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Dynamics of postural control in individuals with ankle instability: Effect of visual input and orthotic use

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ABSTRACT

The present study investigated the effect of visual deprivation and the use of an ankle orthosis on the dynamics of center of pressure (COP) trajectories during unilateral stance in individuals with ankle instability (AI). Sixteen individuals with AI and nine healthy individuals performed four trials of 30s unilateral stance on a force platform with 1) eyes open wearing no orthosis, 2) eyes closed wearing no orthosis, 3) eyes open wearing orthosis, and 4) eyes closed wearing orthosis. The anterior-posterior and mediolateral COP trajectory were extracted. Regularity was quantified by sample entropy, dimensionality was quantified by correlation dimension and level of time dependency was quantified by entropic half-life. The AI individuals had lower sample entropy and longer entropic half-life in their COP trajectories. The effect of visual deprivation did not differ between groups. Wearing an ankle orthosis increased the sample entropy in the anterior-posterior direction and decreased the correlation dimension in the mediolateral direction for the AI group only. Individuals with AI have higher COP trajectory regularity and higher level of time dependency compared to healthy individuals. Additionally, individuals with AI do not alter the dynamics of their postural control during unilateral stance with visual deprivation compared to healthy individuals. This suggests that alterations in visual or somatosensory information differently affect the executed postural movement pattern. Finally, wearing an orthosis significantly alters the COP dynamics of individuals with AI.

1. Introduction

Ankle sprain is one of the most common sport injuries and previous ankle sprain is the strongest predictor of future reinjury [1,2]. Recurring ankle sprains and the subjective feeling of the joint "giving way" during loaded conditions is termed chronic ankle instability, which is related to both functional and mechanical instability of the ankle [3,4]. This condition has been associated with postural control deficits during unilateral stance on the affected ankle due to damaged sensory receptors in the injured ligaments [5,6].

Time-to-boundary measures (e.g. comparisons of the instantaneous center of pressure (COP) velocity and position) have revealed that individuals with chronic ankle instability applied a movement strategy during unilateral stance that placed them close in time to episodes of potential loss of postural control (i.e. moving the COP rapidly close to the boundary of support) [7–10]. Thus, individuals with chronic ankle instability have less time to adjust postural movements due to high COP velocity and/or COP position close to the boundary of support, which has been interpreted as a sign of postural instability [11,12]. However, while including spatiotemporal information from the COP displacement and velocity, these measures do not account for the dynamics in the COP trajectories. This quality has been assessed in several studies aiming to investigate the impairments of postural control following aging or neurological disorders by applying nonlinear methods to the COP displacements [13–20]. These methods include sample entropy (SaEn) which is a measure of regularity based on the number of repetitive vectors of consecutive points in the time series in question [21], and correlation dimension (CoD), which quantifies the fractal dimensionality of the time series in state space [22]. Both measures have been used to assess the underlying motor control strategy [23,24]. Additionally, entropic half-life (ENT½) was recently introduced as a measure of the level of time dependency. The measure estimates the elapsed time before positional information from previous completed movements no longer influences the control of current movements [25,26]. These methods have previously been found to be superior to traditional linear tools when used to investigate the effect of concussion in athletes [27], infant motor control development [28,29] and balance changes with ageing [16]. To the best of our knowledge, using multiple nonlinear tools to assess the dynamics of the COP displacement has not been done in individuals with chronic ankle instability and could reveal

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insights to the underlying motor control [24].

Recently, a meta-analysis by Song and colleagues concluded that individuals with chronic ankle instability rely less on somatosensory information and more on visual information to maintain postural control during unilateral stance [30]. A possible explanation could be that repeated ankle sprains damage the mechanoreceptors of the collateral ligaments, which increases the reliance of visual input [30]. This altered sensory input potentially changes the dynamics of the movement strategy and increases the postural instability. However, wearing an ankle orthosis in individuals with chronic ankle instability has been shown to improve the postural control during unilateral stance by compensating for both the mechanical instability and the limited somatosensory information in the ankle joint [31–34]. It is unknown how wearing an ankle orthosis affects the dynamics of the COP trajectory in individuals with a history of lateral ankle sprains and a subjective feeling of ankle instability (AI). Thus, the purpose of the present study was to investigate the dynamics of postural control in individuals with AI during unilateral stance with and without visual input and with and without an ankle brace. To fulfil this purpose, the present study assessed the dynamics in terms of regularity as quantified by SaEn [21], dimensionality as quantified by CoD [22] and the level of time dependency as quantified by ENT½ [26,35] of COP trajectory in individuals with a history of lateral ankle sprains and a subjective feeling of AI during unilateral stance in the previously mentioned conditions. Assuming that postural instability would be characterized by more random COP movements, we formulated three hypotheses:

1) Individuals with AI would exhibit less regularity (higher SaEn), higher dimensionality (higher CoD) and lower level of time dependency (lower ENT½) of their COP trajectory during unsupported unilateral stance compared to healthy controls.
2) Individuals with AI would exhibit a larger decrease in regularity (increased SaEn), larger increase in dimensionality (increased CoD) and larger decrease in level of time dependency (decreased ENT½) of their COP trajectory when deprived of visual information during unilateral stance compared to healthy controls.
3) Wearing an ankle orthosis during unilateral stance with visual input will significantly increase the regularity (decreased SaEn), reduce the dimensionality (decreased CoD) and increase the level of time dependency (increased ENT½) of the COP trajectory for both individuals with AI and healthy controls compared to the corresponding trial without an ankle orthosis.

2. Methods

2.1. Participants

Sixteen individuals with a history of lateral ankle sprains (males/females: 9/7, mean ± SD age: 30.9 ± 4.7 yrs, body mass: 73.4 ± 11.9 kg, body height: 176.4 ± 9.5 cm and body mass index: 23.5 ± 2.9 kg/m^2) were included in the AI group. Inclusion criteria were: a history of unilaterally recurring ankle sprain following an initial sprain (at least 1 year from initial acute injury), at least 3 months since last ankle sprain, regular athletic activity (at least 3 h weekly), 25–40 years of age, and no regular use of an external ankle support. Exclusion criteria were: the presence of either a neurological disease that affects movement control of the lower limbs or a vestibular condition that affects balance. For the included AI participants, the initial sprain was characterized by a doctor's diagnosis with subsequent conservative treatment with a brace and low weight bearing for 2–4 weeks after the injury. The participants reported to still have a sense of “giving way” feeling on occasion but maintained regular sports participation. Nine healthy individuals (males/females: 7/2, mean ± SD age: 29.3 ± 4.5 years, body mass: 72.5 ± 11.3 kg, body height: 178.6 ± 11.5 cm and body mass index: 22.6 ± 1.9) were included in the control group (CON). All participants gave their written consent to participate after being previously informed of the experimental conditions. The study was approved by the local ethical committee and carried out in accordance with the recommendation in the Declaration of Helsinki.

Fig. 1. Orthotic used for testing within this study, with the embedded energy absorbing system marked in dashed lines and the direction of restraint during sprain-like movements indicated with an arrow.

2.2. Experimental setup

The participants wore their own training shoes and completed four 30-s trials of shod unilateral stance on a force platform (BP400600, AMTI, Watertown, MA, USA) in the following order: 1) eyes open wearing no orthosis (EONO), 2) eyes closed wearing no orthosis (ECNO), 3) eyes open wearing ankle orthosis (EOAO), and 4) eyes closed wearing ankle orthosis (ECAO). During each trial, the AI group stood on their affected leg and the CON group stood on their preferred leg. The orthotic used for testing consisted of a plastic cuff connected to an insole with an unpowered energy-absorbing system, in which only large frontal plane accelerations are restricted via a dilatant fluid module (Fig. 1) (Betterguards GmbH, Berlin, Germany). All participants were instructed to keep their hands on their hips and the contralateral leg lifted without resting it on the standing leg. During the trials with open eyes, the participants were instructed to keep their gaze fixed on a target placed on the wall approximately 5 m in front of them. The trials were separated by a minimum of 2 min rest to avoid fatigue interfering with performance. During each trial, three-dimensional ground reaction forces and moments were recorded at 1000 Hz. The signals were AD-converted (resolution: 16-bit) and regulated through the AD-card within a VICON Giganet Box VICON, Oxford, UK) with a 10 V DC-coupled setup including both dynamic and static components of the force signal.

2.3. Analysis

For each trial, the anterior-posterior (AP) and mediolateral (ML) COP time series were extracted from 20 s after removing the initial 10 s (Fig. 2). Similarly to previous studies, the COP time series were filtered using a Daubechies wavelet (decomposition at level 5 using 5 db) and down-sampled to 100 Hz before further analysis [16,25,35].

Regularity was quantified by SaEn using the equation presented by Richman and Moorman [21] (equation (1)).

$$SaEn(m, r, N) = -\log \frac{A^{m+1}(R)}{B^{m}(R)} \tag{1}$$

Where A is the number of similar vector lengths $(m+1)$ falling within R

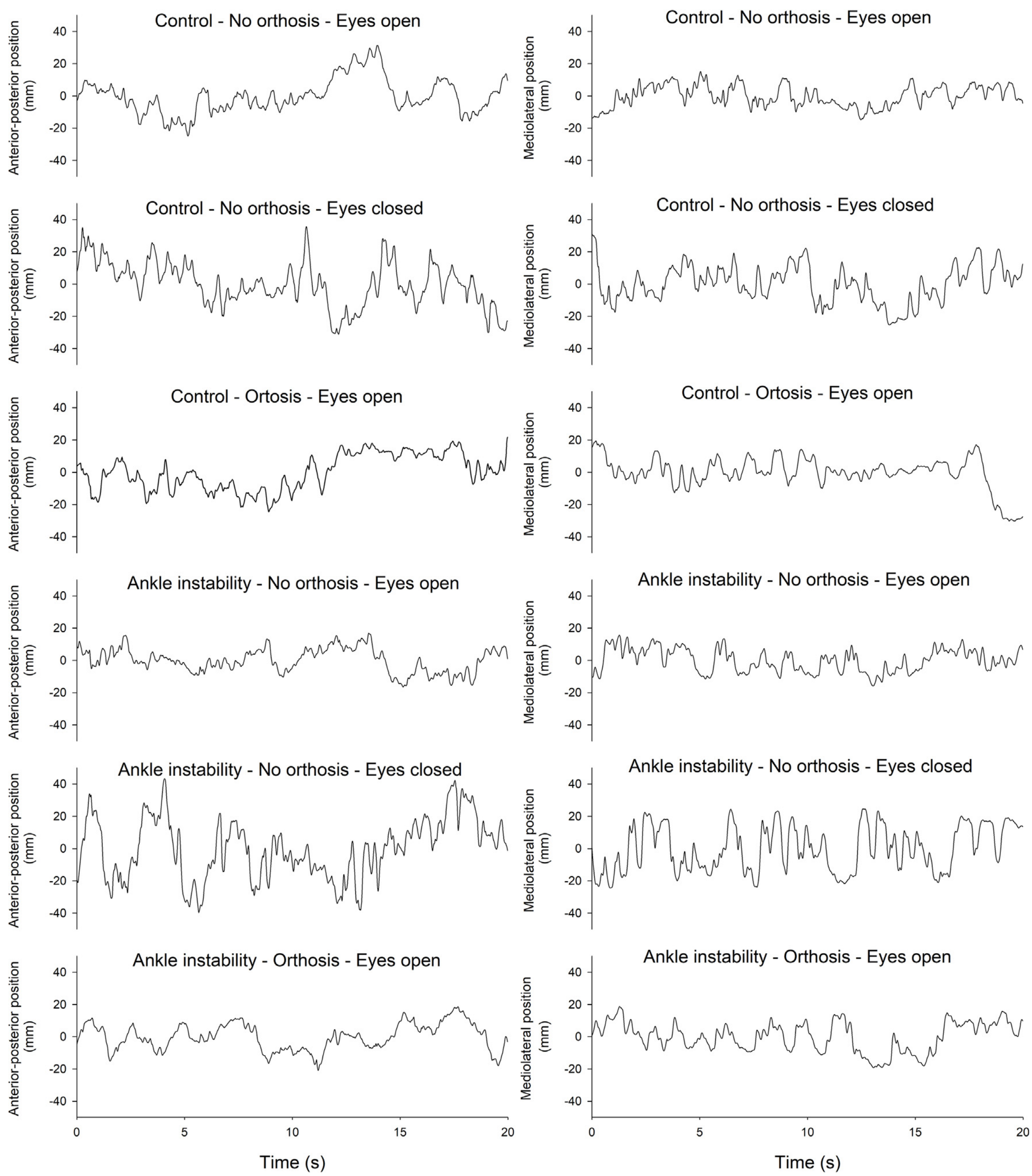

Fig. 2. COP trajectories in AP (left graphs) and ML (right graphs) directions for one individual from the CON group (top three graphs) and one individual from the ankle instability group (bottom three graphs) during the eyes open wearing no orthosis, eyes closed wearing no orthosis and eyes open wearing orthosis trials.

the tolerance limit (equation (2)), B is the number of similar vector lengths (m) falling within the tolerance limit and N is the number of data points.

$$Tolerance\ limit = r \cdot standard\ deviation_{time\ series} \quad (2)$$

To investigate the effect of input parameter choice of m and r, SaEn was calculated using $m = 2$, and $m = 3$ and $r = 0.1$, 0.15, 0.2, 0.25, and 0.3 [36]. The results from SaEn calculated with $m = 2$ and $r = 0.2$ is presented below and the remaining results are presented in the supplementary material. Low SaEn values indicate high regularity and less randomness in the COP displacement and high SaEn values indicate low regularity and more randomness in the COP displacement.

Dimensionality was quantified by CoD using the equation presented by Grassberger and Procaccia [22] (equations (3) and (4)). CoD estimates the spatial correlations between pairs of phase points on the given trajectory attractor in an m-dimensional phase space.

$$D_k = \lim_{t \to 0} \frac{\ln C(t)}{\ln(t)} \quad (3)$$

$$C(r) = \lim_{L \to \infty} \left[\frac{1}{L^2} \sum_{i=1}^{L} \sum_{j=1}^{L} H(t - |x_i - x_j|) \right] \quad (4)$$

Where L is the total number of phase points, H is the Heaviside function, t is the tolerance distance (radius of the sphere), X_k is the kth phase point in the m-dimensional embedded space (k varies from 1 to L), and $|X_i - X_j|$ is the Euclidian norm of the two phase points. Before calculating CoD, state space was reconstructed using the method of delayed embedding [37–39]. Time delay and embedding dimension were calculated using the Average Mutual Information and False Nearest Neighbor algorithms. The time delay and embedding dimension of each subject during each trial are presented in the supplementary material.

The mean time delay and embedding dimension across trials and subjects were: 15.5 ± 3.3 and 4.8 ± 0.7 for the AP direction and 13.4 ± 2.9 and 5.0 ± 0.5 for the ML direction, respectively. To best represent all investigated time series, time delays of 15 and 13 and embedding dimensions of 5 and 5 were chosen for the AP and ML direction, respectively. CoD was then calculated from the time series reconstructed in state space with lower CoD values indicating a lower dimensionality and higher CoD values indicating a higher dimensionality [22].

The level of time dependency was quantified by ENT½ using the procedure described by Zandiyeh and von Tscharner [26] and Baltich and colleagues [35]. ENT½ is based on calculations of SaEn on consecutive gradual randomizations of the COP time series. The gradual randomization was achieved through successive reshaping procedures of the original time series according to the following principle: Assuming that the first reshaped time series equals the original time series [1 2 3 4 5 6 7 8 9 10 11 12], the second reshaped time series equals [1 3 5 7 9 11 2 4 6 8 10 12], and the third equals [1 4 7 10 2 5 8 11 3 6 9 12]. With the a sampling frequency of 100 Hz, the time between two adjacent data points of the original time series was 10 ms, for the second reshaped time series the time between two adjacent data points was 20 ms, and for the third reshaped time series the time was 30 ms and so on. This reshaping procedure was iterated 150 times and SaEn was calculated for each of them using $m = 2$ and $r = 0.2$. The SaEn from each reshaped time series were normalized according to the following equation:

$$Normalized\ SaEn = \frac{SaEn_{RS} - SaEn_{OR}}{SaEn_{RAN} - SaEn_{OR}}$$

Where $SaEn_{RS}$ is the SaEn of the reshaped time series, $SaEn_{OR}$ is the SaEn of the original time series, $SaEn_{RAN}$ is the average of the SaEn from 50 completely randomized time series. The randomized time series were created by a random permutation of the data points in the original time series. Finally, the normalized SaEn values were plotted in a semi logarithmic plot as a function of the reshaped time (increasing with 10 ms for each new reshaped time series). ENT½ was identified as the time at which the normalized SaEn increased above 0.5 [26]. The higher the ENT½, the longer the elapsed time before previous COP movements no longer influence current COP movements [25,35]. All calculations were performed in Matlab (R2017b).

2.4. Statistics

Residual normality and data homogeneity were confirmed using Shapiro-Wilk test and Levene's test of equality before conducting the following statistical analysis. To test the first hypothesis, differences between groups in the SaEn, CoD, and ENT½ during EONO unilateral stance trial were established by applying a two-way mixed design ANOVA with group (between-subjects), COP direction (within-subjects) and the group-direction interaction as independent variables. To test the second hypothesis, the change in each variable between EONO and ECNO was calculated, and group differences in the change were established by applying a two-way ANOVA with group, COP direction and the group-COP direction interaction as independent variables. To test the third hypothesis, the differences between EONO and EOWO in the SaEn, CoD, and ENT½ were established by applying a two-way ANOVA for repeated measures with trial, COP direction and the trial-direction interaction as independent variables for each group. Level of significance was set at 5%. All statistical analyses were performed in SPSS (IBM, SPSS Statistics, version 24, NY, USA).

3. Results

The overall effects and interactions of the performed two-way ANOVAs are presented in Table 1. The results of the post hoc tests are described below. The SaEn, CoD and ENT½ of the COP trajectories in the AP and ML direction for the AI and CON groups during the EONO trail are presented in Fig. 3. The AI group exhibited significantly lower SaEn and longer ENT½ compared to the CON group and the SaEn was significantly higher in ML direction (Fig. 3A and C). The change in SaEn, CoD and ENT½ of the COP trajectories in the AP and ML direction for the AI and CON groups from the EONO trial to the ECNO trial are presented in Fig. 4. While the SaEn increased from the EONO trial to the ECNO trial in the AP direction, it decreased in the ML direction (Fig. 4A). The SaEn, CoD and ENT½ of the COP trajectories in the AP and ML direction for both investigated groups during the EONO and EOWO trails are presented in Fig. 5. For the AI group, the SaEn in the AP direction was significantly higher when wearing a orthosis ($p = 0.049$) (Fig. 5A) and for the CON group the SaEn during the trial without the orthosis was significantly higher in the ML direction compared to the AP direction ($p < 0.001$) (Fig. 5D). The CoD in the ML direction of the AI group was significantly lower when wearing an orthosis ($p = 0.007$) (Fig. 5B).

4. Discussion

The present study aimed to investigate the dynamics of postural control in individuals with AI during unilateral stance. Additionally, the effects of visual input and the mechanical support of an ankle brace were investigated. To the best of our knowledge, the present study is the first to address this topic through the use of multiple nonlinear methods to quantify the characteristics of the COP trajectories. In relation to the aim of the present study, three hypotheses were formulated.

The first hypothesis stated that individuals with AI would exhibit less regularity (higher SaEn), greater dimensionality (higher CoD) and lower level of time dependency (higher ENT½) of their COP trajectory during unilateral stance compared to healthy controls. In contrast to this hypothesis, we observed lower SaEn and higher ENT½ in the AI group compared to the CON group. This suggests that the motor control strategy exhibited by the AI group in order to maintain stable unilateral balance is characterized by a more regular pattern with a higher level of time dependency (i.e. more time elapses before position information from previous completed movements no longer influences the control of current movements). While this appears counterintuitive, since AI individuals presumably have more passive ankle laxity [40], it could be explained by altered muscle activity with increased co-contraction on both sides of the ankle joint, leading to a more rigid and regular movement pattern [6]. Furthermore, the higher level of time dependency in the AI individuals suggests that their motor control strategy is influenced by movements performed further back in time compared to that of the CON individuals [25,35]. This could indicate that the AI individuals are less able to adjust their postural motor control quickly in response to external perturbations. The CoD has previously been interpreted as a measure of the organization of movement solution space with higher values indicating a less structured space with more potential movement solutions, and lower values indicating a more structured space with fewer potential movement solutions [41,42]. Our results suggest that both the AI and CON individuals have a similar organized movement solution space during unilateral stance.

The second hypothesis stated that the AI group would exhibit a larger decrease in regularity (increased SaEn), larger increase in dimensionality (increased CoD) and larger decrease in level of time dependency (decreased ENT½) when deprived of visual information during unilateral stance compared to the CON group. This hypothesis could not be confirmed as no group effect was observed on the change in the investigated variables from the EONO trial to the ECNO trial. Thus, the two groups reacted similarly when visual input was removed. This is in contrast to the conclusion from a recent meta-analysis using time-to-boundary as the outcome measure, which suggested that individuals with chronic ankle instability rely more on visual information

Table 1
Outcome of two-way ANOVA tests for the three raised hypotheses with main effects and interaction. F-value, p-value and eta squared are reported with significant effects marked in bold. AI group: ankle instability group and CON group: control group.

Hypothesis 1	Group effect			Direction effect			Group-direction interaction		
	F-value	p-value	η^2	F-value	p-value	η^2	F-value	p-value	η^2
SaEn	**9.92**	**0.004**	**0.30**	**52.45**	**< 0.0001**	**0.70**	0.01	0.934	0.000
CoD	1.48	0.237	0.06	0.12	0.734	0.01	3.21	0.086	0.122
ENT½	**6.18**	**0.024**	**0.27**	1.12	0.305	0.062	0.03	0.859	0.002
Hypothesis 2	Group effect			Direction effect			Group-direction interaction		
	F-value	p-value	η^2	F-value	p-value	η^2	F-value	p-value	η^2
SaEn	1.48	0.236	0.06	**21.56**	**< 0.0001**	**0.48**	0.61	0.442	0.026
CoD	1.94	0.180	0.097	0.64	0.435	0.03	0.16	0.698	0.009
ENT½	0.09	0.768	0.004	0.23	0.636	0.01	0.05	0.827	0.002
Hypothesis 3	Trial effect			Direction effect			Trial-direction interaction		
AI group									
	F-value	p-value	η^2	F-value	p-value	η^2	F-value	p-value	η^2
SaEn	0.94	0.352	0.058	**22.22**	**< 0.0001**	**0.597**	**9.46**	**0.008**	**0.387**
CoD	1.60	0.226	0.096	1.04	0.342	0.065	**7.57**	**0.015**	**0.335**
ENT½	0.60	0.458	0.063	0.12	0.737	0.013	1.16	0.310	0.114
CON group									
	F-value	p-value	η^2	F-value	p-value	η^2	F-value	p-value	η^2
SaEn	0.00	0.952	0.000	**17.17**	**0.003**	**0.682**	**5.79**	**0.043**	**0.420**
CoD	0.25	0.630	0.030	2.33	0.165	0.226	0.73	0.419	0.083
ENT½	1.45	0.273	0.195	1.29	0.300	0.177	0.70	0.435	0.104

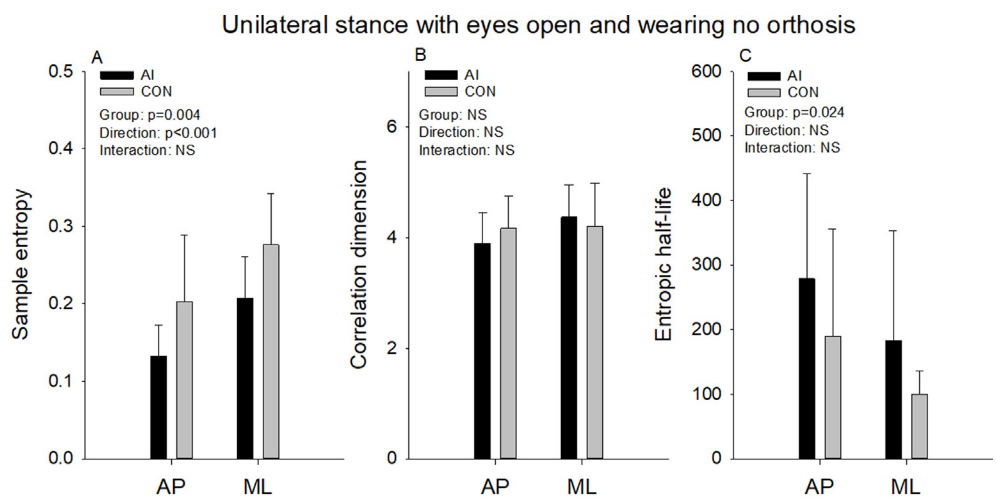

Fig. 3. Sample entropy, correlation dimension and entropic half-life in the AP and ML direction during unilateral stance with eyes open and wearing no orthosis for the AI group (black bars) and CON group (grey bars).

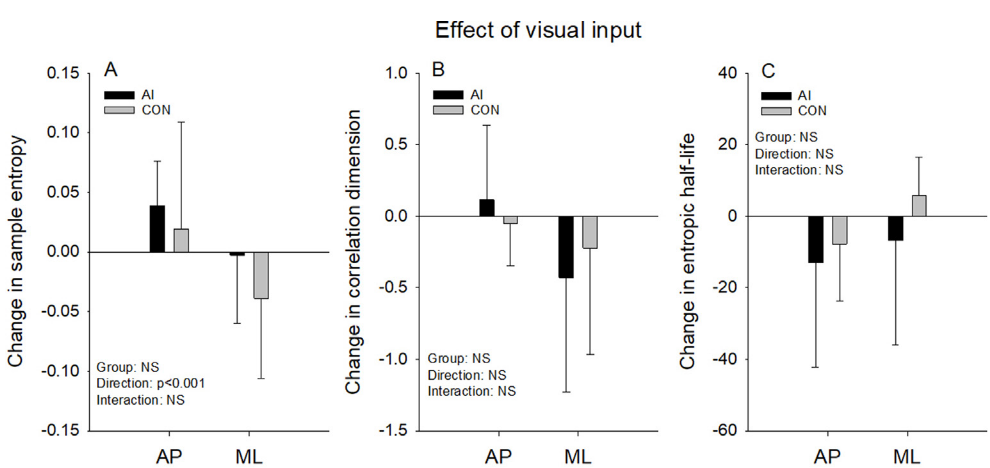

Fig. 4. The change in the sample entropy, correlation dimension and entropic half-life in the AP and ML direction from the unilateral stance trial with eyes open and wearing no orthosis to the eyes closed and wearing no orthosis trial for the AI group (black bars) and CON group (grey bars). Positive values indicate lower regularity (greater sample entropy), higher dimensionality (greater correlation dimension) and higher level of time dependency (higher entropic half-life).

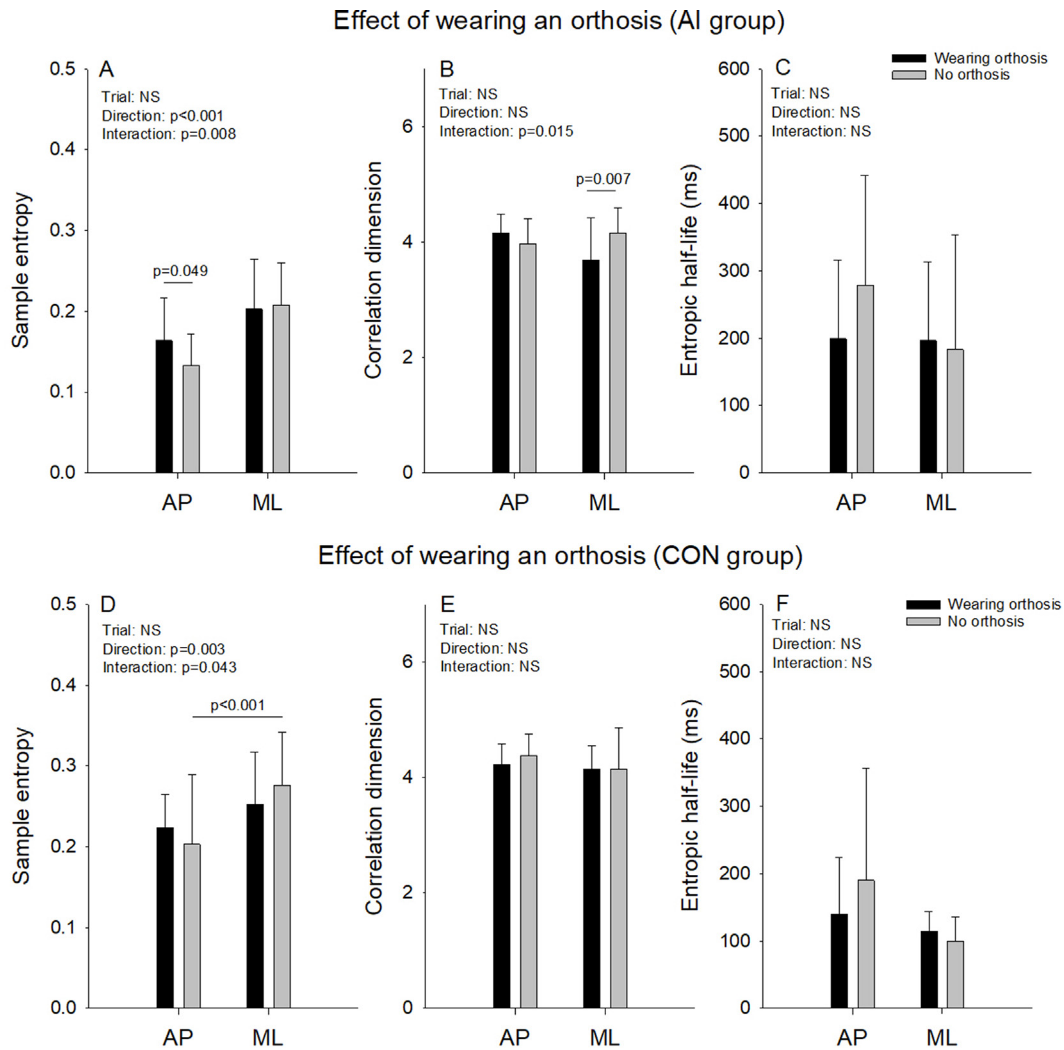

Fig. 5. Sample entropy (A and D), correlation dimension (B and E) and entropic half-life (C and F) in the AP and ML direction during unilateral stance with eyes open and with (black bars) or without (grey bars) an orthosis for the AI group (A, B and C) and the CON group (D, E and F).

to maintain postural control compared to controls [30]. This discrepancy in conclusion could be due to differences in methodology and inclusion criteria between the present study and the included studies of the aforementioned meta-analysis. The present study design did not enable time-to-boundary analysis and it is unknown if the included individuals would display similar results as previously reported [30]. Furthermore, the included nonlinear analyses assess the evolution of the COP trajectories across the entire trial whereas the time-to-boundary method estimates the time it would take the COP to reach the boundary of support [8]. These measures aim to quantify different characteristics of postural control and are not necessarily correlated. The significant effect of CoP direction observed only in SaEn indicates that deprivation of visual input decreased the regularity in the AP direction while increasing the regularity in the ML direction, without a simultaneous effect on dimensionality and time dependency. In light of the results related to the first hypothesis, it is evident that the deprivation or impairment of two different types of sensory input (e.g. visual and somatosensory) results in substantially different motor strategy responses.

The third hypothesis stated that wearing an ankle orthosis would significantly increase the regularity (decrease SaEn), reduce the dimensionality (decrease CoD) and increase the level of time dependency (increase ENT½) of the COP trajectory. This hypothesis was only partly confirmed as both a decrease in AP regularity and ML dimensionality were observed when the AI individuals wore an ankle orthosis. The remaining parts of the raised hypothesis were not confirmed. Wearing an orthosis should restrict ankle inversion during various activities, which seems to be the case for the AI individuals in the present study. In general this supports the use of orthosis for individuals with AI.

The order of completed trials was not randomized and the development of fatigue cannot be excluded. However, the participants were given a minimum of 2 min rest between each trial and none reported any subjective feeling of fatigue interfering with their performance. The present study did not control for leg dominance of the included participants. While leg dominance, ankle injury prevalence and one-legged balance are relevant topics that should be considered as possible influencing factors, we believe that future studies investigating this require a large participation groups than the present study.

In conclusion, the present study observed that individuals with AI exhibited higher regularity and greater time dependency during unilateral stance compared to healthy individuals. In contrast to previous observations, the present study did not observe a higher reliance on visual input to maintain postural balance in individuals with AI compared to healthy individuals. Finally, wearing an orthotic ankle brace affected the dynamics of postural control in AI individuals by reducing the dimensionality of the COP trajectory.

Declaration of interest

None.

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Appendix A. Supplementary data

See the supplementary material for the test of parameter consistency for the sample entropy calculation and the individual time delay and embedding dimension.

Supplementary data to this article can be found online at https://doi.org/10.1016/j.compbiomed.2019.05.018.

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