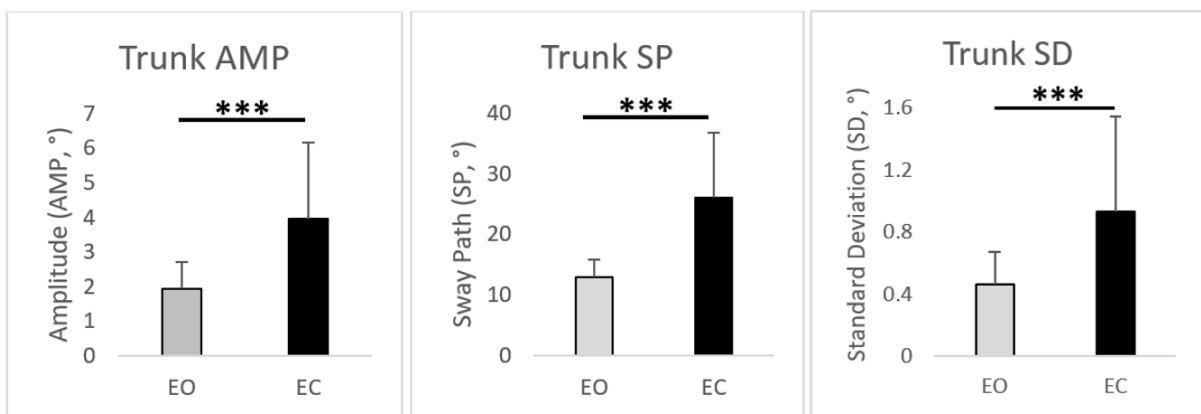


Figure 3: Trunk amplitude (AMP), sway path (SP), and standard deviation (SD) under “eyes open” (EO) and “eyes closed” (EC) conditions.

|                      | AMP    |        | SP    |        | SD     |       |
|----------------------|--------|--------|-------|--------|--------|-------|
|                      | EO     | EC     | EO    | EC     | EO     | EC    |
| <b>R<sup>2</sup></b> | 0.61   | 0.33   | 0.75  | 0.57   | 0.60   | 0.43  |
| <b>F</b>             | 12.0   | 3.85   | 23.8  | 10.1   | 11.5   | 5.91  |
| <b>p</b>             | <0.001 | <0.001 | 0.012 | <0.001 | <0.001 | 0.001 |
| <b>β-Ankle</b>       | 0.70*  | 0.34   | 0.72* | 0.50*  | 0.80*  | 0.59* |
| <b>β-Knee</b>        | 0.15   | 0.02   | 0.06  | 0.16   | 0.04   | -0.08 |
| <b>β-Hip</b>         | 0.11   | 0.21   | -0.04 | -0.06  | -0.03  | 0.15  |
| <b>β-Trunk</b>       | -0.18  | 0.11   | 0.27* | 0.27   | -0.08  | 0.03  |

Table 1: Results of the multiple linear regressions for each kinematic variable (amplitude (AMP), sway path (SP), and standard deviation (SD)) under “eyes open” (EO) and “eyes closed” (EC) conditions. R<sup>2</sup> = coefficient of determination, F= Fisher distribution, \*p≤0.05