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Sensitivity of trunk variability and stability measures to balance impairments induced by galvanic vestibular stimulation during gait

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ABSTRACT

For targeted prevention of falls, it is necessary to identify individuals with balance impairments. To test the sensitivity of measures of variability, local stability and orbital stability of trunk kinematics to balance impairments during gait, we used galvanic vestibular stimulation (GVS) to impair balance in 12 young adults while walking on a treadmill at different speeds. Inertial sensors were used to measure trunk accelerations, from which variability in the medio-lateral direction and local and orbital stability were calculated. The short-term Lyapunov exponent and variability reflected the destabilizing effect of GVS, while the long-term Lyapunov exponent and Floquet multipliers suggested increased stability. Therefore, we concluded that only short-term Lyapunov exponents and variability can be used to asses stability of gait. In addition, to investigate the feasibility of using these measures in screening for fall risk, the presence or absence of GVS was predicted with variability and the short-term Lyapunov exponent. Predictions were good at all walking speeds, but best at preferred walking speed, with a correct classification in 83.3% of the cases.

1. Introduction

Kinematic variability during gait is increased in older adults, especially in those prone to falling [1–3], and may be a predictor of the risk of falling as it reflects the amount and magnitude of the perturbations encountered. Additional information might be gained from estimating stability with nonlinear dynamic stability measures, which are believed to provide information on how the neuromuscular system controls locomotion and responds to small perturbations [3–5]. These perturbations do not have to be superimposed, since the measures depend on perturbations inherent in the system [4,6,7]. Nonlinear dynamic stability of trunk kinematics seems most sensitive to differences between, e.g. elderly and young subjects [8]. Two measures used in this approach are maximum finite time Lyapunov exponents (estimating local stability) and maximum Floquet multipliers (estimating orbital stability). Both measures have a sound theoretical and mathematical basis [7,9,10]. Nevertheless, it remains to be shown whether these measures actually reflect balance impairments, and the probability of falling during walking [11,12].

Simulation studies have shown that Floquet multipliers do not correlate with the chance of falling using a simple walking model [13,14]. For maximum finite time Lyapunov exponents, results were more ambiguous. The Lyapunov exponent calculated for the long-term (3–4 steps) did not correlate with the chance of falling [14], while the short-term exponent (0–0.1 step [14] or 0–1 strides [15]) did. However, in studies on human walking, both Floquet multipliers and maximum finite time Lyapunov exponents were reported to discriminate elderly fallers from non-fallers, and older adults from younger controls [3,12,16]. Still, apart from the fact that these studies were non-experimental, they may have been biased by methodological choices, e.g. measuring at preferred rather than a fixed walking speed [17] and analyzing a fixed time rather than a fixed number of strides [18]. To our knowledge, only one experimental study has compared normal gait stability to balance when gait is impaired by walking on a compliant surface [19], indicating that impaired balance may be detected using maximum finite time Lyapunov exponents at the group level. However, walking speed was again uncontrolled. Thus, although there is evidence from modeling studies that Floquet multipliers and long-term Lyapunov exponents do not quantify the probability of falling while the short-term Lyapunov exponent does, the value

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of these measures for the assessment of balance control in real human walking remains to be further elucidated.

For this purpose, we measured individual subjects both in a stable condition and when balance was impaired. To simulate balance impairments such as caused by pathology or aging, we electrically stimulated the vestibular system [20] by randomly varying bilateral bipolar galvanic vestibular stimulation (GVS). It is known that GVS can be used to influence balance [21] and to study balance control during standing and walking [21–24]. Similar to aging, GVS affects stability mostly in the medio-lateral direction [25]. We thus focused on frontal plane kinematics.

Walking speed substantially affects the variability of trunk accelerations, as well as maximum finite time Lyapunov exponents [5,12,26,27]. Subjects were therefore measured while walking at different speeds; i.e. preferred walking speed and two fixed speeds. If changes in stability could be detected at preferred walking speed, this would be very practical as individuals can be expected to walk at this speed most of the time. Stability and variability at a fixed speed level could be used to study an individual's capacity to attain stable gait under given circumstances.

In summary, we tested whether (combinations of) measures of variability, and local and orbital dynamic stability were sensitive to experimentally induced impaired gait stability, during treadmill walking at several different speeds.

2. Methods

2.1. Subjects

Twelve healthy volunteers participated in this study (Table 1). Exclusion criteria were: history of recent lower extremity injuries or disabilities, any visible gait asymmetries, neurological deficits or mental problems, heart diseases or medication affecting the ability to walk or stand (e.g. anti-histamines). Alcohol consumption was prohibited for 24 h prior to testing. All participants filled in a medical history form and provided informed consent. The protocol was approved by the Local Ethical Committee.

2.2. Instruments

Portable, wireless inertial sensors (Xsens Technology, Enschede, The Netherlands) were attached to the right heel and on the spine at the level of T6 (Fig. 1). These inertial sensors measured 3D linear accelerations and angular velocities during each trial at 50 samples/s. During all trials, subjects wore their own comfortable shoes and clothes and a safety harness that allowed natural arm and leg swing.

In order to apply GVS, two flexible carbon surface electrodes were attached with electro conductive adhesive gel (Tac Gel™) over the mastoid bones. A computer-controlled stimulator (IDEE, Maastricht University, The Netherlands) was used to apply a quasi-random current stimulus with a maximal magnitude of about 2.2 mA to each subject, irrespective of skin and temporal bone conductance, inducing a mild impairment of stability without causing falls. This maximal magnitude was selected because it was beyond the response threshold of 0.2–0.5 mA found in other studies [24,28] and around the level of response saturation [22]. To prevent adaptation, the galvanic stimulus was composed of a linear summation of 5 sinusoids with different frequencies (0.02, 0.07, 0.11, 0.30, and 0.50 Hz), with the same phase and a maximum amplitude of 0.6 mA. We used a composition of these frequencies since earlier research showed that stimuli with frequencies of 0–1 Hz [29], 0.05–5 Hz [30], and 0.2 and 0.5 [22] resulted in continuous modulation of medio-lateral postural sway. The cathode and anode switched when the current became negative. For the subject, the resulting stimulus therefore unpredictably changed not only in magnitude but also in direction.

2.3. Procedure

Prior to testing, the subject's reaction to GVS was tested, although no negative physiological or mental side effects are known [22,23]. When discomfort was experienced, the electrodes were repositioned, which solved the problem in all cases. Subsequently, each subject's preferred walking speed (PWS) was determined during over ground walking, while subjects were unaware of the measurement taking place. The subjects walked three times over a fixed distance (10 m) in steady-state gait and the time taken was measured and averaged.

After familiarisation, measurements were performed while walking at 0.69 m/s (2.5 km/h; trial duration of 3.5 min), at PWS (3 min) and at 1.53 m/s (5.5 km/h; 2.5 min) on a treadmill (Biostar Giant™, Biometrics, Almere, The Netherlands). Speed levels, and starting or ending with the GVS-trial within each speed level were randomized, and a 5 min break was allowed between speed levels. Because galvanic induced body sway is directed lateral to the orientation of the head, regardless of the body orientation [20], participants were instructed to look straight ahead.

2.4. Data analysis

All measures are presented in inertial sensor axes. During normal upright stance the positive X-axis was directed vertically downwards, the positive Y-axis was directed towards the right and the positive Z-axis was directed towards the front of the subject, as illustrated in Fig. 1.

Raw unfiltered data were analyzed to assure that the nonlinearity was not lost or altered due to filtering [31]. MATLAB (version 7.5, The MathWorks BV, Natrick, USA) and SPSS (version 17.0, SPSS Inc., Chicago, USA) were used for analyses. Heel strikes were determined as the maximal vertical acceleration of the right heel sensor and stride time was determined as the time between two consecutive right heel strikes. Strides were time-normalized to the average stride length for each speed level. Note that this yielded different numbers of samples between speed conditions, but since we did not intend to study the effect of speed this was deemed less important than the fact that this allowed for minimizing the change in number of samples in the normalization procedure.

At each percentage of the stride cycle, the standard deviation between strides was calculated for the trunk kinematics in the medio-lateral plane, i.e. linear acceleration in the Y-direction (SDa) and angular velocity around the Z-axis (SDω). Subsequently, these estimates were averaged over the normalized stride cycle resulting in variability measures.

For the nonlinear dynamic measures, a state space was created by using the linear acceleration and angular velocities of the trunk in all three directions and their time-delayed copies [11]. Because these time series all had the same average frequency due to normalization, a fixed time-delay could be used for state space construction [18,26]. We used a time-delay of a quarter of the normalized stride time, which roughly corresponds to the derivative. Local and orbital stability were calculated over time series containing a fixed number of strides to exclude the influence of data series length [18].

Local stability was estimated by calculating short-term (over 0–0.5 stride [11,17,18]) and long-term (over 4–10 strides [4,5,11,17,18,32]) Lyapunov exponents (λ_s and λ_l), which express the exponential rate of divergence or convergence after a small disturbance of nearby orbits in state space [4,9]. Because nearby orbits correspond to nearly identical states, a positive λ indicates that systems with initial differences will soon behave quite differently, and stability is low [4,9].

Orbital stability was calculated using Floquet multipliers (FM) which reflect how the system responds to small perturbations discretely, from one cycle to the next [7]. A limit cycle is orbitally stable if all FM are smaller than one, else the limit cycle is orbitally unstable [7]. For analysis, the mean of all maximum FM at each instant in time was calculated, which gives an index of the instability over the stride cycle.

2.5. Statistical analysis

A 2×3 (with/without GVS $\times$ walking speed) repeated measures ANOVA was performed to test the effects of GVS and walking speed on SDa, SDω, λ_s, λ_l and FM. Subsequently, post hoc paired sample t-tests with Bonferroni correction were performed to determine whether the effect of GVS could be detected at each walking speed. In addition, to assess the sensitivity of the measures at the individual level, for each speed condition, discriminant analyses were performed with GVS as classification variable and selected measures (based on the preceding analyses and after checking for multi-collinearity) as predictor variables. A significant discriminant model allows prediction of whether GVS was present, based on the trunk kinematic variables. The percentage of correct predictions and the specificity and sensitivity were assessed. In all tests, a p-value less than 0.05 was considered significant.

3. Results

Due to technical problems, the 0.69 m/s with GVS condition of one subject did not yield useful data and was excluded from analysis. For all subjects, 115 strides were analyzed for all trials. Preferred walking speed (1.47 (SD 0.14) m/s) was not significantly different from the fast (1.53 m/s) condition ($p = 0.079$, paired t-test). Normalization for the 0.69 m/s was done to 78 samples/stride and for the two faster conditions to 49 samples/stride.

Table 1
Subjects characteristics, means with standard deviations between brackets.

Gender	3 female, 9 male
Age	23.7 (2.4) yrs
Height	1.82 (0.10) m
Body mass	76.3 (11.9) kg

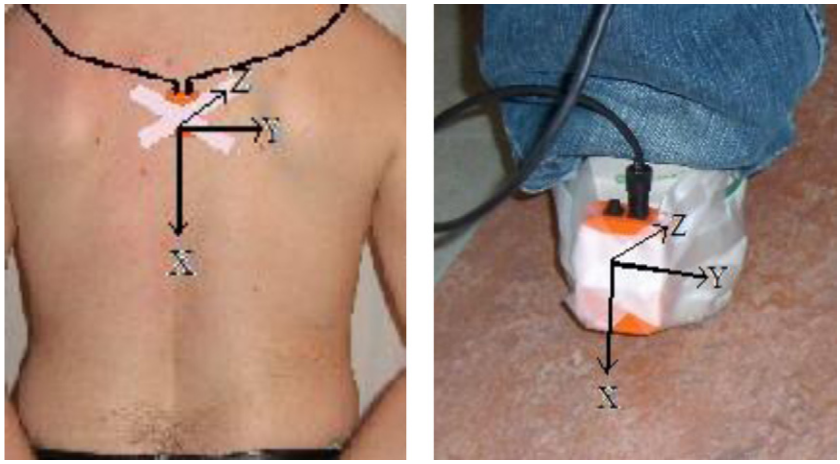

Fig. 1. Inertial sensor orientation, positive axis are drawn.

A significant main effect of GVS was found for all measures (Table 2); SDa, SDω and λ_s were increased relative to the condition without GVS, while λ_l and FM were decreased (Fig. 2). Main effects for walking speed were found for SDa, SDω, λ_s and λ_l (Fig. 2). Post hoc *t*-tests demonstrated that the effect of GVS could not be detected for all measures at individual walking speeds; for SDa and λ_s the effect of GVS was significant for all speeds ($p \leq 0.001$ and $p \leq 0.013$, respectively). However, for SDω the effect of GVS was only significant at 0.69 m/s and 1.53 m/s ($p \leq 0.002$) and for λ_l only at 1.53 m/s ($p < 0.001$).

Based on these results and a high partial correlation (corrected for GVS and walking speed) between SDa and SDω ($r = 0.698$, $p < 0.001$), SDa and λ_s were further analyzed to determine their predictive value for impaired stability at the individual level

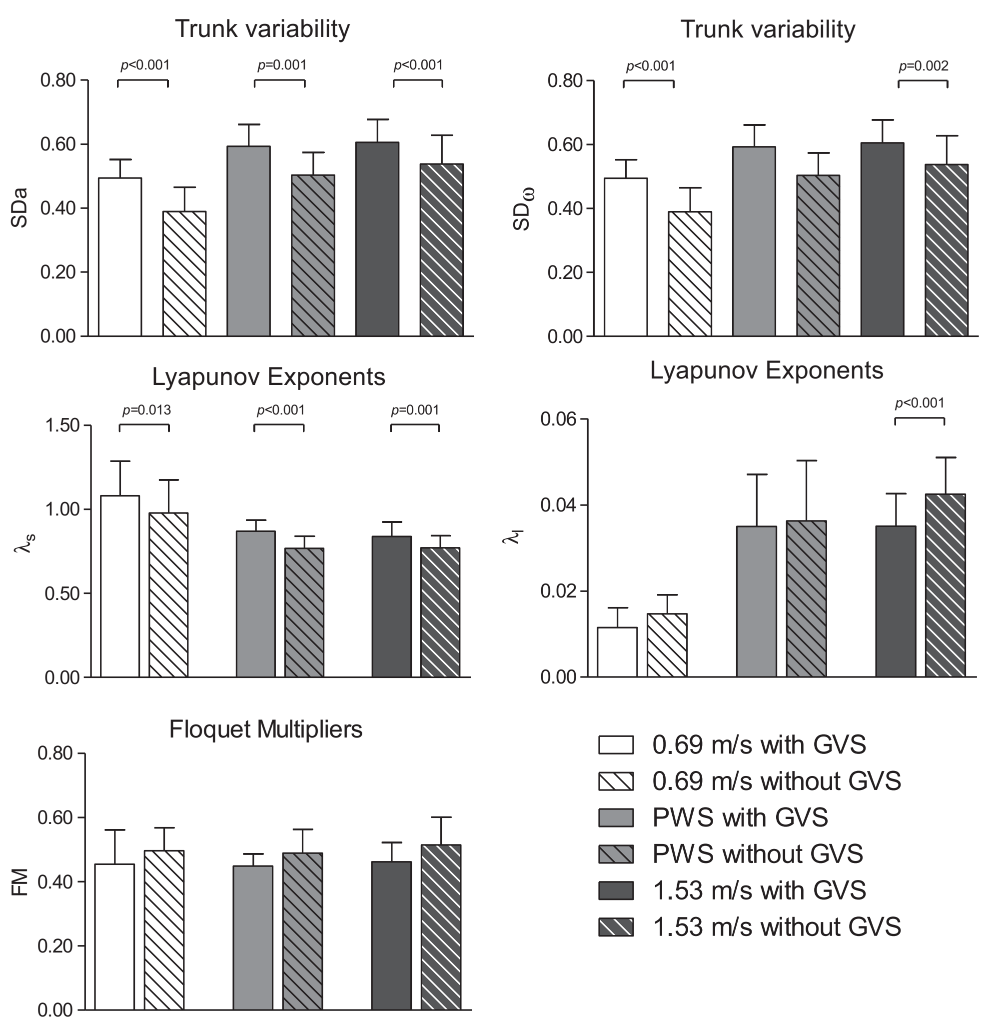

Fig. 2. Gait stability measures with and without GVS, means with standard deviations given in error bars.

Table 2
Results of repeated measures ANOVAs testing the main and interaction effects of GVS and gait speed on each gait stability parameter. NS = non-significant.

	GVS	Walking speed	Interaction effect
SDa	$F(1.00, 10.00) = 64.13$ $p < 0.001$	$F(2.00, 20.00) = 24.47$ $p < 0.001$	NS
SDω	$F(1.00, 10.00) = 32.30$ $p < 0.001$	$F(2.00, 20.00) = 17.16$ $p < 0.001$	NS
FM	$F(1.00, 10.00) = 5.46$ $p = 0.042$	NS	NS
λ_s	$F(1.00, 10.00) = 36.15$ $p < 0.001$	$F(1.17.7446) = 14.30$ $p = 0.001$	NS
λ_l	$F(1.00, 10.00) = 8.79$ $p = 0.014$	$F(2.00, 20.00) = 39.40$ $p < 0.001$	NS

(Fig. 3). Discriminant analysis yielded a significant model at each walking speed. At 0.69 m/s, 81.8% of the 22 trials were correctly classified (Wilk's $\lambda = 0.599$, $p = 0.008$) as with or without GVS with a sensitivity of 72.7% and a specificity of 90.9%. At 1.53 m/s, the discriminant model (Wilk's $\lambda = 0.615$, $p = 0.006$) correctly classified trials in 79.2% of the 24 trials with a specificity of 75% and sensitivity of 83.3%. At PWS, the discriminant model (Wilk's $\lambda = 0.494$, $p = 0.001$) correctly predicted 83.3% of 24 trials, with both a specificity and sensitivity of 83.3%.

4. Discussion

The present study investigated whether (combinations of) medio-lateral variability of trunk accelerations (SDa and SDω) and measures of local and orbital dynamic stability reflect the assumedly impaired balance induced by GVS.

Previous work indicated that increased kinematic variability may reflect balance impairments [1–3]. In line with this, we found that the variability of medio-lateral trunk kinematics was increased with GVS, supporting the use of these measures in assessment of balance control. In apparent contrast, one study has reported lower variability of medio-lateral trunk kinematics in frail compared to fit elderly women [27]. Possibly, this reflects compensatory gait adaptations in the frail women. Our results also support the use of λ_s as a measure to assess balance control, which is in line with previous modeling [14,15], showing that the probability of falling in a simple walking model was correlated to λ_s. For λ_l and FM, the results were opposite to expected from theoretical rationale, as they were lower when balance was impaired, which suggests that these measures cannot be used to assess stability during gait. This is in line with previous findings that FM [13,14] and λ_l [14] are not related to the probability of falling in a simple walking model. Su and Dingwell [14] argued that since the model never "fell over" it did remain "stable" and thus the inherent stability [6] may still have been quantified by these measures. However, we altered the stability of human locomotion in our study using GVS and therefore expected to measure this decreased (inherent) stability, even though the subjects did not fall. The reason that we did not find this, or rather, found the opposite, may be due to compensatory changes, which occur at longer time scales than half a stride. However, such a claim would require additional research. For now, the only conclusion that can be drawn from these results appears to be that λ_l and FM cannot be used to assess stability during gait.

Discriminant analysis was used to test whether the measures used could detect the presence of impaired gait stability on an individual basis. Trials were classified according to presence or absence of GVS, with SDa and λ_s as independent variables. This analysis was successful for all walking speeds, and slightly better for PWS. Overall, a combination of SDa and λ_s allowed reasonably accurate classification, suggesting that these measures combined may be useful in estimating individual fall risk. Moreover, evaluation of intervention effects on fall risk could be an even

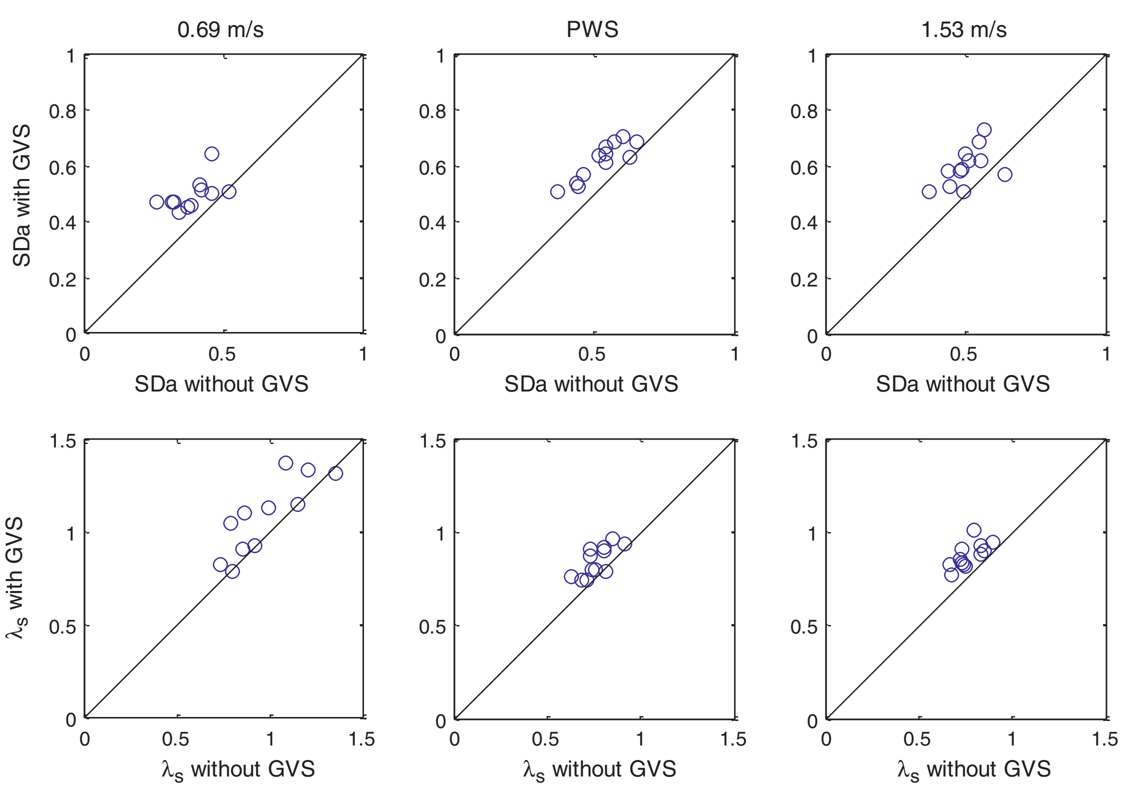

Fig. 3. Scatter plots of the individual data with and without GVS for variability of trunk acceleration and short-term Lyapunov exponents per walking speed. The diagonal lines represent the identity line.

more easily attainable goal, as this would exclude the effects of between-subject variance, which appeared to be substantial in this study (Fig. 2).

In the current study, balance control was impaired by application of randomly varying GVS. The mean absolute effects were 11% for λ_s, 20% for SDa, and 10% for SDω. The effects of aging reported in literature for λ_s vary from 35 to 50% [8,12]. Unfortunately, no reference values were found for SDa and SDω. Although walking speed varied in the above-mentioned studies, which influenced stability, it seems that instabilities induced by GVS are smaller than those due to aging. This would imply that effects induced by aging can be detected quite well.

We used inertial sensors for measuring trunk kinematics rather than more commonly used optical methods, as the inertial sensors can easily be used in clinical practice [11]. We have previously shown that measured local dynamic stability is comparable between these measuring methods [12]. However, values of FM were less well correlated between these measurement systems. This appeared due to the sensitivity of FM to measurement noise, which could also have influenced our results. However, the fact that we did find a significant main effect of GVS on FM, although in a direction opposite than expected, suggests that the measurement noise was of limited importance.

The time-normalization that we used was different between the walking speeds, i.e. to 78 data points/stride for walking at 0.69 m/s and 49 data points/stride for the two other speeds. Consequently, the results at 0.69 m/s cannot be compared to both other walking speeds [26], but extensive up or down sampling, which may influence the results of interest, was prevented in this way. Another limitation of our study was the use of a treadmill to control walking speed, because differences in variability and local dynamic stability of gait between treadmill and over ground walking have been reported [4,32,33]. Since it may be more feasible in clinical practice to assess stability during over ground walking, subsequent studies should address how well these measures work during over ground walking. In addition, heel contact was determined as the maximal vertical acceleration of the right heel sensor. To further simplify the measurement system, subsequent studies may investigate whether the sensor on the spine can also be used to detect heel contact, as this would facilitate clinical implementation.

In conclusion, variability and λ_s of trunk kinematics can be used to assess balance control in gait, while λ_l and FM cannot. Detection of the presence or absence of GVS was fairly accurate with SDa (linear trunk acceleration variability) and λ_s (the short-term Lyapunov exponent of trunk kinematics). It was found that prediction was best at preferred walking speed, with a correct classification of 83.3%. This suggests that a portable system can be used for the diagnosis of stability problems during gait.

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Conflict of interest statement

None of the authors of this paper had any conflict of interest that could inappropriately influence (i.e., bias) the presented work.

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