1 Postural control and gait measures derived from wearable inertial measurement unit devices in
2 Huntington's disease: recommendations for clinical outcomes
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42
43   **Abstract**
44
45   *Background*

46   Postural control impairments begin early in Huntington's disease yet measures most sensitive to

47   progression have not been identified. The aims of this study were to: 1) evaluate postural control

48   and gait in people with and without Huntington's disease using wearable sensors; and 2) identify

49   measures related to diagnosis and clinical severity.

50

51   *Methods*

52   41 individuals with Huntington's disease and 14 age-matched peers performed standing with feet

53   together and feet apart, sitting, and walking with wearable inertial sensors. One-way analysis of

54   variance determined differences in measures of postural control and gait between early and mid-

55   disease stage, and age-matched peers. A random forest analysis identified Features of Importance

56   for Huntington's disease diagnosis. Stepwise and ordinal regressions were used to determine

57   predictors of clinical chorea and tandem walking scores respectively.

58

59   *Findings*

60   There was a significant main effect for all postural control and gait measures comparing early

61   stage, mid stage and age-matched peers, except for gait cycle duration and step duration. Total

62   sway, root mean square and mean velocity during sitting, as well as gait speed had the greatest

63   importance in classifying disease status. Stepwise regression showed that root mean square

64   during standing with feet apart significantly predicted clinical measure of chorea, and ordinal

65   regression model showed that root mean square and total sway standing feet together

66   significantly predicted clinical measure of tandem walking.

67

68 *Interpretations*

69 Root mean square measures obtained in sitting and standing using wearable sensors have the

70 greatest potential to serve as biomarkers of postural control impairments in Huntington's disease.

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72 **Highlights**

73 • Sensor-derived postural control measures are discriminative in Huntington's disease

74 • Root mean square measures predict disease status and correlate to clinical measures

75 • Root mean square during sitting and standing are potential disease biomarkers

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83   **Introduction**

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85   Huntington's disease (HD) is a genetic neurodegenerative disorder marked by a triad of

86   motor, behavioral and cognitive manifestations. (Bunner and Rebec, 2016; Tian et al., 2012) As

87   a result of degeneration of the striatum, HD elicits hallmark motor deficits including impaired

88   postural control from early in the disease process. Postural control deficits in HD are

89   characterized by mis-scaling of accelerations and durations, and excessive excursions in both

90   static and dynamic contexts, (Delval et al., 2011; Moisello et al., 2011; Salomonczyk et al.,

91   2010) that interfere with stability. Although existing clinician-rated measures, such as items on

92   the United Huntington's Disease Rating Scale-Total Motor Score (UHDRS-TMS), (Kieburtz,

93   1996) can provide insight into general balance performance (e.g. tandem walking and

94   retropulsion pull tests), there remains a need for objective measures to accurately assess non-

95   observable aspects of postural control. Objective markers of postural control provide insight into

96   functional independence, and factor into important aspects of clinical decision making in HD,

97   such as fall risk and the likelihood of nursing home admission.(Wheelock et al., 2003)

98   Ultimately, precise and reliable measures can elucidate postural control impairments in HD and

99   function as a marker for disease progression. (Salomonczyk et al., 2010).

100   Postural control encompasses the ability of an individual to maintain equilibrium during

101   both static and dynamic tasks. (Dunsky et al., 2017; Lazarotto et al., 2020) During static tasks,

102   which include sitting and standing, the motor system prioritizes maintaining the center of mass

103   within the base of support to maintain stability. Dynamic tasks, such as walking and reaching,

104   challenge the system to integrate the modulation of center of mass within the constraints of a

105   volitional movement. Under both circumstances, the motor system aims to minimize the risk of

106   falls. In non-HD populations, properties of an individual's postural control, such as total sway

107   and trunk velocity and acceleration, are tempered and scaled to the demands of a task. (Horak,

108   2006) Previous work in HD demonstrated an inability to appropriately scale postural control in

109   the context of dynamic movements such as gait and reaching, (Delval et al., 2007; Mann et al.,

110   2012) which can ultimately contribute to an increase in fall risk. (Grimbergen et al., 2008)

111   Deficits in postural control are apparent early in HD and worsen with progressive striatal

112   degeneration, which both directly and indirectly influence the output of the reticulospinal,

113   vestibulospinal, and corticospinal tracts needed for voluntary postural control. (Jacobs and

114   Horak, 2007) A deeper understanding of specific properties of postural control that are most

115   impaired in persons with HD will lead to targeted rehabilitation programs for motor deficits that

116   negatively affect function and contribute to fall risk (Grimbergen et al., 2008).

117   The presence of involuntary movements, such as chorea, creates a further challenge in

118   measuring postural control in individuals with HD. Chorea consists of writhing involuntary

119   movements of the face, neck, upper extremities, lower extremities, or trunk, and can cause a shift

120   to the center of mass affecting postural control. (Berardelli et al., 1999) Chorea and postural

121   control are typically assessed using subjective clinical ratings from the UHDRS-TMS. This test

122   is inherently subjective, and ratings are typically non-linear, which limits the use of sensitive

123   parametric statistical methods. Clinical rating scales also only provide a brief 'snapshot' into any

124   disease, which remains problematic given the amount of heterogeneity of people with HD.

125   Quantitative analysis of movement has traditionally involved use of marker-based motion

126   capture camera systems or use of force plates to measure spatiotemporal features of gait and

127   postural control.(Miller et al., 2016) However, marker-based systems are costly, require an

128   extensive setup time, and often require advanced programming skills for data processing. More

129   recently, wearable sensors have been shown to be capable of providing reliable and valid

130   measurements of motor impairments, including postural control.(Kobsar et al., 2020; Simoes,

131   2011; Washabaugh et al., 2017) The benefits and utility of wearables has been demonstrated in

132   people with neurodegenerative diseases such as Parkinson's disease (PD) to measure tremor,

133   gait, and postural control. (Horak and Mancini, 2013; Washabaugh et al., 2017) Several recent

134   studies have also demonstrated their clinical utility in individuals with HD. (Dalton et al., 2013;

135   Porciuncula et al., 2020; Purcell et al., 2020; Tortelli et al., 2021; Trojaniello et al., 2015) IMUs

136   can measure discrete biomechanical properties of postural control quality and impairments, such

137   as jerk, total sway, and acceleration values, which may be more sensitive to motor impairments

138   than current clinical measures. (Adams et al., 2017; Andrzejewski et al., 2016)

139   In order to address this need for quantitative measures of postural control, we analyzed

140   postural control in sitting and standing conditions. The aims of this study were to 1) evaluate

141   differences in postural control and gait between non-HD individuals and those with HD from

142   wearable sensors; and 2) identify measures related to diagnosis and clinical measures of manifest

143   HD. The aims addressed in this study will ultimately provide recommendations for candidate

144   measures for use as clinical biomarkers.

145

146   **Methods**

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148   <u>Study Design</u>

149   This cross-sectional study evaluated participants during one session at three sites (George

150   Huntington Institute, Munster, Germany; Teachers College, Columbia University, New York,

151   USA and Wayne State University, Detroit, Michigan, USA). The study was approved by the

152   Ethics or Institutional Review Boards at each site. Cardiff University was the study sponsor and

153   the Wales Research Ethics Committee also approved the study protocol. The data presented here

154   was part of a larger study integrating wearable technologies to quantify meaningful activity in

155   Huntington's disease (iWEAR-HD).

156

157   Participants

158   Participants were included if they met the following inclusion criteria: 1) genetically

159   confirmed diagnosis of HD; 2) Total Functional Capacity (TFC) score ≥7; 3) 18 years of age or

160   older; and 4) able to walk 10 meters independently without assistance devices. Participants were

161   excluded if they had: 1) A diagnosis of juvenile onset HD; 2) history of comorbid neurological

162   conditions such as stroke or multiple sclerosis; 3) acute orthopedic conditions; and 4) the

163   inability or unwillingness of participant or legal guardian to give written informed consent.

164   Demographics of all participants are listed in Table 1.

165   TABLE 1 HERE

166

167   Procedures

168   UHDRS (TMS and TFC) (Kieburtz, 1996) were collected within three months of their

169   study visit. If this was not available, the UHDRS was administered as a part of the study visit.

170   The UHDRS TFC is a measure of functional ability, with scores ranging from 0 to 13, where

171   higher scores indicate greater functioning. HD1 (early stage) manifest-HD was categorized as

172   TFC scores of 11-13 while HD2 (mid stage) was TFC scores 7-10. These classifications were

173   based on previous research that aimed to identify imaging measures as clinical outcome

174   measures within HD. (Tabrizi et al., 2013) The UHDRS-TMS ranges in scores from 0 to 120,

175   with higher scores indicating a greater level of motor impairment. Each item is rated on a $0 - 4$

176   scale with 0 = Absent, 1 = Slight/ Intermittent, 2 = Mild/Common or Moderate/Intermittent, 3 =

177   Moderate/ Common, 4 = Marked/ Prolonged. The UHDRS chorea sub score was calculated from

178   the UHDRS-TMS by summing trunk and extremities chorea scores (5 items). The UHDRS

179   chorea sub score ranges from $0 - 20$, with lower scores indicating little to no chorea and higher

180   scores indicating greater chorea. Face and bucco-oral-lingual chorea scores were removed due to

181   their lack of association with the postural measures of interest. Chorea sub scores were used to

182   identify postural control measures that could best predict chorea. Clinical balance measures in

183   this study were represented by UHDRS tandem walking scores. Tandem walking is rated on a 0-

184   4 scale, where 0 indicates normal heel-to-toe walking for 10 steps and a score of 4 indicates an

185   inability to attempt the motor task.

186   Participants were fitted with six Opal V2 IMUs (Ambulatory Parkinson's Disease

187   Monitoring (APDM) opal sensors (ERT, Portland, OR, USA). Opal sensors have triaxial

188   accelerometers that measure linear acceleration, a gyroscope to measure angular velocity, and a

189   magnetometer for positional orientation.(Horak and Mancini, 2013). Data were processed using

190   Mobility Lab $2.0^{©}$ software (2015). Postural measures included total sway area, root mean

191   square (RMS) of sway, mean velocity and jerk. Total sway area signifies the extent of movement

192   around the center of gravity. RMS of sway is a time-domain measure, representing

193   the average variance. Mean velocity is the average rate of change in position during a certain

194   period of time. Jerk is a derivative of acceleration that reflects smoothness of movement. The

195   wearable sensors captured movement as participants sat quietly for 30 seconds in an armless

196   chair with their feet flat on the floor, hands resting in their lap, and their lower back away from

197   the back of the seat. Participants also stood quietly for 45 seconds with their feet hip width apart

198   (standardized using the APDM foot plate) and 45 seconds with their feet together. During the

199   gait task, participants walked 7 meters to a marking on the floor, turned around and returned to

200   the initial position. Gait variables that were extracted from Mobility Lab included gait speed,

201   percentage of time spent in stance and double support, and stride length.

202   Data Analysis

203   Statistical analyses were performed using R (Version 4.02; R Foundation for Statistical

204   Computing, 498 Vienna, AT). Two methods were used to determine the most sensitive measures

205   of dysfunction in postural control between individuals with HD and non-HD peers: Standardized

206   Mean Difference (*SMD*) and Random Forest. The *SMD* is a measure of effect size determined by

207   the following two equations:

208

209
$$SMD = \frac{\overline{X_1} - \overline{X_2}}{S}$$

210
211

212
$$S = \left(\frac{SD_1^2}{n_1} + \frac{SD_2^2}{n_2}\right) \times \left(\frac{n_1 \times n_2}{n_1 + n_2}\right)$$

213
214   where X_1 and X_2 are the means of the HD and non-HD groups, respectively. *S* is the pooled

215   standard deviation of the groups, where SD_1, n_1, SD_2, and n_2 are the standard deviation and

216   sample size of the HD group and the non-HD peer group, respectively. *SMD* values of >0.50

217   were retained as sensitive measures of UHDRS chorea sub score.

218   A Pearson correlation matrix was computed for participants with HD, which determined

219   associations between all postural control and gait variables, in addition to TMS, TFC, UHDRS

220   chorea subscore, and tandem walking score. A second Pearson correlation matrix was computed

221   for non-HD peers, which included all postural control and gait variables. Furthermore, between-

222   group differences in all postural control and gait variables were tested in the non-HD peer, HD1

223   (TFC<11), and HD2 (TFC 11-13) groups, using one-way analysis of variance (ANOVA). Post-

224   hoc differences were assessed using Tukey's test.

225   The random forest model was conducted by using the 'randomForest' package on R. The

226   random forest model is a machine learning approach that creates an ensemble of classification

227   and regression trees using a training dataset. In this experiment, a classification tree was

228   constructed using 80% of the total data. Once constructed, the remaining 20% of the data was

229   used to test the error rate of the classification tree. The random forest yields a variable ranking

230   metric termed 'mean decrease in Gini', which was used to represent features of importance of

231   predictor variables and was compared to *SMD*s. Mean decrease in Gini was also used for feature

232   selection to find a significant ordinal regression model to predict UHDRS tandem walking

233   scores.

234   Finally, we wanted to determine which measures derived from IMUs were best able to

235   predict clinical HD measures of chorea (UHDRS-TMS chorea subscore) and dynamic balance

236   (UHDRS-TMS tandem walking). To determine wearable measures capable of predicting chorea,

237   variables from the wearables (i.e., jerk, total sway area, RMS, mean velocity, gait speed, gait

238   stride length) of all postural control and gait conditions were entered into a forward stepwise

239   regression model, as chorea sub scores were treated as ratio data. Variables with

240   multicollinearity (VIF $\geq$ 10) were removed from the stepwise regression. Lowest Akaike

241   information criterion (AIC) was used for model comparison to determine the final regression

242   model. To determine measures capable of predicting UHDRS tandem walking scores, the

243   aforementioned variables of postural control and gait were entered into a random forest model.

244   The parameters with the highest mean decrease index of Gini were entered into an ordinal

245   regression model. Significance level was set at $P < 0.05$.

Results

246

247

248   There was a significant main effect for all postural control and gait measures comparing

249   HD1 and HD2 and non-HD peers, except for gait cycle duration and step duration ($P > .05$)

250   (Table 2). Post-hoc comparisons revealed a significant difference between the HD2 and non-HD

251   peers for all measures, except jerk feet together ($P > .05$). Post-hoc comparisons between HD1

252   and non-HD peers revealed differences in RMS sitting ($P = .022$). Further, there were significant

253   differences between HD1 and HD2 groups in RMS feet apart, RMS feet together, jerk feet apart,

254   jerk sitting, and gait speed (Table 2).

255

256   INSERT TABLE 2 HERE

257

258   *SMD*s that compared differences between each HD group to non-HD peers are shown in

259   Table 3. The highest *SMD*s found for both HD groups occurred in RMS feet apart (HD2: 19.94,

260   HD1: 5.58), RMS feet together (HD2: 19.65, HD1: 6.36), and step duration (HD2: 6.44, HD1:

261   5.87).

262   INSERT TABLE 3 HERE

263

264   Pearson correlations were calculated for both HD and non-HD peers groups separately

265   and are graphically represented in matrices found in Figure 1a and 1b.

266   INSERT FIGURE 1 HERE

267

268   Mean decrease in Gini derived from the random forest model showed that total sway

269   sitting (3.57), RMS sitting (2.22), mean velocity sitting (2.09), and gait speed (1.32) had the

270   greatest features of importance for accurate classification of individuals into either the HD or

271   non-HD group.

272   Stepwise regression revealed a significant linear model, shown in Figure 2 in which RMS

273   during standing feet apart (β = 9.19, P < .001) predicted chorea subscore (F (1, 38) = 11.74, P <

274   .0015, R^2 = 0.216). The random forest model revealed that RMS feet together (Odds Ratio: 6.74)

275   and total sway feet together (Odds Ratio: -0.002) predicted UHDRS tandem walking scores

276   (χ^2(2) = 13.38, P <.001, Nagalkerke R^2 = 0.30).

277

278   INSERT FIGURE 2 HERE

279

280   **Discussion**

281   Consistent with previous findings, our results indicated significant differences between HD

282   and non-HD peers across most variables of postural control that were measured within and across

283   conditions. (Porciuncula et al., 2020; Salomonczyk et al., 2010; Tian et al., 1991) Individuals with

284   HD performed significantly worse than non-HD peers on variables of postural control including

285   greater sway area, greater acceleration values in static standing and sitting, and greater jerk values

286   during quiet standing and sitting. These differences were more pronounced between non-HD peers

287   and HD individuals with TFC scores below 11 (HD2). We found that RMS during standing feet

288   apart best predicted UHDRS chorea subscore, and RMS and total sway during standing feet

289   together best predicted UHDRS tandem walking scores. RMS, total sway, and mean velocity

290   during sitting were postural measures most associated with an HD diagnosis. Given their ability to

291 predict preexisting clinical measures and HD diagnosis, we identified RMS measures in sitting

292 and standing as candidate biomarkers of postural control. Among RMS measures, we determined

293 that RMS during sitting is a suitable biomarker of postural control for its capacity to discriminate

294 between high TFC groups and non-HD peers, and its feasibility for data collection within in an

295 HD population.

296 RMS is the average composite value of acceleration of sway derived from the lumbar

297 sensor. The *SMD* of RMS between non-HD and HD groups indicated a moderate effect size across

298 all tasks (above .5) (Table 3). Interestingly, although RMS measures during standing successfully

299 discriminated between HD1 and HD2, RMS during sitting remained the only variable with the

300 discriminative capacity to discern between the HD (higher functioning) and non-HD peers.

301 Although jerk values exhibited acceptable *SMD* values and differences between low and high TFC

302 groups, within group variability was too large, as represented by large standard deviations, and

303 therefore not included in the random forest tree model.

304 Often HD individuals with higher TFC scores do not exhibit motor symptoms that can be

305 easily identified by clinicians. However, non-observable deficits in motor control are evident in

306 pre-manifest and early-stage HD, (Delval et al., 2011; Salomonczyk et al., 2010) which are more

307 easily captured by inertial sensors. (Andrzejewski et al., 2016; Fusca et al., 2018) Our findings on

308 RMS values during sitting further confirm this and introduce the idea that static sitting may elicit

309 more acceleration-based, measurable responses in force modulation impairments in HD. It is

310 possible that in standing, the motor system prioritizes stability and fall prevention by introducing

311 constraints to limit excessive sway. During sitting, the risk of fall is minimal relative to standing

312 thereby decreasing the demand of constraints needed to prioritize stability. The overall result is

313 that when seated, participants with HD could generate greater movement with less force

314   modulation without the risk of falling. Sitting minimizes the need for risk-related constraints,

315   leaving greater room for non-consequential errors in sway, force modulation, and chorea.

316   The UHDRS chorea subscore used in this study was comprised of scores assessing the

317   severity of chorea present in the trunk and extremities. Among the IMU-derived measures of

318   postural control our findings indicated that RMS and mean velocity during sitting predicted chorea

319   subscores. Aside from the association between lumbar total sway and visual trunk displacement,

320   it is likely that pronounced chorea in extremities would be associated with greater displacement of

321   the trunk due to postural disturbances and intersegmental dynamics. (Grimbergen et al., 2008;

322   Kegelmeyer et al., 2017) Therefore, RMS and mean velocity during sitting may function as an

323   indirect measure of chorea.

324   From our findings, we propose that RMS during sitting and standing derived from IMUs

325   can serve as clinical outcome measures for postural control in individuals with HD. These

326   measures are recommended due to their ability to predict both HD status and UHDRS tandem

327   walking scores. These measures also had relatively low variability and larger discriminative

328   validity, as determined by *SMD* values, relative to the remaining measures. Furthermore, the effect

329   sizes of RMS values are consistent with previous findings indicating that larger accelerations and

330   force productions within HD groups occur across motor tasks. (Moisello et al., 2011; Reilmann

331   and Schubert, 2017) Our results show that IMU sensors can capture force modulation impairments,

332   as represented by larger acceleration values, that are characteristic in HD. This further confirms

333   the notion that IMU wearable devices can detect subtle differences in postural control that cannot

334   be captured by clinical scales. The subjective nature of the UHDRS tandem walking score makes

335   it unlikely that this clinical measure could detect more subtle changes in postural control in HD.

336   Wearable-derived candidate measures of postural control were both associated with the

337 aforementioned clinical scales and captured non-observable and reliable measures of posture

338 specific to HD as compared to non-HD peers, and characteristic of the disease. The wearable-

339 derived outcome measures of postural control identified in this study can ultimately allow for the

340 ability to discriminate between disease stages and potentially to evaluate intervention effectiveness

341 as clinical endpoints.

342 Some limitations that can be addressed for future studies include the use of a single vs

343 multiple trials. Results from our study are from a single trial of each task across participants. This

344 may limit the generalizability as multiple trials can improve sensitivity of each measure. Future

345 work should also address the reliability of these measures over the course of HD, including a pre-

346 manifest HD group, where postural impairments are more subtle. Lastly, future work can

347 determine whether these postural measures are sensitive to intervention. Identifying response to

348 time with HD and intervention are integral next steps for evaluating the usability of these measures

349 in clinical trials.

350

351 **Conclusion**

352

353 RMS measures derived from IMUs during sitting and standing were most related to HD status

354 and severity and have the potential to serve as clinical outcome measures for postural control in

355 HD. Measures of total sway area of FA and sitting are likely influenced by choreic movements,

356 whereas RMS FT, jerk FA and RMS sitting are likely to best reflect postural control

357 impairments. Differences in postural control variables were more pronounced between non-HD

358 peers and HD individuals with TFC scores below 11. Future studies can determine sensitivity to

359 time and intervention in these postural control measures.

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508 Table 1. Participant Characteristics
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	All Participants with HD (n=43)	TFC 7 -10 (n=28)	TFC 11-13 (n=15)	Non-HD peers (n=15)
Age(years)	56.9 (11.7)	55.9 (9.6)	50.9 (14.2)	53.1 (13.3)
Sex (M/F)	25/19	14/14	11/4	7/7
Height (cm)	173.6 (9.6)	172.3(9.4)	175.7 (9.5)	169.2 (10.8)
Weight(kg)	75.0 (14.8)	76.4(16.3)	73.2 (12.1)	79.6 (12.1)
UHDRS TMS	40.6 (16.2)	48 (12)	27 (14)	NA
UHDRS SDMT	27.5 (12)	22 (9)	37.5 (11)	NA
UHDRS Chorea Subscore	10 (5)	12 (5)	9 (5)	NA
UHDRS FA	22 (3)	20 (3)	24 (1)	NA
UHDRS Ind.	84 (9)	80 (10)	90 (6)	NA
# of Fallers	24	19	5	NA

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511 Table 1: Demographics of all participants are represented as means(SD) except sex, which is
512 reported number of participants. Unified Huntington's Disease Rating Scale (UHDRS) total
513 motor score (TMS) is a sum of all sub scores rated on a 0-4 scale, where 0 represents an absence
514 of impairment and 4 represents a pronounced impairment. UHDRS Symbol Digit Modalities
515 Test (SDMT) measure executive functioning as represented by values correct from 0 to 110.
516 UHDRS Chorea sub score is a sum of sub scores of 0-4, where 0 represents an absence of chorea
517 and 4 represents pronounced presence, totaling in 28 points. UHDRS Functional Assessment
518 (FA) scores are out of 25, where larger values represent higher functioning, and independence
519 (Ind) scores, where higher scores represent greater independence, are rated 0-100.
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534 Table 2: Raw mean values, standard deviations, and coefficient of variation for all postural control
535 and gait variables in persons with HD and non-HD peers. Standardized mean difference is a
536 measure of effect size
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|		Non-HD Peers	HD (TFC11-13)	HD (TFC<11)	*SMD* (TFC11-13)	*SMD* (TFC<11)
RMS	Feet Apart	0.10(0.04)	0.21(0.19)	0.33(0.17)	5.58	19.94
	Feet Together	0.12(0.05)	0.25(0.19)	0.39(0.18)	6.36	19.65
	Sitting	0.06(0.05)	0.54(0.47)	0.63(0.60)	4.40	4.44
Jerk	Feet Apart	1.50(1.06)	22.0(37.6)	70.5(80.8)	0.029	0.030
	Feet Together	3.42(4.3)	19.6(25.4)	114.0(183)	0.049	0.0094
	Sitting	1.15(1.60)	36.7(31.8)	92.7(97.41)	0.070	0.028
Mean Vel.	Feet Apart	0.33(0.18)	0.45(0.41)	0.68 (0.49)	1.22	3.40
	Feet Together	0.32(0.15)	0.61(0.58)	0.80 (0.58)	1.59	3.63
	Sitting	0.11(0.11)	0.80 (0.71)	1.21 (1.24)	2.71	2.02
Total Sway	Feet Apart	1.80 (1.8)	14.8 (24.8)	36.6 (39.8)	0.042	0.062
	Feet Together	4.06(4.0)	20.4(27.0)	46.4 (51.7)	0.044	0.045
	Sitting	0.29 (0.37)	56.9 (66.8)	79.9(105)	0.025	0.021
Gait	Cycle Duration	1.06(0.12)	1.09(0.09)	1.10(0.13)	3.18	3.70
	Speed	1.32(0.26)	1.22(0.20)	1.04(0.21)	-1.84	-4.66
	Step Duration	0.53(0.06)	0.55(0.04)	0.56(0.07)	5.87	6.44
	Stride Length	1.37(0.19)	1.31(0.15)	1.14(0.19)	-1.76	-6.13

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559 Table 3. Comparison of postural control and gait variables between Low TFC (LTFC; TFC <11),
560 High TFC (HTFC; TFC≥11) and non-HD peer, using ANOVA and Tukey post-hoc pairwise
561 comparison. * Denotes a main significant effect of group. Only significant pairwise post-hoc
562 tests reported in table. D.F = Degrees of Freedom.

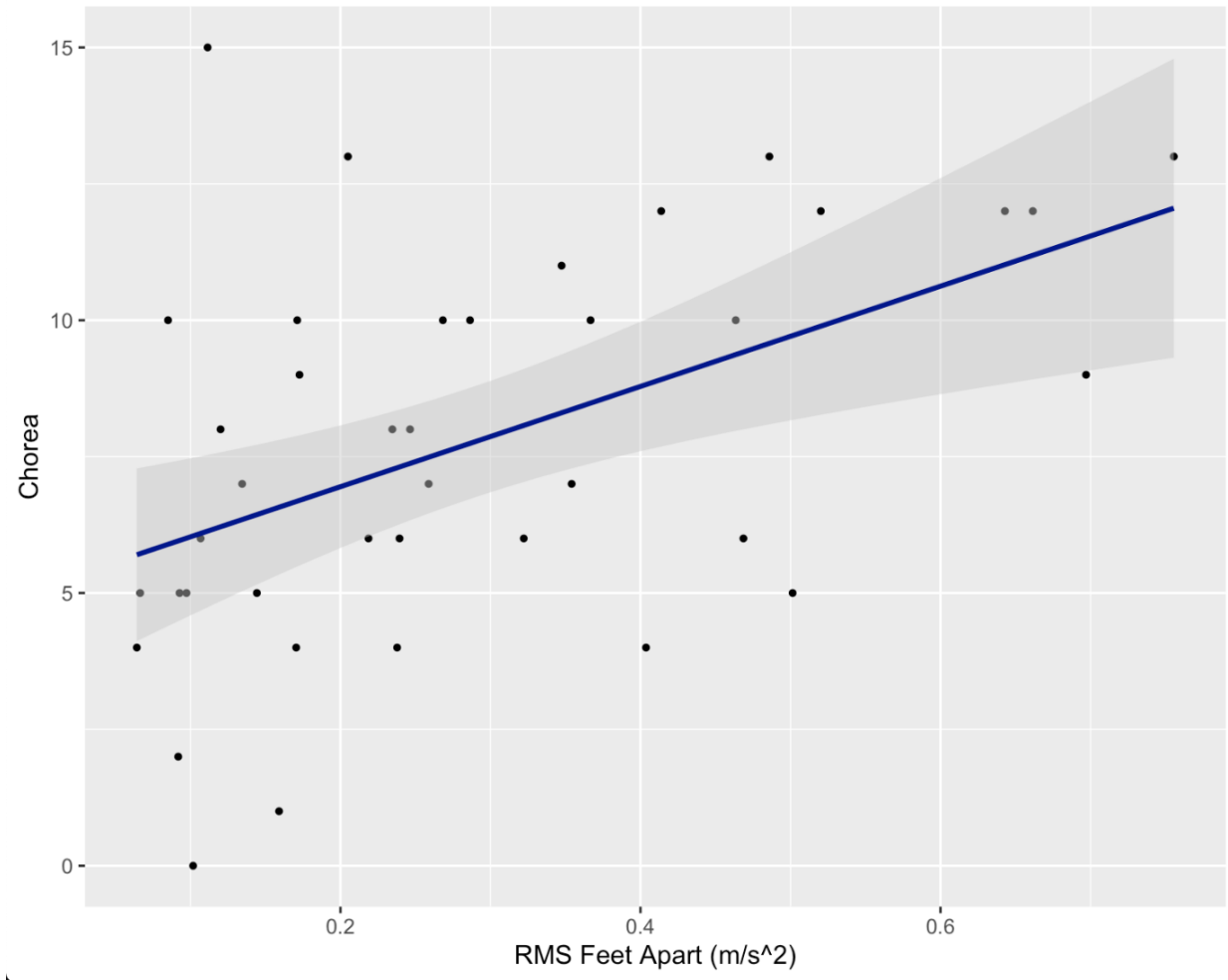|		D.F.	F-value	p-value	Tukey post-hoc
RMS	Feet Apart	(2,54)	11.13	<.001*	HD 2-HD 1: p = .04 HD 2-Non HD: p <.001
	Feet Together	(2,54)	13.24	<.001*	HD 2-HD 1: p = .027 HD 2-Non HD: p < .001
	Sitting	(2,55)	7.036	.0019*	HD 2-HD 1: p = .0016 HD 1-Non HD: p = .022
Jerk	Feet Apart	(2,54)	7.457	.0014*	HD 2-HD 1: p = .036 HD 2-Non HD: p =.0018
	Feet Together	(2,54)	4.593	.014*	HD 2-Non HD: p = .026
	Sitting	(2,55)	8.974	<.001*	HD 2-HD 1: p = .041 HD 2-Non HD: p < .001
Mean Vel.	Feet Apart	(2,54)	3.93	.026*	HD 2-Non HD: p = .027
	Feet Together	(2,54)	4.355	.0018*	HD 2-Non HD: p = .013
	Sitting	(2,55)	6.44	.0026*	HD 2-Non HD: p =.0017
Total Sway	Feet Apart	(2,54)	6.845	.0022*	HD 2-Non HD: p =.0022
	Feet Together	(2,54)	6.289	.0035*	HD 2-Non HD: p =.0034
	Sitting	(2,55)	4.742	.0126*	HD 2-Non HD: p =.0091
Gait	Cycle Duration	(2,54)	1.014	0.37	–
	Speed	(2,54)	8.271	<.001*	HD 2-HD 1: p = .04 HD 2-Non HD: p <.001
	Step Duration	(2,54)	0.862	.428	–
	Stride Length	(2,54)	8.881	<.001*	HD 2-Non HD: p <.001

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Figure 1. a) Pearson correlation tables of all IMU derived postural control and gait, in addition to clinical measures, and UHDRS chorea subscores for patients with HD. b) Pearson correlation tables of all IMU derived postural control and gait for non-HD peers.
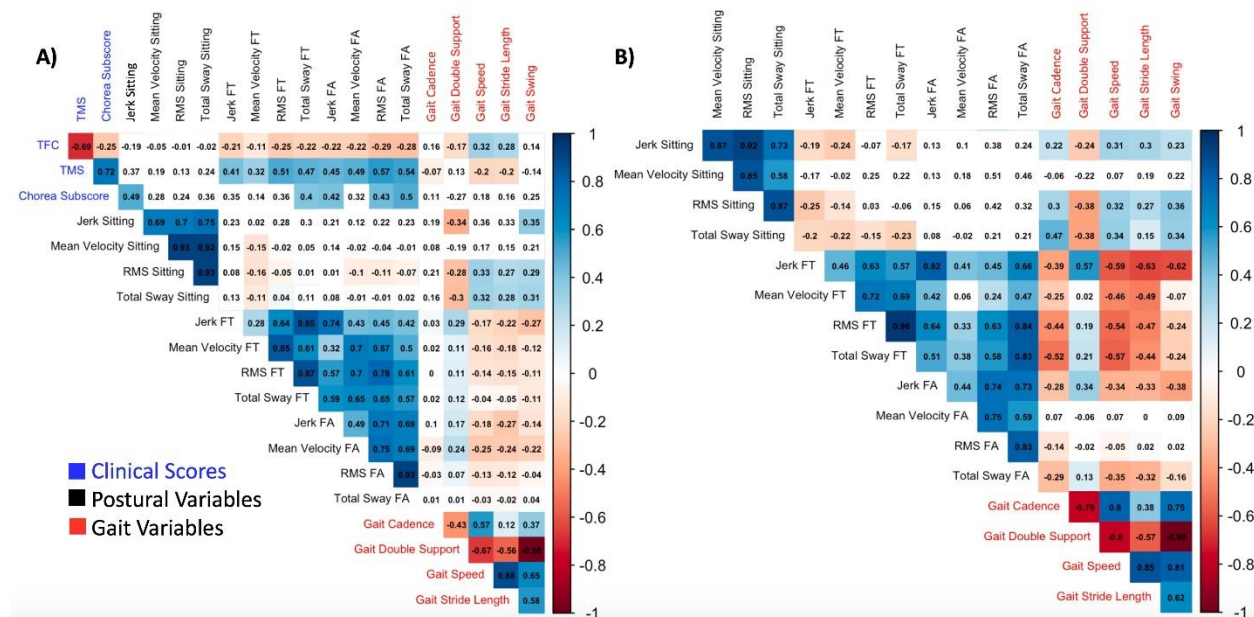
590   Figure 2: Linear regression plot indicating significant model, where UHDRS Chorea sub score is
591   predicted by Root Mean Square (RMS) Feet Apart.
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