Published in final edited form as:

Gait Posture. 2013 May ; 38(1): 109–114. doi:10.1016/j.gaitpost.2012.10.025.

Quantitative analysis of gait and balance response to deep brain stimulation in Parkinson's disease

Thomas O. Mera, MS^{a,*}, Danielle E. Filipkowski^a, David E. Riley, MD^b, Christina M. Whitney, PhD^b, Benjamin L. Walter, MD^b, Steven A. Gunzler, MD^b, and Joseph P. Giuffrida, PhD^a

^aDivision of Movement Disorders, Great Lakes NeuroTechnologies Inc., 10055 Sweet Valley Drive, Cleveland, Ohio, 44125, USA

^bMovement Disorders Center, Neurological Institute, University Hospitals and Case Western Reserve University School of Medicine, 1611 South Green Road Suite 204, South Euclid, OH, 44121, USA

Abstract

Gait and balance disturbances in Parkinson's disease (PD) can be debilitating and may lead to increased fall risk. Deep brain stimulation (DBS) is a treatment option once therapeutic benefits from medication are limited due to motor fluctuations and dyskinesia. Optimizing DBS parameters for gait and balance can be significantly more challenging than for other PD motor symptoms. Furthermore, inter-rater reliability of the standard clinical PD assessment scale, Unified Parkinson's Disease Rating Scale (UPDRS), may introduce bias and washout important features of gait and balance that may respond differently to PD therapies. Study objectives were to evaluate clinician UPDRS gait and balance scoring inter-rater reliability, UPDRS sensitivity to different aspects of gait and balance, and how kinematic features extracted from motion sensor data respond to stimulation. Forty-two subjects diagnosed with PD were recruited with varying degrees of gait and balance impairment. All subjects had been prescribed dopaminergic medication, and twenty subjects had previously undergone DBS surgery. Subjects performed seven items of the gait and balance subset of the UPDRS while wearing motion sensors on the sternum and each heel and thigh. Inter-rater reliability varied by UPDRS item. Correlation coefficients between at least one kinematic feature for six of the seven items and UPDRS were greater than 0.75. Kinematic

© 2012 Elsevier B.V. All rights reserved.

***Address Correspondence to:** Thomas Mera, MS, Great Lakes NeuroTechnologies Inc., 10055 Sweet Valley Drive, Cleveland, Ohio 44125, USA, Phone: 001 216-361-5410, Fax: 001 216-361-5420, tmera@glneurotech.com.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

Conflicts of Interest Statement

- !! Thomas Mera, MS is a paid employee of Great Lakes NeuroTechnologies Inc.
- !! Danielle Filipkowski, BS is a paid employee of Great Lakes NeuroTechnologies Inc.
- !! David Riley, MD claims no conflict of interest related to this research.
- !! Christina Whitney, PhD claims no conflict of interest related to this research.
- !! Benjamin Walter, MD receives research funding from NIH, and is a consultant for Deringer-Ney, Glaxosmithkline, Teva, Medtronic, Novartis, Ipsen, and Boehringer Ingelheim
- !! Steven Gunzler, MD received research funding from Allon Therapeutics Inc., Synosia Therapeutics Inc. (now Biotie Therapeutics Inc.), and the Kinetics Foundation.
- !! Joseph Giuffrida, PhD is a paid employee of Great Lakes NeuroTechnologies Inc.

features improved ($p < 0.05$) from DBS-OFF to DBS-ON for three UPDRS items. Despite achieving high correlations with the UPDRS, evaluating individual kinematic features may help address inter-rater reliability issues and rater bias associated with focusing on different aspects of a motor task.

1. Introduction

Parkinson's disease (PD) causes both motor and non-motor signs, but is primarily recognized by resting tremor, rigidity, and bradykinesia (slowness of movement) [1]. However, other manifestations such as balance and gait disturbances, especially in advanced patients, can be debilitating and may lead to increased fall risk [2].

As PD advances and patients become potential candidates for deep brain stimulation (DBS), targeted symptoms for surgical management typically include tremor and medication-responsive bradykinesia, rigidity, and freezing of gait in the presence of intolerable control due to motor fluctuations and dyskinesia. In general, compromised gait and balance is more difficult to manage with DBS, and as patients advance into later stages of the disease, these impairments become more common and problematic. The same set of stimulation parameters may not address all of the cardinal motor signs of PD regardless of disease severity. In addition, as severity increases, response to specific stimulation parameters may change. Strategies to better treat these more complex symptoms have been proposed. Moreau, et. al. demonstrated that high-frequency ($>100\text{Hz}$) stimulation improved tremor and bradykinesia, but was not as effective for gait and balance. In contrast, low-frequency (60Hz) stimulation with proportionally higher-amplitude settings resulted in greater improvement in gait [3]. Currently, there are no standardized DBS programming guidelines for gait and balance. To further this point, in a study of 108 DBS patients, primarily diagnosed with PD, subjects sought referral to a movement disorder specialist after experiencing unsatisfactory symptomatic benefit. Gait and balance was the second leading complaint (34.3%) [4]. Therefore, developing improved neuromodulation tools for improving gait and balance is still needed. This may be especially beneficial as new stimulation targets are being investigated including the pedunculopontine nucleus; a brain region believed to play an important role in locomotion function, specifically initiation and modulation of gait [5–7].

A wide range of clinical rating scales have been developed to quantify and rate gait and balance impairment. The Unified Parkinson's Disease Rating Scale (UPDRS) is the standard clinical evaluation for PD, including DBS programming. A study conducted by Kerr evaluated multiple gait and balance clinical scales and their ability predict fall risk, a measure of functional gait impairment [8]. The motor section of the UPDRS (UPDRS-III) motor examination differentiated between fallers and non-fallers ($p < 0.012$) with a sensitivity and specificity of 0.64 and 0.60, respectively. The Tinetti Mobility Test (TMT) was predictive of prospective falls ($p < 0.001$) with a sensitivity and specificity of 0.67 and 0.59, respectively. Kegelmeyer also reported that the TMT correlated with the UPDRS-III ($r = -0.45$, $p < 0.05$) and resulted in a sensitivity and specificity of 0.76 and 0.66, respectively [9].

Although clinical rating scales have been shown to correlate with fall risk, inter-rater reliability may introduce bias and washout important features of gait and balance that may respond differently to PD therapies. Espay demonstrated that clinicians differentially weighed amplitude and speed deficits when rating UPDRS-III upper extremity bradykinesia tasks [10]. It was also shown that angular velocity, a measure of speed, improved more in

response to dopaminergic medication than excursion angle or variability in tapping angular velocity, measures of amplitude and rhythm, respectively [11].

Motion sensor technology has been previously commercialized as a general activity monitor to quantify gait. The ActivPal (Pal Technologies, Scotland, UK), AMP 331 (Dynastream Innovations, Alberta, Canada), and StepWatch (Orthocare Innovations, Seattle, WA) are examples of single unit accelerometer sensors which output various gait measures (e.g. cadence, walking speed). In contrast, we investigated a motion capture system that utilized multiple upper and lower body-worn motion sensor units to capture kinematic features specific to the gait and balance task subset of the UPDRS and changes in these features in response to DBS. Three hypotheses were tested: 1) UPDRS gait and balance items are compromised by lack of inter-rater reliability, 2) UPDRS is not sensitive enough to capture specific aspects of gait and balance, and 3) sensor-based kinematic features significantly change in response to DBS during a typical clinical visit.

2. Methods

This work was approved by the institutional review board in which this study was performed, and all subjects gave prior informed consent. All clinical investigations were performed in accordance with the ethical standards of the Declaration of Helsinki (2008).

2.1 Technology overview

KinetiSense™ (Great Lakes NeuroTechnologies Inc., Cleveland, OH) captures synchronized three-dimensional kinematic data from five sensor units (Figure 1). Each sensor unit contains three orthogonal accelerometers for measuring linear acceleration (a_x , a_y , and a_z) and three orthogonal gyroscopes for measuring angular velocity (ω_x , ω_y , and ω_z). The command module provides power and streams data to a computer at 128 samples per second.

2.2 Data collection protocol

Forty-two subjects (31 males, 11 females, 67 ± 13 years old) diagnosed with idiopathic PD in accordance with UKPDS Brain Bank criteria [12] with varying gait and balance impairment were recruited from a movement disorders center (Cleveland, OH). All subjects were on dopaminergic medication during the study and twenty subjects had also previously undergone DBS surgery. A motion sensor unit was positioned on each heel using a u-shaped shoehorn, on the central anterior aspect of each quadriceps muscle using an elastic Velcro strap, and on the base on the sternum using double-sided adhesive electrode washers (Figure 1).

The movement disorder center clinician or nurse practitioner guided subjects through seven items of the UPDRS-III to evaluate gait and balance impairment: 1) Toe Tapping, 2) Leg Agility, 3) Arising from Chair, 4) Gait, 5) Freezing of Gait, 6) Postural Stability, and 7) Posture [13]. Non-DBS subjects were each evaluated once while DBS subjects were evaluated twice, first with stimulation turned on using parameters from the previous clinical visit and again within a minute of turning the stimulation off, typical of the time allotted in a routine clinical visit. While motor tasks were performed, kinematic data were transmitted wirelessly and recorded to a laptop, and subjects were videotaped for later clinician scoring.

These videos were randomized and loaded onto a secure online server. Three movement disorder neurologists were blinded to stimulation state and rated the videos per UPDRS guidelines. Scores were averaged across raters to minimize variability. Agreement between clinicians was measured based on the correlation coefficient between scores given by each combination of two of the three clinical raters.

2.3 Kinematic feature extraction

Kinematic data collected from the KinetiSense device were processed into a range of quantitative features based on our previous studies evaluating speed, amplitude, and rhythm deficits of upper extremity hand movement motor tasks [10, 11] and other work using sensors to quantify gait [14, 15]. These studies also showed that variability over time and the logarithm of these features were correlated to clinician ratings and indicators of gait impairment. Correlation coefficients were calculated between the quantitative features and mean clinician UDPRS scores. A paired t-test compared motor severity in DBS-OFF and DBS-ON states for each kinematic feature.

Item 1 Toe Tapping—Three variables were calculated from motion data collected by the heel sensor during ankle flexion and extension. Data were low-pass filtered at 10Hz (Butterworth, 2nd order) to remove high-frequency noise. Log root mean square (RMS) was calculated for 1) angular velocity and 2) angle after integrating and high-pass filtering at 0.3Hz (Butterworth, 2nd order) to remove drift. 3) The coefficient of variation (CV) of toe tap cadence was calculated from peaks in the angle signal using a threshold at half of the RMS angle.

Item 2 Leg Agility—Six variables were calculated from motion data collected by the heel and thigh sensors during vertical leg lifts. Data were low-pass filtered at 10Hz (Butterworth, 2nd order) to remove high frequency noise. Log RMS was calculated for heel 1) linear velocity after integrating linear acceleration and high-pass filtering at 0.3Hz (Butterworth, 2nd order) to remove drift and 2) amplitude from double integration. 3) The coefficient of variation (CV) of time duration between leg lifts was calculated from peaks in the linear velocity signal using a threshold at half of the RMS velocity. Log RMS was calculated for thigh 4) angular velocity and 5) angle after integrating and high-pass filtering at 0.3Hz (Butterworth, 2nd order) to remove drift. 6) The coefficient of variation (CV) of time duration between leg lifts was calculated from peaks in the angle signal using a threshold at half of the RMS angle.

Item 3 Arising from Chair—Two variables were calculated from motion data collected by the sternum sensor. 1) The peak angular velocity was calculated as a measure of speed when standing up from the chair. 2) Task difficulty is associated with the number of forward and back sternum movements necessary to stand up. Therefore, data were low-pass filtered at 2Hz (Butterworth, 2nd order) to remove all activity other than the gross movement of the sternum, and the number of times angular velocity changed from positive to negative was calculated.

Item 4 Gait/ Item 5 Freezing of Gait (FOG)—Ten variables were calculated from motion data collected by the heel, thigh, and sternum sensors. Data collected by the heel sensor was low-pass filtered at 5Hz (Butterworth, 2nd order) to remove high frequency noise. Each forward step was identified by marking the peaks in angular velocity from the leg swing using a threshold of half the RMS angular velocity. The mean and CV were calculated for (1–2 leg swing peak angular velocity, (3–4 leg swing range of rotation, and (5–6 time between leg swings.

Four additional variables were calculated during the turn and the first step following the turn. 7) Peak angular velocity about the vertical axis of the sternum during the turn was calculated after low-pass filtering at 0.5Hz (Butterworth, 2nd order). 8) Turn time was calculated by integrating angular velocity and measuring the time required to rotate the sternum from 20 to 160 degrees. 9) The time delay to initiate a step after completing the turn and taking the first step forward was calculated. 10) The step count required to complete the

turn was calculated by marking the number of peaks greater than 50 degrees/second in the heel sensor.

Item 6 Postural Stability—Two variables were calculated from motion data collected by the sternum and heel sensors. 1) Reaction time was calculated from the time of the initial pull as measured by the sternum sensor to the first left or right step backward from the heel accelerometer. 2) Steps required after the initial pull to recover balance was calculated using a threshold at 1.25 meters/second² of the left and right heel accelerometers.

Item 7 Posture—One variable was calculated from motion data collected by the sternum sensor. Angle of body inclination angle relative to gravity was calculated from the inverse sine of the mean of the accelerometer normal to the sternum.

3. Results

Inter-rater reliability (Hypothesis 1) was assessed for each pair combination of the three clinician UPDRS raters. Mean correlation coefficients across the three comparisons were the lowest for items 1 ($r = 0.64$), 2 ($r = 0.55$), and 7 ($r = 0.62$) while agreement of the remaining four items (3, 4, 5, and 6) was 0.89, 0.78, 0.82, and 0.95, respectively.

Correlations between kinematic features and UPDRS scores (Hypothesis 2) were evaluated by first analyzing sensor-based gait and balance kinematics during UPDRS motor tasks. Figure 2 illustrates how items 4 and 5 were segmented into walking and turning in order to calculate task-specific features. Correlation coefficients between kinematic features and mean UPDRS clinician scores varied greatly (Table 1). Body inclination angle for item 7 resulted in the lowest correlation to the UPDRS ($r = 0.1$) while mean leg swing range of rotation for item 4 resulted in the best correlation ($r = -0.89$). Regression plots for items 1–6 display the mean clinician UPDRS score versus the most correlated kinematic feature (Figure 3).

Changes in kinematic features between DBS-OFF and DBS-ON (Hypothesis 3) were also tested (Table 2). Items 1, 2, and 4 had at least one feature that significantly changed ($p < 0.05$).

4. Discussion

The first hypothesis tested whether the UPDRS gait and balance items were affected by inter-rater reliability. Results demonstrated a large range of correlation coefficient values between clinicians, which may introduce rater bias and washout important features of gait and balance. This may most likely be attributed to raters prioritizing different aspects of UPDRS motor tasks and as a result make comparing scores across patients particularly challenging.

The second hypothesis tested whether the UPDRS was sensitive to specific aspects of gait and balance impairment. In favor of the UPDRS, specific clinically-relevant kinematic features were highly correlated with clinician scores [14, 16]. The item 4 feature, time between steps, was poorly correlated to the UPDRS ($r = 0.18$), while the coefficient of variability, a measure of stability, was more correlated ($r = 0.65$). Hausdorff, et. al. showed that stride-to-stride fluctuation was a better indicator of fall risk than stride time [15]. Clinician scoring was also sensitive to freezing of gait characterized by stutter stepping or difficulty initiating a step, which would increase turn time. Although the same kinematic features were calculated for item 4 (gait) and item 5 (freezing of gait), the correlation coefficient between the UPDRS and turn time was greater for item 5.

In opposition to the UPDRS, the coefficient of variation, a measure of rhythm or hesitations, was consistently less correlated to the UPDRS than other kinematic features such as log RMS angular velocity and angle, measures of speed and amplitude deficits, respectively (Table 1). The same can be said for item 7. While raters demonstrated good agreement ($r = 0.62$), UPDRS scores and the sternum inclination feature were poorly correlated ($r = 0.1$). Especially if PD treatments differentially affect individual kinematic features, combining them into a single score may washout clinically-relevant changes in gait and balance impairment.

Another limitation is the logarithmic nature of visual rating scales and the reduced sensitivity to just-noticeable differences at higher severities [17]. Therefore, the logarithm was calculated for several kinematic features including item 1 heel angle and item 2 thigh angular velocity, which improved correlations to the UPDRS. This observation has also been shown for upper extremity tremor [18] and bradykinesia [10].

The third hypothesis tested whether DBS significantly changed sensor-based kinematic features during a typical clinical visit. Of the three items responsive to stimulation ($p < 0.05$), all features increased or decreased as expected. For example, item 1 log RMS angular velocity and angle and item 4 and 5 leg swing angular velocity increased in response to DBS, while variability in these features decreased.

The non-significant changes from the other kinematic features may be attributed to limitations of this study. First, subjects were evaluated with their usual medication regimen and not in a practically defined ‘off’ therapy state. Therefore, withholding PD medication would have likely produced greater changes in response to stimulation. However, these study conditions more closely resemble real-life programming for long-term follow-up where patients are not as frequently programmed in the ‘off’ therapy state as compared to the initial programming following surgery. Second, stimulation settings may not have been optimized for gait and balance. Programming sessions often involve fixing frequency around 130Hz [3], an effective setting for tremor and bradykinesia, and varying voltage to maximize therapeutic benefit. However, gait improvements are more responsive to lower stimulation frequencies [3]. This also leads into the third limitation of this study and in-clinic programming in general. The short time between turning stimulation off and repeating the assessment may not have been sufficient to capture the full gait and balance response. While tremor typically improves within minutes and bradykinesia within 15–30 minutes after modulating DBS parameters, gait and balance changes can exceed three hours [19]. This may help explain the discrepancy in p-values between items 1–2, which evaluate lower extremity bradykinesia, and the remaining items in Table 2 since this study was limited by the clinician’s available time.

Another study limitation was the positioning of motion sensors. Specifically, item 7 was poorly correlated to the UPDRS and showed no change to DBS, most likely because the sternum position was not sensitive to varying degrees of postural inclination. Posture deficits may still be present when the torso is vertical, but the neck is angled forward. Repositioning the sensor on the upper midline back may better capture forward spinal flexion without being confounded by voluntary head movements.

A variety of motion sensor systems have been developed to quantify gait activity. The KinetiSense device used in this study provided synchronized data acquisition from five motion sensor units, each containing triaxial accelerometers and gyroscopes, to capture body movements specific to each of the UPDRS items. In contrast, most other devices [20] including the Activpal, Amp 331, and StepWatch are limited to just one accelerometer-based motion sensor. These systems were designed as general activity monitors while the

present study investigated PD-specific gait attributes such as freezing of gait, which can manifest during turns and before initiating forward stepping, not just general cadence measures. Angular velocity recorded by the KinetiSense gyroscopes may also better characterize natural limb rotation during movement. Furthermore, KinetiSense-based algorithms have been validated in other studies to quantify upper extremity PD tremor and bradykinesia [10, 11, 18], which when combined with lower-limb measures presented here, can provide a more complete assessment of PD symptom response to treatment.

This study demonstrated the limited inter-rater reliability of the UPDRS and how features extracted from body-worn sensors were more sensitive to specific aspects of gait and balance than the UPDRS. While some kinematic features did respond within a minute of turning stimulation off, in order to capture the full response, clinicians simply cannot evaluate a full range of stimulation settings during a single office visit. Therefore, in-clinic assessments may not reflect the patient's experience at home or capture motor fluctuations. In the future, motion sensors could be worn in the home to provide a more accurate measure of response to PD therapies including DBS.

Acknowledgments

This work was supported by the National Institutes of Health, National Institute on Aging, 5R43AG033947. We would like to acknowledge Dustin Heldman, PhD, for his conscientious assistance with editing the manuscript and Ellen Walter, CNP, for her assistance with subject recruitment.

Literature references

1. Pflanzner, RG.; Rhoades, R. Human Physiology. 3rd ed.. Forth Worth: Saunders College; 1996.
2. Cattaneo D, De Nuzzo C, Fascia T, Macalli M, Pisoni I, Cardini R. Risks of falls in subjects with multiple sclerosis. Arch Phys Med Rehabil. 2002; 83:864–867. [PubMed: 12048669]
3. Moreau C, Defebvre L, Destee A, Bleuse S, Clement F, Blatt JL, et al. STN-DBS frequency effects on freezing of gait in advanced Parkinson disease. Neurology. 2008; 71:80–84. [PubMed: 18420482]
4. Kluger BM, Foote KD, Jacobson CE, Okun MS. Lessons learned from a large single center cohort of patients referred for DBS management. Parkinsonism Relat Disord. 2011; 17:236–239. [PubMed: 20627660]
5. Garcia-Rill E, Houser CR, Skinner RD, Smith W, Woodward DJ. Locomotion-inducing sites in the vicinity of the pedunculopontine nucleus. Brain Res Bull. 1987; 18:731–738. [PubMed: 3304544]
6. Pahapill PA, Lozano AM. The pedunculopontine nucleus and Parkinson's disease. Brain. 2000; 123:1767–1783. [PubMed: 10960043]
7. Skinner RD, Kinjo N, Henderson V, Garcia-Rill E. Locomotor projections from the pedunculopontine nucleus to the spinal cord. Neuroreport. 1990; 1:183–186. [PubMed: 2129877]
8. Kerr GK, Worringham CJ, Cole MH, Lacherez PF, Wood JM, Silburn PA. Predictors of future falls in Parkinson disease. Neurology. 2010; 75:116–124. [PubMed: 20574039]
9. Kegelmeyer DA, Kloos AD, Thomas KM, Kostyk SK. Reliability and validity of the Tinetti Mobility Test for individuals with Parkinson disease. Phys Ther. 2007; 87:1369–1378. [PubMed: 17684089]
10. Heldman DA, Giuffrida JP, Chen R, Payne M, Mazzella F, Duker AP, et al. The modified bradykinesia rating scale for Parkinson's disease: reliability and comparison with kinematic measures. Mov Disord. 2011; 26:1859–1863. [PubMed: 21538531]
11. Espay AJ, Giuffrida JP, Chen R, Payne M, Mazzella F, Dunn E, et al. Differential response of speed, amplitude, and rhythm to dopaminergic medications in Parkinson's disease. Mov Disord. 2011; 26:2504–2508. [PubMed: 21953789]
12. Gibb WR, Lees AJ. The relevance of the Lewy body to the pathogenesis of idiopathic Parkinson's disease. J Neurol Neurosurg Psychiatry. 1988; 51:745–752. [PubMed: 2841426]

13. Goetz CG, Tilley BC, Shaftman SR, Stebbins GT, Fahn S, Martinez-Martin P, et al. Movement Disorder Society-sponsored revision of the Unified Parkinson's Disease Rating Scale (MDS-UPDRS): scale presentation and clinimetric testing results. *Mov Disord.* 2008; 23:2129–2170. [PubMed: 19025984]
14. Salarian A, Russmann H, Vingerhoets FJ, Dehollain C, Blanc Y, Burkhard PR, et al. Gait assessment in Parkinson's disease: toward an ambulatory system for long-term monitoring. *IEEE Trans Biomed Eng.* 2004; 51:1434–1443. [PubMed: 15311830]
15. Hausdorff JM. Gait variability: methods, modeling and meaning. *J Neuroeng Rehabil.* 2005; 2:19. [PubMed: 16033650]
16. Schaafsma JD, Giladi N, Balash Y, Bartels AL, Gurevich T, Hausdorff JM. Gait dynamics in Parkinson's disease: relationship to Parkinsonian features, falls and response to levodopa. *J Neurol Sci.* 2003; 212:47–53. [PubMed: 12809998]
17. Elble RJ, Pullman SL, Matsumoto JY, Raethjen J, Deuschl G, Tintner R, et al. Tremor amplitude is logarithmically related to 4- and 5-point tremor rating scales. *Brain.* 2006; 129:2660–2666. [PubMed: 16891320]
18. Giuffrida JP, Riley DE, Maddux BN, Heldman DA. Clinically deployable Kinesia technology for automated tremor assessment. *Mov Disord.* 2009; 24:723–730. [PubMed: 19133661]
19. Temperli P, Ghika J, Villemure JG, Burkhard PR, Bogousslavsky J, Vingerhoets FJ. How do parkinsonian signs return after discontinuation of subthalamic DBS? *Neurology.* 2003; 60:78–81. [PubMed: 12525722]
20. Yang CC, Hsu YL. A review of accelerometry-based wearable motion detectors for physical activity monitoring. *Sensors (Basel).* 2010; 10:7772–7788. [PubMed: 22163626]

Highlights

- !! Clinician inter-rater reliability varied greatly by UPDRS gait and balance item.
- !! UPDRS scores were highly correlated with specific sensor-based kinematic features.
- !! Sensor-based kinematic features significantly improved in response to DBS.

Figure 1. KinetiSense™ (Great Lakes NeuroTechnologies Inc., Cleveland, OH) consists of a command module and five motion sensor units, each containing three orthogonal accelerometers and three orthogonal gyroscopes. One sensor unit was positioned on each heel, one on each thigh, and one on the sternum.

Figure 2.

Gait Analysis. Subjects were instructed to walk forward, turn 180 degrees, and walk back. Data from a single subject collected from the heel and sternum sensor during the gait task were used to calculate various features of walking, turning, and freezing of gait (FOG).

Figure 3.

Scatter plots for items 1–6 quantitative features versus average clinician score. Item 7 posture was poorly correlated to clinician scores ($r=0.1$) and was not statistically significant ($p<0.4$). The dotted line represents the linear regression fit.

Table 1

Correlation coefficients (r) between mean clinician UPDRS score and item-specific kinematic features detailed in section 2.3. The same kinematic features were calculated for item 4 (gait) and item 5 (freezing of gait). RMS: root mean square. CV: coefficient of variation

Task	Feature	r
1	Log RMS angular velocity	-0.77
	Log RMS angle	-0.78
	CV of time between taps	0.71
2	Heel Log linear velocity	-0.76
	Heel log amplitude	-0.68
	Heel CV of time between leg lifts	0.64
	Thigh log RMS angular velocity	-0.84
	Thigh log RMS angle	-0.74
	Thigh CV of time between leg lifts	0.55
3	Peak angular velocity	-0.50
	Zero crossings	-0.79
4/5	Mean leg swing angular velocity	-0.84/-0.68
	CV of leg swing angular velocity	0.56/0.62
	Mean range of rotation	-0.89/-0.78
	CV of range of rotation	0.54/0.64
	Mean time between leg swings	0.18/0.05
	CV of time between leg swings	0.65/0.69
	Peak sternum angular velocity during turn	-0.67/-0.64
	Turn time	0.68/0.78
	Time delay after turn/before next step	0.45/0.49
	Steps during turn	0.47/0.41
6	Reaction time	0.34
	Steps required to recover balance	0.83
7	Sternum forward inclination angle	0.1

Table 2

A paired t-test was used to determine whether kinematic features detailed in section 2.3 significantly changed in response to stimulation. N.S. indicates p-values greater than 0.05. RMS: root mean square. CV: coefficient of variation.

Task	Feature	p
1	Log RMS angular velocity	0.005
	Log RMS angle	0.005
	CV of time between taps	0.002
2	Heel Log linear velocity	0.001
	Heel log amplitude	0.001
	Heel CV of time between leg lifts	0.005
	Thigh log RMS angular velocity	0.001
	Thigh log RMS angle	0.001
	Thigh CV of time between leg lifts	0.005
3	Peak angular velocity	N.S.
	Zero crossings	N.S.
4/5	Mean leg swing angular velocity	0.02
	CV of leg swing angular velocity	0.04
	Mean range of rotation	0.002
	CV of range of rotation	N.S.
	Mean time between leg swings	N.S.
	CV of time between leg swings	N.S.
	Peak sternum angular velocity during turn	0.05
	Turn time	N.S.
	Time delay after turn/before next step	N.S.
6	Steps during turn	N.S.
	Reaction time	N.S.
7	Steps required to recover balance	N.S.
	Sternum forward inclination angle	N.S.