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Analysis of gait and balance through a single triaxial accelerometer in presymptomatic and symptomatic Huntington's disease

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ABSTRACT

Purpose: To investigate the capacity of a single triaxial accelerometer sensor in detecting gait and balance impairments in pre-manifest and manifest Huntington's disease (HD) subjects.
Methods: Fourteen manifest HD (MHD) (age: 51.83 ± 14.8), ten pre-manifest HD (PHD) (age: 44.8 ± 11.7) and ten healthy subjects (HLY) (age: 56.4 ± 10.9) were recruited. The sensor was attached to the upper sternum as subjects completed gait and Romberg balance tests. An inverted pendulum model of the body's centre of mass and an unbiased autocorrelation procedure were employed to derive gait parameters from the triaxial accelerometer signal. The accuracy of the gait measurements was compared to those recorded by a computerized walkway.
Results: Strong agreement was seen between the sensor and the walkway; cadence (ICC = 0.95, CI = [0.75, 0.97]), velocity (ICC = 0.94, CI = [0.75, 0.97]) and step length (ICC = 0.89, CI = [0.77, 0.95]). Sensor derived velocity was significantly higher in HLY ($p < 0.001$) and PHD ($p < 0.005$) when compared to MHD. Step and stride length was significantly longer in HLY ($p < 0.05$) and PHD ($p < 0.001$) when compared to MHD. Significant differences between subject groups across all four balance tasks ($p < 0.001$) were found.
Conclusion: An accelerometer based sensor may be an effective means of differentiating between pre-manifest and manifest Huntington's disease subjects.

1. Introduction

Huntington's disease (HD) is an autosomal dominant neurodegenerative condition with a principal symptom of progressive movement disorder [1]. The prevalence of the disease has been estimated at 4–10 affected individuals per 100,000 with higher incidence in women than in men. Considerable research has been conducted into the progression of gait disorder through the various stages of HD. Specifically, manifest HD (MHD) subjects experience decreased stride length, gait velocity and cadence [2–4], higher variability in stride length and step-time [5] and significant degradation in balance [5]. Further investigation into gait impairments in pre-manifest HD (PHD) subjects (gene carriers not yet demonstrating motor symptoms and functional decline) has also generated attention. Rao et al. [3] employed a computerized walkway and found PHD subjects demonstrated decreased gait velocity, stride length and time in double support. They further identified a high correlation between these gait parameters and predicted years to onset. Delval et al. [5,7] investigated the role of akinesia in HD and found that PHD subjects demonstrated a shorter first step duration and lower-amplitude postural adjustments. Panzera et al. [8] found MHD subjects generated significantly less rising force and significantly higher sway velocity at the centre of gravity while performing three functional postural tasks.

These studies highlight the advantages of quantitative gait analysis in HD populations however; laboratory-based systems are typically expensive and are not available in all clinical settings. Furthermore, Rao et al. [9] has highlighted the limitations of ordinal based clinical tests by showing that the Functional Reach Test and the traditional Timed up and Go (TUG) test are not sensitive in detecting motor impairments in PHD subjects. Therefore, a significant interest has grown in the development of alternative gait analysis tools.

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Accelerometry has shown significant potential in the measurement of spatio-temporal gait parameters [10–13] however, caution must be exercised as reduced accuracy at decreased gait speed has been shown for commercially available activity monitors [13]. By modelling the trajectory of the body's centre of mass using an inverted pendulum model Zijlstra et al. [10] was able to accurately derive several gait parameters. Moe-Nilssen et al. [11] demonstrated how an unbiased autocorrelation procedure may be employed to derive cadence and step length. Significantly, the autocorrelation method also provides the opportunity to calculate a measure of gait regularity and symmetry. These techniques demonstrated high reliability when applied to a population of 121 elderly by Bautmans et al. [12].

Accelerometry has also been used to investigate balance [14–17]. O'Sullivan et al. [14] demonstrated a high correlation between accelerometry and the Berg Balance Scale (BBS) in a population of 21 elderly subjects. Accelerometry measured during quiet standing has also been adapted to differentiate between young and old healthy subjects [15]. In a HD population fourteen MHD subjects demonstrated lower balance confidence and impairments on clinical measures of balance when compared to nine healthy controls [17].

The objective of this study was to investigate the capacity of a single accelerometer based sensor to derive spatio-temporal parameters of gait and balance in a HD population and investigate its potential to differentiate between subject groups. This is a continuation of previously presented work where an investigation of an instrumented TUG test was performed [18].

2. Methods

2.1. Participants

Fourteen manifest HD (age: 51.83 ± 14.8), ten pre-manifest HD (age: 44.8 ± 11.7) and ten healthy subjects (age: 56.4 ± 10.9) were recruited from the South Wales HD service. All subjects gave their written informed consent before participation in accordance with local research ethics requirements (09/WSE02/24). The healthy (HLY) control group consisted of 2 students, 2 staff, 2 family members and 4 friends recruited from outside the University. The inclusion criterion for the PHD subjects was a positive genetic test for HD, a diagnostic confidence score of three or less as rated by a neurologist indicating the absence of definitive motor signs of HD. The inclusion criterion for MHD subjects was a positive genetic test for HD and a score of four on the motor diagnostic confidence scale of the Unified Huntington's Disease Rating Scale (UHDRS) [19]. This scale accesses motor function, cognition, behaviour and functional abilities therefore providing a uniform assessment of the clinical features and course of HD. Subject exclusion criteria included a history of coexisting neurological conditions such as stroke and severe visual problems. For HD subjects scores on the motor section of the UHDRS and the Total Functional Capacity (TFC) scale were recorded. The total motor score has 31 items with a maximum possible score of 124 (indicating maximum disability), and includes tests of chorea, dystonia, rigidity, bradykinasia, coordination, balance and gait

2.2. Apparatus and procedure

2.2.1. Accelerometer

An accelerometer based sensor [18] was attached to the thorax of each subject as they completed the clinical tests. The AD_BRC sensor contains a ±2.5–10 g triaxial accelerometer, a Texas Instruments microprocessor, a 2000 mAh lithium-ion battery and samples data at a frequency of 250 Hz. The sensor was calibrated prior to attachment by rotation of the accelerometer through six different known angles as outlined by Bourke et al. [20]. To correct for any misalignment or tilt of the sensor caused by the site of attachment the accelerometer's capacity as an inclinometer was employed as outlined by Moe-Nilssen et al. [15]. Data from the sensor were high-pass filtered with a 3rd order normalized elliptical filter with a passband frequency of 0.25 Hz, 0.01 dB of ripple in the passband and 100 dB of attenuation in the stopband [10].

2.2.2. Gait analysis

The inverted pendulum model proposed by Zijlstra et al. [10] was used to extract spatio-temporal gait parameters. Furthermore, by employing an unbiased autocorrelation procedure [11] and subsequently analyzing the ratio of the correlation coefficients at the first and second dominant periods observations on gait regularity and symmetry were made. Step and stride regularity were calculated across the anteriorposterior (AP), mediolateral (ML) and longitudinal (LD) axes respectively. Coefficient of variation ($CV = 100 \times \mu/\sigma$) for step time, step length, stride length and step time asymmetry [12] were also calculated. Step time asymmetry calculates the ratio of the difference between mean step time of individual legs to the combined mean step time of both legs. The accuracy of the accelerometer was gauged against the parameters provided by a GAITRite® system (CIR Systems, Inc.: Havertown, PA). This 4.8 m long instrumented walkway uses embedded sensors to record pressure applied at each footfall as a function of time. Menz et al. [21] investigated the reliability of the GAITRite® system in a young and old population in quantifying spatio-temporal gait parameters and reported excellent ICC values (ICC = 0.88–0.92). These results were mirrored in an independent study by Webster et al. [22] (ICC = 0.91–0.99). Finally, validity and reliability of the GAITRite® system has also been established in a HD population [3] (ICC = 0.86–0.95). Each participant was asked to perform five trials of walking at their comfortable pace with all statistical analysis performed on respective mean values. As noted by Hof et al. [23] several gait parameters are speed dependant and thus the spatio-temporal variables were normalized using subject specific height prior to statistical analysis. Subjects were instructed to begin walking two meters before the edge of the walkway and stop two meters beyond the end of the walkway. Finally, subjects were given a practice trial at the beginning of the testing session.

2.2.3. Balance

Balance and postural control were assessed using the Romberg (RB) test [24]. The test measures the length of time (max 30 s) the subject can stand with ankle malleoli touching, arms crossed with palms touching the opposite shoulder. Each test was conducted twice with all statistical analysis performed on the mean of both attempts. As previously discussed the sensor was calibrated and compensation for possible tilt was considered. A further issue specific to these type of balance tests is that a subject may slowly shift their position during the testing period (a shift unrelated to balance control) and this can result in a low frequency drift over time [15]. Thus the triaxial signal was detrended using a second-order polynomial fit curve [14]. Finally, for the centre 80% of data of each task the root mean square of the ML and AP axes along with the combined instantaneous vector sum ($\sqrt{a_{ML}^2 + a_{AP}^2 + a_{LD}^2}$) were calculated.

2.3. Statistical analysis

Signal processing and parameter extraction was performed using MATLAB (Matlab 7.10.0, The MathWorks, USA) and statistical analysis was performed using PASW-Statistics 17.0.1 (SPSS Inc., IL, USA). The levels of agreement between the sensor and GAITRite® were assessed by intraclass correlation coefficients (ICCs) of the type (2, k). For each parameter the assumption of a normal distribution was assessed by a Kolmogorov–Smirnov Goodness of Fit test ($p < 0.05$). Dependent on this analysis group differences were assessed using either a one-way ANOVA ($p < 0.05$) or a Kruskal–Wallis ($p < 0.05$) test, respectively. Post hoc analysis was performed with Tukey HSD tests or a Mann–Whitney U-test with appropriate Bonferroni correction (adjusted alpha: ($p < 0.0125$) for multiple comparisons. Using Receiver Operating Characteristics (ROC) curve analysis the sensitivity and specificity of statistically significant variables was investigated. Specifically, the discriminatory power of a parameter can be evaluated by the area under the ROC curve (AUC) – a larger area is indicative of higher sensitivity and specificity [25].

3. Results

Table 1 summarizes subject characteristics. HLY subjects were generally older and taller than both PHD and MHD subjects but these differences were not significant ($p < 0.05$). PHD subject's weight was generally greater than that of HLY and MHD subjects but again this was not significant ($p < 0.05$). All PHD subjects scored either 0 ($n = 6$) or 1 ($n = 4$) on diagnostic confidence. PHD had UHDRS total motor scores (4.8 ± 5.3) and TFC (13 ± 0) indicating that clinical examination detected no specific motor

Table 1
Subject characteristics ($HLY_n = 10$, $PHD_n = 10$, $MHD_n = 14$) $\bar{x} \pm s$ = mean ± standard deviation.

Characteristic	Healthy ($\bar{x} \pm s$)	Pre-manifest HD ($\bar{x} \pm s$)	Manifest HD ($\bar{x} \pm s$)
Age (year)	56.45 ± 10.93	44.81 ± 11.79	51.83 ± 14.82
Sex	5M/5F	4M/6F	8M/6F
Weight (kg)	75.91 ± 7.62	81.86 ± 27.94	68.38 ± 9.35
Height (cm)	171.15 ± 6.66	171.05 ± 9.94	167.58 ± 8.72
Right leg (cm)	93.00 ± 5.52	92.55 ± 5.63	89.88 ± 3.62
Left leg (cm)	92.60 ± 5.24	92.15 ± 5.87	89.67 ± 3.68
UHDRS MS	NA	4.80 ± 5.30	54.15 ± 13.02
UHDRS TFC	NA	13 ± 0	6.33 ± 2.18

Table 2
Gait analysis (all subject groups combined. $n = 34$) $\bar{x} \pm s$ = mean ± standard deviation.

Parameter	GAITRite® ($\bar{x} \pm s$)	AD_BRC ($\bar{x} \pm s$)	ICC [CI]
Velocity (m/s)	1.17 ± 0.29	1.21 ± 0.31	0.94 [0.75, 0.97]
Cadence (s/m)	109.76 ± 13.84	108.86 ± 14.95	0.95 [0.75, 0.97]
Step length (m)	0.58 ± 0.11	0.57 ± 0.13	0.89 [0.77, 0.95]
Step time (s)	0.56 ± 0.08	0.54 ± 0.09	0.88 [0.72, 0.94]
Stride length (m)	1.18 ± 0.23	1.23 ± 0.22	0.88 [0.71, 0.95]

impairments and no functional limitations in these individuals. MHD subjects had UHDRS total motor scores (54.15 ± 13.02) and TFC (6.33 ± 2.18).

3.1. Gait analysis

Table 2 compares the gait parameters derived from the accelerometer to those recorded by the GAITRite® software across all subject groups. The accelerometer sensor displayed excellent agreement to the GAITRite® system for cadence (ICC = 0.95 [0.75, 0.97]) and velocity (ICC = 0.94 [0.75, 0.97]). Agreement was very good for step length (ICC = 0.89 [0.77, 0.95]), step time (ICC = 0.88 [0.72, 0.94]) and stride length (ICC = 0.88 [0.71, 0.95]) respectively. Table 3 presents means and standard deviations across each subject group and the respective statistical group differences. Fig. 1(A–E) presents box plots representing the parameters which were statistically significant between subject groups.

Sensor derived velocity was significantly higher in HLY ($p < 0.001$) and PHD ($p < 0.005$) when compared to MHD. Step and stride length was significantly longer in HLY ($p < 0.001$) and PHD ($p < 0.005$) when compared to MHD. No significant group differences were found for cadence and step time. Interestingly, significant group differences were found between each subject category for each coefficient of variation parameter. HLY step time CV, step length CV and stride length CV were significantly different from PHD ($p < 0.005$) and from MHD ($p < 0.001$). PHD step time CV, step length CV and stride length CV were significantly different from MHD ($p < 0.01$). No significant difference was found for step time asymmetry. Finally, step and stride regularity were significantly different between all three groups in the ML axis (ML step: ($p < 0.001$), ML stride: ($p < 0.005$)) and LD axis (LD step: ($p < 0.002$), LD stride: ($p < 0.002$). Similarly, for the AP axis step and stride regularity demonstrated significant group differences (AP step: ($p < 0.002$) AP stride: ($p < 0.002$)) with the exception of stride regularity between PHD and MHD.

Table 3
Sensor gait analysis ($HLY_n = 10$, $PHD_n = 10$, $MHD_n = 14$). ML = mediolateral, AP = anteriorposterior, LD = longitudinal. $\bar{x} \pm s$ = mean ± standard deviation.

Parameter	Healthy ($\bar{x} \pm s$)	Pre-manifest HD ($\bar{x} \pm s$)	Manifest HD ($\bar{x} \pm s$)
Velocity (m/s)	1.44 ± 015[†,d]	1.30 ± 0.21[†,b]	0.96 ± 0.29
Cadence (s/m)	110.74 ± 7.95	111.80 ± 15.82	105.20 ± 18.33
Step length (m)	0.65 ± 0.06[†,d]	0.64 ± 0.10[†,b]	0.48 ± 0.13
Step time (s)	0.51 ± 0.03	0.52 ± 0.06	0.60 ± 0.11
Stride length (m)	1.38 ± 0.10[†,d]	1.33 ± 0.15[†,b]	1.02 ± 0.19
Step time coeff. of var.	2.61 ± 0.51[*,b,d]	5.45 ± 2.40[†,a]	11.78 ± 6.21
Step length coeff. of var.	2.13 ± 0.39[*,b,d]	5.34 ± 2.82[†,a]	16.31 ± 10.70
Stride length coeff. of var.	2.01 ± 0.25[*,b,d]	4.70 ± 2.29[†,a]	13.81 ± 8.64
Step time asymmetry	2.17 ± 1.39	3.54 ± 2.11	5.43 ± 3.08
ML step regularity	−0.75 ± 0.10[*,d]	−0.54 ± 0.12[†,d]	−0.35 ± 0.10
ML stride regularity	0.69 ± 0.10[*,b]	0.53 ± 0.10[†,b]	0.34 ± 0.13
AP step regularity	0.81 ± 0.09[*,c]	0.67 ± 0.05[†,c]	0.50 ± 0.14
AP stride regularity	0.79 ± 0.08[*,c]	0.57 ± 0.12	0.50 ± 0.11
LD step regularity	0.78 ± 0.09[*,c]	0.58 ± 0.09[†,c]	0.37 ± 0.13
LD stride regularity	0.72 ± 0.09[*,c]	0.52 ± 0.12[†,c]	0.29 ± 0.16
Romberg balance tests			
Eyes Open Feet Together	0.025 ± 0.003[†,e]	0.026 ± 0.004[†,e]	0.035 ± 0.003
Eyes Closed Feet Together	0.026 ± 0.004[*,d,e]	0.031 ± 0.004[†,e]	0.038 ± 0.005
Eyes Open Feet Apart	0.020 ± 0.005[†,e]	0.022 ± 0.005[†,e]	0.029 ± 0.005
Eyes Closed Feet Apart	0.023 ± 0.003[*,b,e]	0.025 ± 0.003[†,e]	0.034 ± 0.004

[a] $p < 0.01$.
[b] $p < 0.005$.
[c] $p < 0.002$.
[d] $p < 0.001$.
[e] $p < 0.0001$.
[*] Significantly different from pre-manifest HD and manifest HD.
[†] Significantly different from manifest HD.

Table 4
ROC analysis ($HLY_n = 10$, $PHD_n = 10$, $MHD_n = 14$). HLY = healthy, PHD = pre-manifest HD, MHD = manifest HD. ML = mediolateral, AP = anteriorposterior, LD = longitudinal. Step regularity: first dominant peak from unbiased autocorrelation procedure (A_{d1}): regularity of the acceleration between consecutive steps.

Parameter (optimal threshold)	HLY vs. PHD and MHD		HLY vs. PHD	
	Sens.	Spec.	Sens.	Spec.
Velocity (1.31 m/s)	83	80	60	80
Step length (0.58 m)	76	90	50	90
Stride length (1.35 m)	78	90	50	90
Stride length coeff. of var. (2.65)	91	90	80	90
Step time coeff. of var. (3.06)	83	100	80	100
ML step regularity (−0.64)	100	100	100	100
AP stride regularity (0.69)	91	90	50	80
LD step regularity (0.69)	91	90	80	90
Eyes Closed Feet Together (0.029)	93	100	50	90

Stride regularity: second dominant peak from unbiased (A_{d2}) autocorrelation procedure: regularity of the acceleration between consecutive strides.
Sens. = sensitivity ($\sum TP/(\sum TP+\sum FN) \times 100$); probability of a positive test among subjects with HD.
Spec. = specificity ($\sum TN/(\sum TN+\sum FP) \times 100$); probability of a negative test for subjects without HD.

3.2. Balance analysis

The lower section of Table 3 presents the calculated root mean square values after appropriate adjustment for tilt and low frequency drift. Although separate values were calculated for the ML and AP axes, very similar results were found from analysis of the vector sum and thus only these values are presented.

Kruskal–Wallis analysis revealed significant differences between subject groups across all four balance tasks ($p < 0.0001$). Post hoc analysis found significant differences between HLY & MHD and PHD & MHD across all balance tasks ($p < 0.0001$) however, significant differences between HLY and PHD were only found for tasks requiring eyes closed; Eyes-Closed Feet-Together ($p < 0.001$) and Eyes-Closed Feet-Apart ($p < 0.005$). Table 4 presents the ROC curve analysis. Several parameters displayed very high sensitivity and specificity for HLY vs. PHD and MHD. However, these values are slightly reduced for HLY vs. PHD. Notably, mediolateral step regularity demonstrated perfect distinction across subjects with and without the HD mutation.

4. Discussion

Our results demonstrate the capacity of a single triaxial accelerometer attached at the sternum in differentiating between pre-manifest and manifest Huntington's disease subjects. Specifically, MHD subjects exhibited lower velocity, shorter step and stride length, inferior gait symmetry/regularity and greater postural sway when compared to HLY and PHD subjects.

The strong agreement between the AD_BRC sensor and the GAITRite® software across all five spatio-temporal parameters (velocity, cadence, step length, step time and stride length) demonstrates the accuracy of the sensor and reiterates the reliability of the inverted pendulum [10] model. As noted in Section 2.2.2 normalization of spatio-temporal variables by subject

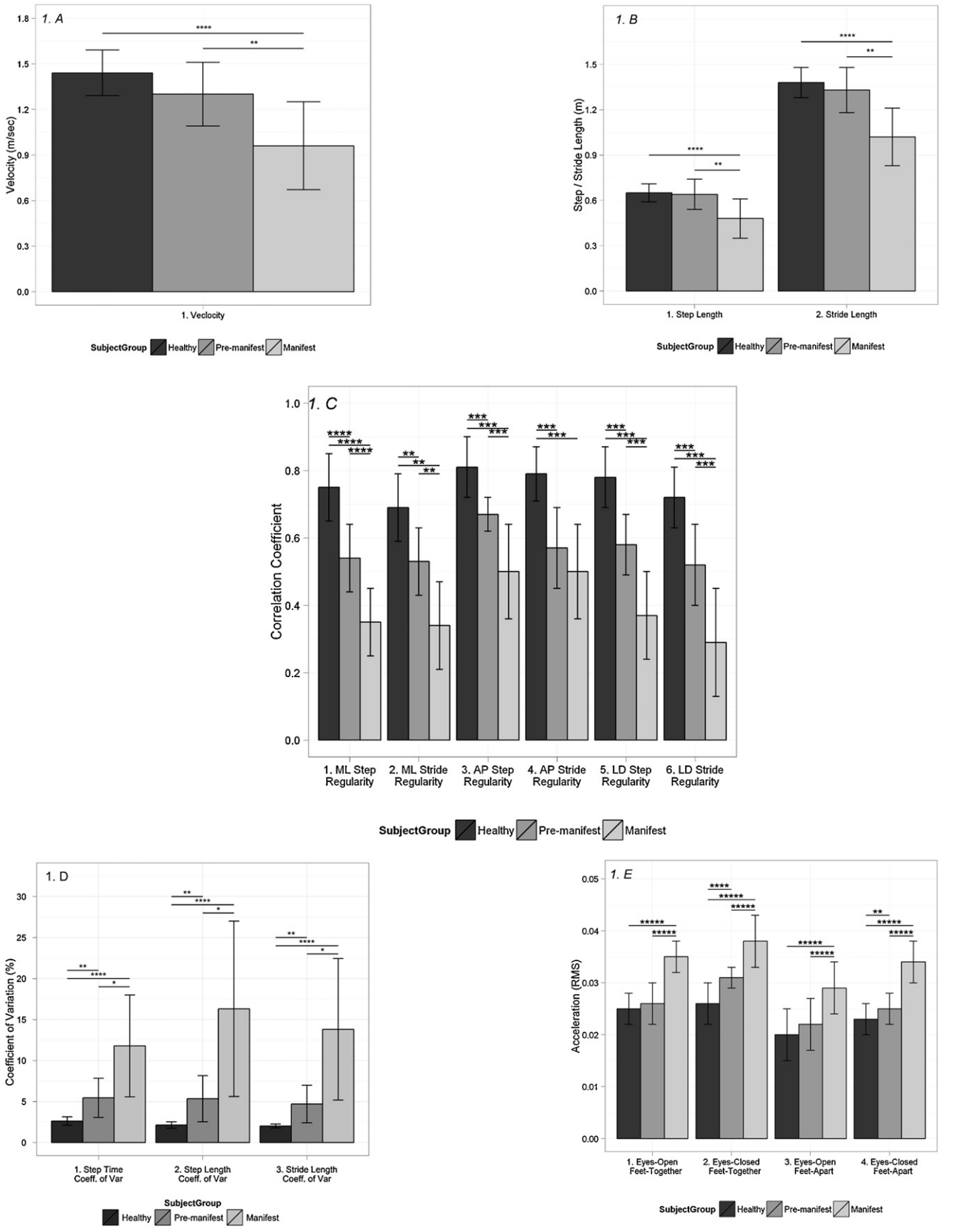

Fig. 1. Box plots (mean ± standard deviation) presenting the statistically significant parameters from the gait analysis. (A) velocity, (B) step length, stride length, (C) step and stride regularity across the ML, AP and LD axes respectively, (D) step time coefficient of variation, step length coefficient of variation and stride length coefficient of variation, (E) Romberg balance tasks (note the absolute values of ML step regularity are presented). Step regularity: first dominant peak from unbiased autocorrelation procedure (A_{d1}) Regularity of the acceleration between consecutive steps. Stride regularity: second dominant peak from unbiased (A_{d2}) autocorrelation procedure: regularity of the acceleration between consecutive strides $^{*}p < 0.01$; $^{**}p < 0.05$; $^{***}p < 0.002$; $^{****}p < 0.001$; $^{*****}p < 0.0001$.

specific height was performed however, we found this procedure had little impact on the raw data. We believe this can be attributed to the lack of dispersion among our sample height data and as noted by Hof et al. [23] in homogenous datasets the normalization procedure may have minimal effect.

Significant group differences were found for several gait parameters. Specifically, step time CV and stride length CV were significantly different across all subject groups. They results are consistent with studies of a larger sample size conducted by Rao et al. [6], Delval et al. [7] and Tabrizi et al. [26]. Our results did not reveal significant group differences between HLY and PHD for velocity ($p < 0.06$) and stride length ($p < 0.06$). Furthermore, cadence did not demonstrate any significant group differences. In contrast, both Delval et al. [7] and Rao et al. [6] found velocity and stride length to be significantly different across all subject groups. For cadence Delval et al. [7] found significant differences between HLY and PHD ($p < 0.01$) and Rao et al. [6] found significant differences between healthy subjects and subjects in stages two ($p < 0.01$) and three ($p < 0.003$) of the disease; we did not create subgroups by manifest HD stage.

Step and stride length decreased with each subject group. Furthermore, significant group differences were found across all gait regularity components and mediolateral step regularity demonstrated strong discriminative power. Although we were unable to find published work investigating step and stride regularity using the unbiased autocorrelation procedure and step time asymmetry in a HD population, these results correspond to published work employing alternative measurement techniques and varying populations. Rao et al. [6] noted that PHD subjects tended to spend a larger percentage of time in stance and have increased step time variability with the severity of these symptoms increasing with disease progression. Tura et al. [27] found that AP step regularity and LD stride regularity could distinguish between ten amputees and ten healthy subjects. Stride-to-stride variability has also proven useful in distinguishing between Parkinsonian subjects with freezing of gait and those without [28]. For step time asymmetry Bautmans et al. [12] also found no significant difference in a cohort of 121 elderly subjects.

For balance analysis the root mean square values increased with task complexity from Eyes-Open Feet-Apart to Eyes-Closed Feet-Together across all subject groups. Although, we were unable to find published work investigating accelerometry data for balance analysis in a HD population several studies have employed similar techniques in varying populations. In an investigation of postural instability, stance with feet close together was found to be a highly sensitive test in a sample of 20 HD subjects [29]. Using accelerometry O'Sullivan et al. [14] was able to distinguish between various test conditions except between that of Eyes-Open and Eyes-Closed. They speculate since both balance tasks only represent a marginally different challenge to balance control and as their subjects stood with feet apart, increased mediolateral stability was demonstrated. Our dataset tends to agree with this observation as a significant difference was found for Eyes-Open Feet-Apart vs. Eyes-Closed Feet-Together ($p < 0.0007$). Several studies investigated balance while standing on a mat or foam surface [14–17] and it is plausible that the sensitivity and specificity of our results might change with the addition of such a surface in the protocol.

The optimal thresholds from the ROC curve analysis, velocity (1.3142 m/s) and stride length (1.349 m), align with recent studies [6,7]. The parameters derived from both gait and balance demonstrated high sensitivity and specificity thus reinforcing the argument for inclusion of quantitative evaluation in clinical assessment of HD. However, we note the caution raised by Delval et al. [7]. With a large study population of 57 healthy subjects they found standard gait parameters were not sensitive enough to detect PHD status however, an increase in stride-to-stride variability was found sufficient to differentiate between PHD and HLY (sensitivity and specificity < 0.9). In comparison Tabrizi et al. [26] found GAITRite® stride length CV to be a sufficient biomarker in a very large study population. Although measures of step and stride regularity and balance were not investigated in these studies we recognize the necessity of large scale longitudinal studies employing accelerometry in a HD population.

Our site of attachment for the triaxial accelerometer was the thorax just below the suprasternal notch and this placement might be considered a limitation of the study. We chose this location as we wish to incorporate fall detection in future development. The majority of referenced work affixed a sensor over the L3 region of the spine which is close to where COM is believed to be during quiet standing [10]. From analysis of the data we note the accelerometry patterns at the level of the thorax are an attenuated version of those closer to the COM. This conclusion is also supported in the literature where accelerations measured at the level of the head correlate with and are dampened versions of those at inferior levels [30]. A significant advantage of our work is the incorporation of a high sampling frequency (Fs = 250 Hz). A low sampling rate increases the probability of temporal aliasing especially in heel strike detection [29] and a higher sampling rate (>100 Hz) could improve the accuracy of step/stride regularity and step-time asymmetry.

5. Conclusion

The objective of this study was to investigate the capacity of a single accelerometer based sensor to derive spatio-temporal parameters of gait and balance in a HD population and investigate its potential to differentiate between subject groups. The sensor showed excellent agreement to a computerized walkway across a range of spatio-temporal parameters and demonstrated significant discriminatory power between healthy, pre-manifest HD and manifest HD subjects across a host of gait and balance parameters. Specifically, we note the excellent discriminatory power of velocity, stride length coefficient of variation, ML step regularity and LD step regularity. These results highlight the sensors capacity to detect and quantify the variability in task performance across subject groups and its significant potential for employment in monitoring for disease onset in HD.

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Conflict of interest statement

None declared.

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