Gait Assessment in Parkinson's Disease: Toward an Ambulatory System for Long-Term Monitoring

Arash Salarian*, *Member, IEEE*, Heike Russmann, François J. G. Vingerhoets, Catherine Dehollain, *Member, IEEE*, Yves Blanc, Pierre R. Burkhard, and Kamiar Aminian, *Member, IEEE*

***Abstract*—An ambulatory gait analysis method using body-attached gyroscopes to estimate spatio-temporal parameters of gait has been proposed and validated against a reference system for normal and pathologic gait. Later, ten Parkinson's disease (PD) patients with subthalamic nucleus deep brain stimulation (STN-DBS) implantation participated in gait measurements using our device. They walked one to three times on a 20-m walkway. Patients did the test twice: once STN-DBS was ON and once 180 min after turning it OFF. A group of ten age-matched normal subjects were also measured as controls. For each gait cycle, spatio-temporal parameters such as stride length (SL), stride velocity (SV), stance (ST), double support (DS), and gait cycle time (GC) were calculated. We found that PD patients had significantly different gait parameters comparing to controls. They had 52% less SV, 60% less SL, and 40% longer GC. Also they had significantly longer ST and DS (11% and 59% more, respectively) than controls. STN-DBS significantly improved gait parameters. During the stim ON period, PD patients had 31% faster SV, 26% longer SL, 6% shorter ST, and 26% shorter DS. GC, however, was not significantly different. Some of the gait parameters had high correlation with Unified Parkinson's Disease Rating Scale (UPDRS) subscores including SL with a significant correlation ($r = -0.90$) with UPDRS gait subscore. We concluded that our method provides a simple yet effective way of ambulatory gait analysis in PD patients with results confirming those obtained from much more complex and expensive methods used in gait labs.**

***Index Terms*—Biomedical signal processing, gait analysis, gyroscope, Parkinson's disease (PD), subthalamic nucleus deep brain stimulation (STN-DBS), wearable technology.**

Manuscript received July 2, 2003; revised November 1, 2003. This work was supported in part by the Swiss National Science Foundation under Grant FNRS 3152-062006.00/1. *Asterisk indicates corresponding author.*

*A. Salarian is with the Laboratory of Movement Analysis and Measurement (LMAM), Swiss Federal Institute of Technology (EPFL), 1015 Lausanne, Switzerland (e-mail: arash.salarian@epfl.ch).

H. Russmann is with the Department of Neurology, Centre Hospitalier Universitaire Vaudois, 1005 Lausanne, Switzerland.

F. J. G. Vingerhotes is with the Neurodegenerative Disorders Unit, Centre Hospitalier Universitaire Vaudois, 1005 Lausanne, Switzerland (e-mail: francois.vingerhoets@chuv.hospvd.ch).

C. Dehollain is with the Institut de Microélectronique et Microsystèmes, Swiss Federal Institute of Technology (EPFL), 1015 Lausanne, Switzerland.

Y. Blanc is with the Laboratoire de Cinésiologie, University Hospital of Geneva, 1211 Geneva, Switzerland.

P. R. Burkhard is with the Movement Disorders Clinic, Neurology Department, University Hospital of Geneva, 1211 Geneva, Switzerland (e-mail: pierre.burkhard@hcuge.ch).

K. Aminian is with the Laboratory of Movement Analysis and Measurement (LMAM), Swiss Federal Institute of Technology (EPFL), 1015 Lausanne, Switzerland (e-mail: kamiar.aminian@epfl.ch).

Digital Object Identifier 10.1109/TBME.2004.827933

I. Introduction

PARKINSON'S disease (PD) [1] is one of the most common degenerative diseases in the general population. Cardinal symptoms, including tremor at rest, rigidity, akinesia, and postural instability, are essentially motor in nature and result from a selective and progressive loss of dopaminergic neurons of the *substantia nigra pars compacta*. While a large armamentarium of symptomatic dopamine-replacement therapies is currently available, including levodopa, dopamine agonists, and COMT inhibitors among others, all of them have been associated with long-term motor and nonmotor complications, in particular end-of-dose deteriorations known as the wearing-off phenomenon and abnormal involuntary movements known as dyskinesia, which may add significantly to the functional impairment induced by the disease itself [2], [3]. Advanced PD is thus typically characterized by severe, unpredictable, and abrupt changes of the patient's motor function whereby OFF periods, characterized by the temporary loss of drugs' efficacy and the return of most parkinsonian symptoms, alternate, sometimes within minutes, with ON periods, during which the medication effects generate dyskinesia. The clinical assessment of these ON–OFF fluctuations, whose accuracy will determine the therapeutic interventions necessary to overcome them, is difficult and relies mostly on subjective historical data obtained from the patient or from relatives and on clinical scales such as the Unified Parkinson's Disease Rating Scale (UPDRS) [4] which are completed at the time of the patient's visit. To delineate precisely the temporal evolution of these complications, their characteristics, and their severity, more objective instrumental methods are needed [5].

Gait is a particular semi-automatic motor task which is specifically sensitive to ON–OFF changes of Parkinsonian state. When OFF, PD patients tend to walk slowly with short shuffling steps, reduced arm swing, stooped posture and they may present start hesitations and freezing episodes when turning around or facing an obstacle. During an ON state, the same patients may walk nearly normally with or without "dancing" steps as a result of the presence of dyskinesia involving the lower limbs. Analysis of gait parameters may therefore constitute a reliable paradigm to assess global motor function over time in PD patients. However, until now there has been a limited number of ambulatory systems to analyze human gait. Some of these systems need special footwear with foot switches or other pressure sensitive devices inside [6], [7]. Using special footwear is not always possible and may also hinder subject's normal gait. Moreover, PD patients often tend to shuffle while walking, making the initial and terminal contact detection

difficult. In these cases, the gait temporal parameters cannot be calculated precisely. In addition, the foot-switch techniques do not provide spatial parameters. An accelerometer which does not need to be fixed under the foot has been used as an alternative [8], [9]. An automated algorithm for gait temporal parameters estimation was proposed by [10] and validated on osteoarthritis patients. More recently, using accelerometers attached to the trunk an original method to estimate mean step length and walking speed was described by [11]. However, this system was not validated in pathological gait.

The aim of this study was to develop a new method for ambulatory gait analysis in PD patients based on body fixed sensors. For this purpose, a portable analysis system, Physilog[1] [12], developed in our institute, was used. The system is based on gyroscopes with long-term recording capability to study spatio-temporal gait parameters in a particular group of PD patients treated with bilateral subthalamic nucleus (STN) deep brain stimulation (DBS) [13], [14] in whom the switching ON or OFF the stimulators mimicked severe motor fluctuations naturally occurring in drug-treated PD patients.

II. Methods

A. Patients and Experimental Setup

A group of ten PD patients, 20±3 months after implantation of bilateral STN-DBS was selected. The group included five males and five females with average age of 61.5 years ($\max = 75.1$, $\min = 48.7$, $\mathrm{STD} = 7.8$) and average body mass and height of 71.1±14.7 kg and 168.7±11.1 cm. A group of neurologically intact subjects was also selected as normal controls. They were five males and five females with an average age of 63.6 years ($\max = 82.8$, $\min = 45.2$, $\mathrm{STD} = 10.5$) and average body mass and height of 70±10.9 kg and 170.9±8.4 cm. The study has been approved by the Ethics Committees of both involved hospitals and written informed consent had been given to all patients prior to enrollment.

Before each measurement, the PD patients were evaluated using the UPDRS (motor Section III, [4]). The motor section of the UPDRS is composed of 14 subscores in a zero to four scale, where four represents the worst disability and zero means no disability (see Table III). This way, the range of the possible values will be from zero to 108. The subject is asked to perform specific tasks and a specialist gives a subjective score for each activity.

Each PD patient participated twice in the study: once during Stim ON (i.e., when both stimulators have been turned on) and once during Stim OFF (i.e., when both stimulators have been turned off). The "Stimulation OFF" measurement was recorded 180 min after the STN-DBS was turned off [15].

Subjects stood up from a sitting position on a chair, walked 20 m on a straight line toward a second chair, and sat on it. Each subject performed three trials at a comfortable speed. In the case that PD patients, especially during Stim OFF period, had difficulties walking the full distance, they were asked to walk as far as they could.

[1]Registered trademark BioAGM, CH.

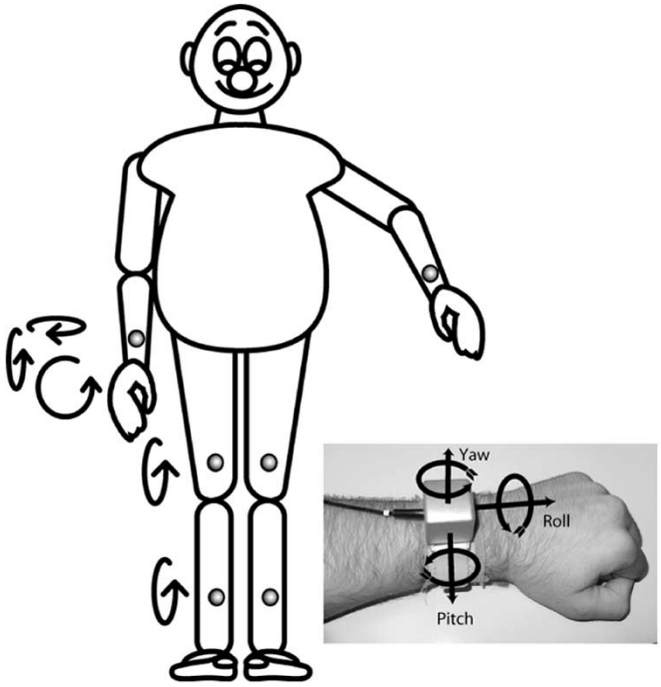

Fig. 1. Attachment of the sensors on the body. Circles represent sensors: One sensor on each left and right forearms, thighs, and shanks. Close-up view of sensor on forearm shows its three sensitive axes (pitch, roll, and yaw).

B. Measurement System

During the walking trials, subjects carried a Physilog portable data logger. Gyroscopes that measure the angular rate of the rotations were attached to selected body segments (see Fig. 1).

Four miniature uni-axial piezoelectric gyroscopes (Murata, ENC-03J) were attached on the lower limbs: one on each thigh and shank. Rubber bands were used to fix the sensors which were aligned to the medio-lateral axis, hence measuring rotations in sagittal plane. The sensors on the shank had a range of ±600°/s while the sensors on the thighs had a range of ±400°/s. To record movements of the upper extremities during walking, two three-dimensional (3-D) gyroscopes were attached to the forearms. These sensors had a range of ±1200°/s for each of the three axes of rotation (pitch, roll, and yaw). In summary, ten gyroscopes in six sensor sites were used. Data was recorded with a sampling rate of 200 Hz and 12 bits/sample and stored in an 8-MB memory card.

All measurement sessions were recorded using a portable video camera. After each session, a reviewer carefully examined the video tape and counted the number of gait cycles in each gait cycle to calculate the sensitivity of the algorithm in gait detection.

C. Temporal Parameters Estimation

To cancel the effect of drift of the gyroscopes, all signals were filtered using a high-pass IIR filter before any other processing (see Appendix A). Using the signals from shanks, gait cycles and related events were detected and temporal parameters of gait were estimated. The first step was to detect initial and terminal contact of feet with the ground (IC and TC). By simultaneously recording the gait signal using Physilog and force-plate (Kistler, CH), the intervals where these events occurred were determined

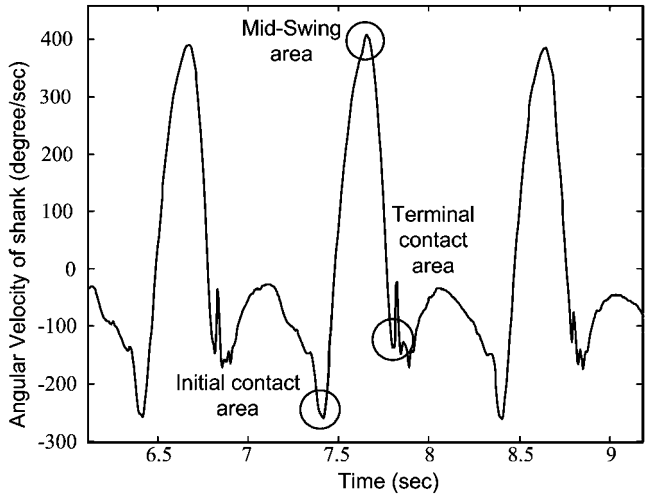

Fig. 2. Shank angular velocity. Marked areas show where important gait events occur.

based on the shank angular velocity (Fig. 2). The swing phase of a gait cycle is characterized by a positive shank angular velocity reaching its highest values at around midswing. Prior to swing phase, a negative angular velocity peak can be observed which is associated with TC. At the end of the swing period, the IC area is characterized by a several negative angular velocity peaks. The first negative peak in this area is associated with IC. With different populations of normal and pathologic, young, and elderly subjects, previous studies confirm the presence of these peaks [12], [16].

Taking advantage of these facts, a new algorithm was developed to extract the precise instances of IC and TC from right and left shank angular velocity of the respective foot. For the clarity of the explanation, let us consider the right shank angular velocity as input. It is obvious that the similar method applies to the signal from the left side. The starting point was the identification of the time events corresponding to the midswing ($t_{\rm ms}$) of shank angular velocity. The $t_{\rm ms}$ samples represent approximately the moment of midswing during a gait cycle, however their exact significance is not of interest. They were only used as references in order to select the intervals in which negative peaks reminiscent of TC and IC were to be found. First, the local maximum peaks of the signal were detected. Those peaks that were larger than 50°/s were candidates for marking the midswing area. If multiple adjacent peaks within a maximum distance of 500 ms were detected, the peak with the highest amplitude was selected and the others were discarded. This prominent peak in the swing area was taken as the midswing.

In the next step, local minimum peaks of shank signal inside interval $[t_{\rm ms} - 1.5\ {\rm s}, t_{\rm ms} + 1.5\ {\rm s}]$ were searched. The nearest local minimum after the $t_{\rm ms}$ was selected as IC. As the negative peak associated to TC was generally a small peak, to smooth the signal and to get rid of spurious peaks, the signal was filtered using a low-pass FIR filter with cutoff frequency of $f_c \approx 30$ Hz and pass-band attenuation of less than 0.5 dB. The local minima in the signal were searched and for each detected midswing the minimum prior to $t_{\rm ms}$ with amplitude less than -20°/s was selected as the terminal contact. The -20°/s threshold was used to avoid detecting a wrong peak in the swing area instead of the TC.

To validate these algorithms, a data base of prerecorded gait cycles (see Appendix C for the details) recorded both with Physilog and force-plate and a camera-based motion-capture system (Elite) was used [16].

After detection of ICs and TCs, gait cycles were formed to calculate gait temporal parameters. Each complete gait cycle had five associated time events. In order of occurrence they are: initial contact of right foot $({\rm IC}_R)$, terminal contact of left foot $({\rm TC}_L)$, initial contact of left foot $({\rm IC}_L)$, and terminal contact of right foot $({\rm TC}_R)$. The fifth time event was the next initial contact of right foot that was also the start of the next gait cycle. This way, the conditions that time events within kth gait cycle was valid were

$$ {\rm IC}_R(k) < {\rm TC}_L(k) < {\rm IC}_L(k) < {\rm TC}_R(k). \tag{1} $$

To form gait cycles conforming to (1), for each ${\rm IC}_R$ a simple algorithm was used to find the correct corresponding gait events. In the case a valid gait event could not be detected for a gait cycle, a special *unknown* value was assigned to it, practically stopping further related calculations on that particular gait cycle. This could happen if:

- the subject suddenly stopped walking, and then started to walk again *with the same* foot he took the last step with;
- the walking was changed to running, i.e., no double-stance period could be detected;
- in rare cases one or more respective IC and TC events could not be detected

Based on these time events, temporal parameters could be calculated as follows (formulas written for the right leg; same formulas were used for the left leg with the change in the corresponding symbols):

Gait Cycle Time:

$$ {\rm GCT}(k) = {\rm IC}_R(k+1) - {\rm IC}_R(k) \tag{2} $$

Stance:

$$ {\rm ST}(k) = \frac{{\rm TC}_R(k) - {\rm IC}_R(k)}{{\rm GCT}(k)} \times 100 \tag{3} $$

Initial Double Support:

$$ {\rm IDS}(k) = \frac{{\rm TC}_L(k) - {\rm IC}_R(k)}{{\rm GCT}(k)} \times 100 \tag{4} $$

Terminal Double Support:

$$ {\rm TDS}(k) = \frac{{\rm TC}_R(k) - {\rm IC}_L(k)}{{\rm GCT}(k)} \times 100 \tag{5} $$

Double Support:

$$ {\rm DS}(k) = {\rm IDS}(k) + {\rm TDS}(k) \tag{6} $$

Limp:

$$ {\rm Limp}(k) = \left| {\rm IDS}(k) - {\rm TDS}(k) \right|. \tag{7} $$

Fig. 3 summarizes different steps taken in the calculation of temporal parameters.

D. Estimating Spatial Parameters of Gait

To find the instantaneous angle of each segment, the angular velocity of that segment was integrated during each gait cycle (see Fig. 3). This way, having discrete values of $\omega[n]$ for each sample and sampling rate of Δ, instantaneous angle $\theta[n]$ of the segment will be

$$\theta[n] = \theta[n-1] + \frac{\Delta}{2}(\omega[n] + \omega[n-1]). \tag{8}$$

In the above equation, the initial condition θ_0 is unknown and was assumed zero for the start of each gait cycle. The range of rotation of the segments is

$$\tilde{\theta}_n = \underset{i=\mathrm{IC}_n}{\overset{\mathrm{IC}_{n+1}}{\mathrm{MAX}}}(\theta[i]) - \underset{i=\mathrm{IC}_n}{\overset{\mathrm{IC}_{n+1}}{\mathrm{MIN}}}(\theta[i]) \tag{9}$$

where IC_n stands for initial contact of nth gait cycle. This value is independent of θ_0. Based on the instantaneous segment angles, the calculation of joint angles was trivial. If we have the angle of segments $s1$ and $s2$ making the joint j, then the joint angle for each sample will be

$$\theta_j[n] = \theta_{s1}[n] - \theta_{s2}[n] + \theta_{j0} \tag{10}$$

where again θj_0 or the initial condition is unknown, and we took the value as zero for the start of each gait cycle. The range of joint rotations was also calculated in the same way as (9). Using this method, the range of rotations of each shank and thigh were calculated. Similarly, for each axis of the sensors on forearms the range of the rotations during each gait cycle were calculated.

To validate the results, a database of prerecorded gait cycles was used and estimated ranges of angles were compared to those obtained from the reference system (see Appendix C).

Stride length and velocity were then calculated using the range of thigh and shank rotations with our previously developed gait model (see Appendix B).

E. Outcomes

The following parameters were finally estimated and reported: spatio-temporal gait parameters including gait cycle time, double support, limp and stance, stride length, and velocity; as well as range of rotations of shanks, knees, forearms, and peak angular velocity of shanks.

Double support, limp, and stance were normalized by gait cycle duration and presented as percentage of it (0%–100%). The stride length and velocity were normalized to subject's height and presented as a percentage of stature.

Mean and standard deviation of parameters were calculated. To compare variability of the stride-to-stride parameters, the coefficient of variation (CV) (standard deviation/mean) was calculated. To compare the mean values of different parameters between Stim ON and Stim OFF groups, Wilcoxon's nonparametric paired test, the sign-rank test [17], was used and to compare between control group and the PD patients, rank-sum test was used. When needed, the Jarque-Bera [18] test for goodness-of-fit to a normal distribution was used. To estimate the significance of the correlation coefficients, the Pearson test was used.

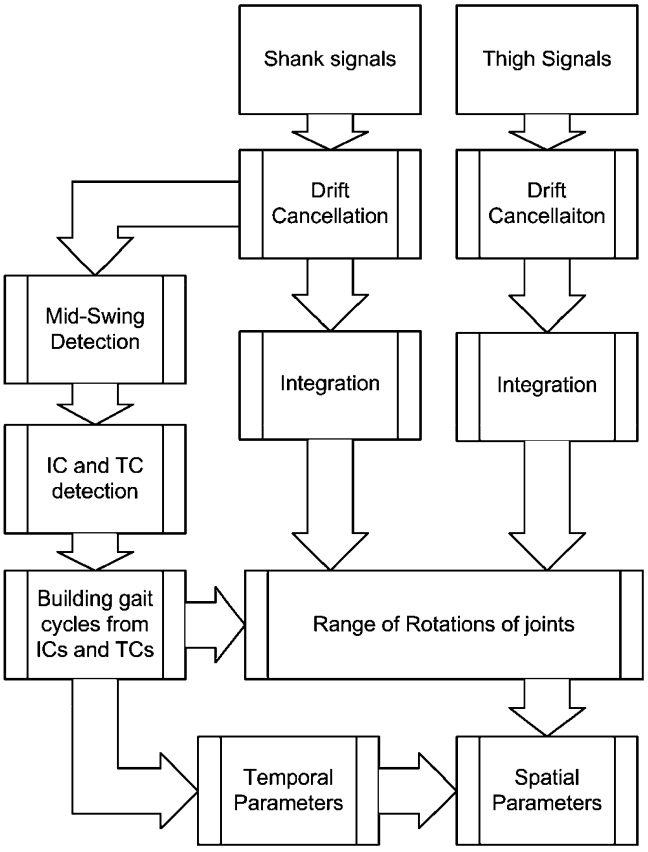

Fig. 3. Flowchart of spatio-temporal gait parameter estimation algorithm.

TABLE I
ERROR IN ESTIMATION OF GAIT PARAMETERS BASED ON THE NEW METHOD IN COMPARISON TO THE REFERENCE MOTION-CAPTURE SYSTEM. AS FOR COMPARISON, THE ERROR OF THE PREVIOUS METHOD [12] IS ALSO REPORTED

Error in estimation of parameter	New Method		Previous Method	
	Mean	STD	Mean	STD
Initial Contact (ms)	-8.7	12.5	-12.6	14.3
Terminal Contact (ms)	-2.9	26.8	6.6	29.2
Gait Cycle time (ms)	2.2	23.2	2.4	24.1
Stance (ms)	5.9	29.6	19.2	31.4
Range of Shank rotation (deg)	0.7	3.3	-0.3	3.3
Range of Thigh rotation (deg)	3.5	4.2	2.4	4.2
Stride Length (cm)	3.5	8.5	0.4	9.6
Stride Velocity (cm/s)	3.0	7.6	2.5	8.3
Total execution time (s)	10.5		94.6	

III. RESULTS

A. Error in Estimation of Gait Parameters Compared to Reference Motion-Capture Systems

Table I summarizes the error in estimating gait parameters using our new method against the reference motion-capture systems (see Appendix C). An error was defined as the difference between values for the reference system and the values estimated by our algorithms. Mean and STD of this error, across all gait cycles and all subjects, was then calculated. Mean of the error in independent parameters (like IC and TC) signifies the presence of a systematic error which can be later corrected and the STD of the error signifies the range of the accuracy of the system in comparison to the references. Further, the Jarque-Bera

TABLE II
SUMMARY OF MEASURED TRIALS AND GAIT CYCLES

	PD subjects Stim ON	PD Subjects Stim OFF	Controls
Trials	21	17	30
Total Gait Cycles	274	248	514
True-Positive Gait Cycles	274	247	514
False-Positive Gait Cycles	3	4	0
True-Positive Gait Events	272	239	512
False-Positive Gait Events	1	4	2

goodness-of-fit test confirms that the error has a normal distribution ($p < 0.0001$).

B. Sensitivity of Gait Cycle and Gait Event Detection

All controls performed the three walking trials. PD patients, however, were not consistently able to perform all trials and did one to three trials each. For each trial, the first and last two gait cycles were omitted to avoid the effects of gait initiation and termination.

Table II summarizes the measured trials and number of gait cycles eventually obtained for each group. *Total Gait Cycles* comes from the observation based on the video tapes: a reviewer counted the number of gait cycles in each trial. There are two possible types of detection errors: errors in detection of a gait cycle (i.e., error in finding midswing peak) and errors in finding related gait events (IC and TC). Based on these values, we had a very high sensitivity in detection of gait cycles (100% for controls and 100% for PD patients during Stim ON and 99.6% during Stim OFF). Also, sensitivity in the detection of gait events was very high (99.6% for controls and 99.3% for PD patients during Stim ON and 96.4% during Stim OFF). The positive prediction value (PPV) in the detection of gait cycles has been 100% for controls, 98.9% for Stim ON, and 98.4 for the Stim OFF group. PPV for detection of IC and TC has been 99.6% for controls and 99.6% for Stim ON and 98.4% for the Stim OFF group.

Fig. 4 shows a sample of the recorded signal on the shank. As it can be seen here, during the Stim ON, the subject moved his shank significantly faster; however, both signals show noticeable variability in the peak speed of the shank.

C. UPDRS Motor Score

The UPDRS motor scores and subscores for all PD patients during Stim ON and Stim OFF periods are presented in Table III. Stimulation significantly improved (decreased) UPDRS motor scores ($p = 0.002$). All UPDRS subscores were also significantly improved ($p = 0.042$) but not the subscores for Speech and Posture (items 18 and 28)

D. Gait Parameters

Gait parameters are reported in Table IV for both states of stimulation and also for controls in (mean $\pm$ S.D.) format. The results of the statistical hypothesis tests of equivalence of means and also CV of the three groups are also presented. A paired test was used when comparing the Stim ON and Stim OFF groups.

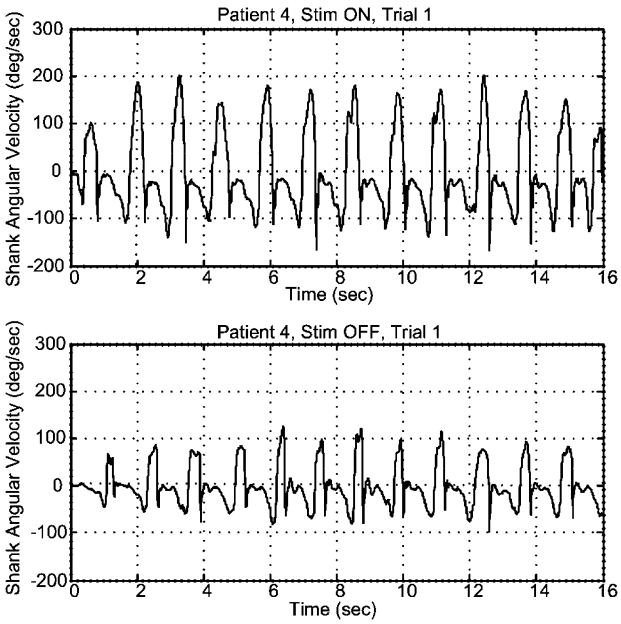

Fig. 4. Sample of the recorded signal on shanks. PD patient with and without stimulation shows a different pattern of walking.

IV. DISCUSSION AND CONCLUSION

A. Spatio-Temporal Parameters

During Stim OFF, PD patients had significantly (for the p-values and also standard deviations see Table IV) less stride velocity than controls (40.5%h/s versus 77.4%h/s) due to significantly shorter stride lengths (46.2%h versus 77.1%h) and significantly longer gait cycle times (1.4 versus 1.0s). Also, duration of stance (65.7% versus 59.4%) and double support (31.4% versus 18.7%) was significantly longer for PD patients as compared to controls. Limp, defined as the difference between initial and terminal double support, in PD patients during Stim OFF was significantly larger than controls (7.2% versus 1.4%). These results are consistent with the clinical observation of a PD gait which is typically characterized by shortened, shuffling steps, reduced speed, and difficulty to initiate lower limb movements. These abnormalities are mostly due to the basal ganglia dopamine deficiency-related symptom akinesia, and to a lesser extent rigidity, which are the main determinants of gait impairment in moderate to advanced PD. This is supported by a striking, although incomplete, improvement of these symptoms under levodopa replacement therapy observed in a majority of PD patients and possibly following STN-DBS; although, the latter has been recently the matter of some debate.

Indeed, in this study during Stim ON, STN-DBS significantly improved stride velocity (53.1%h/s versus 40.5%h/s), stride length (58.5%h versus 46.2%h), stance (61.5% versus 65.7%), and double support (23.0% versus 31.4%) but not the gait cycle time, which is in agreement with previously reported findings [19]–[21]. Limp, however, was not changed significantly. Despite the improvement obtained during Stim ON, all parameters remained significantly different from controls except stance and double support; i.e., gait cycle time (1.2 versus 1.0 s), stride

TABLE III
UPDRS PART III (MOTOR FUNCTIONS) SCORES AND SUBSCORES DURING STIM ON AND STIM OFF STATES

Patient	1		2		3		4		5		6		7		8		9		10	
Stimulation	ON	OFF	ON	OFF	ON	OFF	ON	OFF	ON	OFF	ON	OFF	ON	OFF	ON	OFF	ON	OFF	ON	OFF
UPDRS Part III sub-scores																				
18. Speech	2	2	2	2	1	2	1	1	0	1	1	4	2	2	1	1	0	0	2	2
19. Facial Expression	1	3	1	4	2	4	1	2	1	2	0	2	2	2	1	2	1	2	2	2
20. Tremor at Rest																				
20.a Face	0	0	1	2	1	2	0	0	1	2	1	1	1	1	0	1	0	2	1	1
20.b Upper Extremity R, L	0, 1	4, 4	1, 0	3, 1	0, 0	1, 1	0, 0	3, 1	0, 0	1, 1	1, 0	4, 4	1, 1	1, 1	0, 0	2, 1	0, 0	0, 0	0, 0	1, 1
20.c Lower Extremity R, L	0, 0	0, 0	1, 0	3, 0	0, 0	0, 0	0, 0	0, 0	0, 0	0, 0	0, 0	2, 2	0, 0	1, 1	2, 0	2, 1	0, 0	0, 0	0, 0	0, 1
21. Action or Postural Tremor (R, L)	1, 1	4, 4	0, 1	2, 1	1, 1	1, 1	0, 0	3, 1	0, 0	0, 1	1, 1	2, 2	1, 1	1, 2	1, 1	1, 2	0, 1	0, 1	1, 0	1, 2
22. Rigidity																				
22.a Neck	1	4	1	4	2	4	1	4	1	1	0	2	2	3	2	2	1	2	2	3
22.b Upper Extremity R, L	1, 1	4, 4	1, 0	4, 2	2, 2	3, 4	0, 0	4, 4	0, 0	1, 0	0, 0	1, 2	2, 2	1, 2	0, 0	2, 1	0, 1	1, 2	2, 2	3, 3
22.c Lower Extremity R, L	0, 1	2, 3	0, 1	3, 3	2, 2	3, 4	2, 2	3, 4	1, 0	0, 0	0, 0	1, 1	2, 1	2, 2	1, 1	3, 3	0, 0	2, 2	0, 0	2, 3
23. Finger taps	1, 2	3, 4	4, 2	4, 4	2, 3	3, 4	1, 1	4, 4	0, 0	2, 2	0, 0	2, 3	2, 2	3, 4	1, 1	2, 2	1, 1	3, 3	2, 1	2, 2
24. Hand Movements R, L	1, 3	4, 4	4, 2	4, 4	2, 3	3, 4	1, 1	3, 4	0, 0	1, 2	0, 0	2, 3	2, 2	4, 4	2, 2	3, 3	1, 2	2, 3	2, 1	2, 2
25. Rapid Alternate Movements R, L	3, 3	4, 4	4, 3	4, 4	3, 3	4, 4	2, 2	4, 4	0, 0	1, 2	0, 0	1, 2	1, 1	4, 4	2, 2	3, 3	1, 1	2, 3	3, 1	3, 3
26. Leg Agility R, L	0, 0	4, 4	0, 1	3, 4	1, 2	3, 4	1, 2	3, 3	0, 0	2, 1	0, 0	1, 1	0, 0	0, 1	1, 1	3, 3	0, 0	2, 2	2, 1	3, 4
27. Arising From Chair	0	4	0	3	3	4	0	1	0	0	0	0	0	0	0	2	0	0	0	2
28. Posture	1	1	0	0	2	2	1	1	0	0	0	0	1	2	1	2	0	0	1	2
29. Gait	0	3	2	3	3	4	2	2	0	1	0	0	1	1	2	3	0	0	1	2
30. Postural Stability	1	4	1	3	3	4	1	1	0	0	0	1	1	2	2	3	0	1	2	2
31. Body Bradykinesia	1	4	2	4	3	4	1	3	0	1	1	2	0	1	1	2	0	1	1	2
Total	26	85	35	78	49	77	23	67	4	25	6	48	31	52	28	58	11	36	30	56

TABLE IV
MEASURED GAIT PARAMETERS FOR PATIENTS DURING STIM ON AND OFF STATE AND CONTROLS. p-VALUES MORE THAN 0.05 WERE CONSIDERED AS NOT SIGNIFICANT (N.S.)

	Values for each group in mean±S.D.			p-value for the equivalence of mean of parameters			p-value for equivalence of mean of C.V,		
Gait Parameters	Stim OFF	Stim ON	Control	ON/OFF paired	ON v.s. Control	OFF v.s. Control	ON/OFF paired	ON v.s. Control	OFF v.s. Control
Gait Cycle Time (s)	1.4± 0.6	1.2±0.2	1.0±0.1	N.S.	0.0312	0.0312	N.S.	0.0211	N.S.
Stance (%)	65.7± 8.6	61.5±4.5	59.4±1.2	0.0488	N.S.	0.0312	N.S.	0.0173	0.0211
Double Support (%)	31.4±17.1	23±9.1	18.7±2.5	0.0488	N.S.	0.0312	N.S.	N.S.	N.S.
Limp (%)	7.2±8.6	4.2±2.2	1.4±0.5	N.S.	0.0010	0.0006	N.S.	N.S.	N.S.
Stride Length (%h)	46.2±19.4	58.6±17.9	77.1±6.5	0.0020	0.0073	0.0004	N.S.	N.S.	0.0017
Stride Velocity (%h/s)	40.5±23.5	53.1±20.2	77.4±9.2	0.0039	0.0022	0.0003	0.0137	N.S.	N.S.
Range of Shank rotation (deg)	45.6±19.5	56.5±18.5	76±5.9	0.0020	0.0022	0.0002	0.0020	0.0113	0.0028
Range of Thigh rotation (deg)	28±8.5	34.4±8.6	34.4±11.8	0.0098	N.S.	N.S.	N.S.	0.0257	0.0028
Range of Knee rotation (deg)	39.4±13.7	45.4±15.8	60.4±7.9	0.0195	0.0113	0.0003	N.S.	N.S.	0.0140
Range of hand rotation, Pitch axis (deg)	8.4±5.1	17.8±12.6	20.2±6.4	N.S.	N.S.	0.0013	N.S.	N.S.	N.S.
Range of hand rotation, Roll axis (deg)	14±13.3	18.2±6.7	22.9±5	N.S.	N.S.	0.0028	N.S.	N.S.	N.S.
Range of hand rotation, Yaw axis (deg)	10.3±5.3	24.6±12.1	47.6±8.4	0.0039	0.0013	0.0002	N.S.	0.0452	N.S.
Peak Shank angular Velocity (deg/s)	225.2±103.5	275.4±110	386.3±40.1	0.0020	0.0058	0.0003	0.0020	N.S.	0.0452

length (58.6%h versus 77.1%h), stride velocity (53.1%h/s versus 77.4%h/s), and limp (4.2% versus 1.4%). These results confirm that STN-DBS provides a substantial and measurable improvement of most gait parameters. However, according to our data, this benefit appears selective to certain aspects of gait disturbances (stride length and velocity, stance, double support), since other parameters were found unchanged (gait cycle time and limp) between Stim ON and Stim OFF situations. These data suggest that these unmodified variables may reflect dopamine-independent features of PD gait, well-known examples of which are freezing episodes while ON, festination, and kinesia paradoxica.

B. Segment and Joint Rotations and Angular Velocities

On the lower limbs, during Stim OFF, measured parameters showed differences between controls and PD patients. Range of knee flexion (39.4 versus 60.4 degree) was significantly different due to different range of shank rotation (45.6 versus 76.0 degree) but not for the range of thigh rotation which had not a significant difference. The reduction of range of knee flexion confirms findings of [21]–[23] and is consistent with the common clinical observation of bent knees in PD patients, which is mostly due to limb rigidity. Also, peak velocity of shank was significantly lower for the PD patients (225 versus 386.3 degree/s). On the upper limbs, PD patients during Stim OFF had significantly less range of rotation of the forearms, namely in yaw axis (10.3 versus 47.6 degree), roll axis (14.0 versus 22.9 degree), and pitch axis (8.4 versus 20.2 degree). This represents the neurophysiological counterpart of the clinical feature of reduced arm swing which is an early sign of akinesia exhibited by PD patients while walking.

STN-DBS improved most of these parameters. During Stim ON, PD patients had a significantly larger range of knee flexion than controls (45.4 versus 39.4 degree) due to both larger thigh and shank rotation. This finding is in agreement with [21]. Peak velocity of the shank was also significantly higher during Stim ON versus Stim OFF (275 degree/s versus 225). On the upper limbs, however, STN-DBS only significantly improved range of forearm rotation in yaw axis (24.6 versus 10.3 degree). Despite these improvements, PD patients during Stim ON had significantly lower range of knee flexion than controls (45.4 versus 60.4 degree) due to lower thigh and shank rotation which also agree with [21]. Moreover, the peak shank angular velocity is also lower than controls (275.4 versus 386.3 degree/s). Also, the range of forearm rotations in yaw axis was significantly lower in PD patients during the ON state comparing to the controls (24.6 versus 47.6 degree). However, range of thigh rotation, maximum knee flexion, and rotations of forearm in roll and pitch axis was not significantly different from controls during Stim ON.

In summary, based on the data reported in Table IV, it can be inferred from the significant differences found for most, but not all, parameters in Stim ON patients compared to healthy controls that the magnitude of improvement provided by STN-DBS, however significant, is by no mean sufficient to "normalize" the gait in PD.

C. Correlation Between Gait Parameters and UPDRS

Six different subscores defined as u1 to u6 (see Table V) were made based on UPDRS III subscores (Table III). Estimated gait parameters were then compared to these subscores.

As seen from Table VI, range of rotation in the yaw axis of the forearms always has the highest significant correlation with bradykinesia and rigidity subscores amongst the three forearm sensors' axis. Stride-length, stride-velocity, and range of shank rotation show a very good correlation with gait subscore. Fig. 5 shows a scatter plot of u5 (gait and posture subscore) and estimated stride-length where significant and high correlation was found between a typical estimated outcome and UPDRS. The significance of this correlation was confirmed by using the bootstrapping method [24] where a range of $(-0.98, -0.72)$ was obtained for the 95% confidence interval of correlation coefficient.

TABLE V
UPDRS SUBSCORES USED IN CALCULATION OF CORRELATION COEFFICIENTS

Sub-score	Related symptoms	UPDRS III sub-scores
u1	Bradykinesia	23+24+25+26
u2	Rigidity	22
u3	Tremor	20+21
u4	Gait	29
u5	Gait and Posture	27+28+29+30
u6	Posture	28+30

TABLE VI
COEFFICIENT OF CORRELATION BETWEEN UPDRS SUBSCORES AND GAIT PARAMETERS. p VALUES MORE THAN 0.05 WERE CONSIDERED AS NONSIGNIFICANT (N.S.). ROR STANDS FOR *RANGE OF ROTATION*

Parameter	u1	u2	u3	u4	u5	u6
Gait Cycle Time	N.S.	N.S.	N.S.	0.61	0.65	0.65
Stance	0.48	0.45	N.S.	0.69	0.67	0.64
Double Support	0.48	0.45	N.S.	0.69	0.67	0.64
Limp	N.S.	N.S.	N.S.	0.51	0.49	0.50
Stride Length	-0.68	-0.74	N.S.	-0.90	-0.87	-0.80
Stride Velocity	-0.58	-0.65	N.S.	-0.84	-0.83	-0.79
Shank ROR	-0.64	-0.72	N.S.	-0.88	-0.84	-0.76
Thigh ROR	-0.59	-0.72	N.S.	-0.79	-0.81	-0.78
Range of Knee flexion	N.S.	-0.57	N.S.	-0.68	-0.68	-0.63
Forearm, Pitch axis ROR	-0.47	N.S.	-0.46	N.S.	N.S.	N.S.
Forearm, Roll axis ROR	-0.51	-0.56	N.S.	-0.60	-0.63	-0.55
Forearm, Yaw axis, ROR	-0.71	-0.69	-0.56	N.S.	-0.59	-0.52
Peak Shank angular Velocity	-0.56	-0.63	N.S.	-0.81	-0.78	-0.72

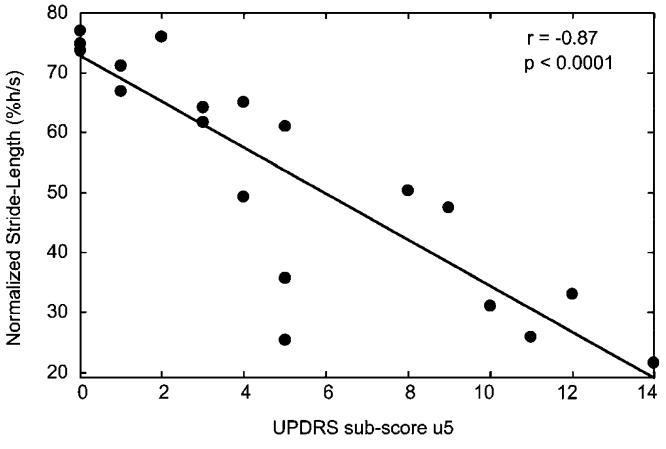

Fig. 5. Scatter plot comparing UPDRS subscore u5 and normalized stride length.

D. Validations

Two validation studies have been performed. In the first study, the sensitivity of the algorithm in detection of gait and temporal

parameters has been assessed; in the second one, the error of the system in comparison to reference systems has been calculated.

For the first study using a portable video camera, after each session a reviewer counted the actual gait cycles in each walking trial. The results were compared to the output of the algorithm. Despite the severe abnormalities during the Stim OFF period, the algorithm could detect gait cycles and related gait events (IC and TC) with very high sensitivity (more than 96%) and with PPV more than 98%.

In the second study, we have used the data recorded with Physilog and a reference system based on a motion-capture system (Elite) to find the accuracy of the algorithm in detection of gait temporal and spatial parameters. For this validation, however, we only used the gait data from normal subjects and pathological gait (see Appendix C). The estimated error (see Table I), considering the 200-Hz sampling rate of the system, was round two samples for IC and five samples for TC. The relative error in estimation of the gait cycle time was 2% and for stride length and stride velocity less than 8%. These results have been proved to be accurate enough to show significant differences between Stim ON and Stim OFF states of the PD patients.

E. Comparison to Other Ambulatory Systems

Our gyroscope-based method has several advantages over other ambulatory systems. Unlike some of the foot-switch or other pressure sensitive devices, no special footwear is needed for long-term monitoring and this way is more comfortable for long-term monitoring. Also, available foot-switch-based devices limit the gait analysis to the temporal parameters [6], [7], while ours can provide both temporal and spatial parameters. The same thing is true for many of the accelerometry methods, as they do not provide spatial parameters [10].

The idea of using gyroscopes to assess gait has been also used in different studies [25]; however, besides [12] there has been no other ambulatory method suitable for long-term monitoring with successful validations compared to reference systems. Our new method differs from [12] in the totally new signal processing approach. While we kept using the same model and sensors (only sensors on the forearms has been added), more traditional IIR and finite-impulse response filters was used instead of the wavelet approach. The results of the validations show that compared to the previous method, the new method has higher precision (STD in Table I) for spatio-temporal parameters estimation, higher accuracy (mean in Table I) for temporal event detection, while lower accuracy is observed for spatial parameters. Actually, the systematic error in detection of IC and TC has been reduced by a large margin (31.0% and 56.1% reductions, respectively). As a result, systematic error in detection of stance is now 69.3% smaller. Also, standard deviation of error in estimation of IC is now 12.5%, of TC is 8.2%, and of stance is 5.7% reduced. Moreover, the standard deviation of error in estimation of stride length and stride velocity is now 11.5% and 8.4% reduced, respectively. This improvement has been achieved while the new method performed the analysis nine times faster than the pervious method. Also, the new algorithm specifically addresses the cases that IC or TC cannot be correctly detected to prevent the propagation of an error to the next gait cycle.

F. Conclusion

An ambulatory method capable of quantifying a number of parameters related to the gait of PD patients has been introduced. The results can be summarized as follows.

- Using minimal attachment sites and without any need for per-person calibrations, the system could successfully estimate gait parameters with a high degree of accuracy.
- The method has been validated to assess gait changes in PD since the results obtained with this instrumental method reliably parallel clinical scores obtained with commonly used scales, such as the UPDRS as well as visual observation.
- The obtained results confirmed previous findings that were obtained using sophisticated gait analyses methods restricted to gait labs while preserving the possibility of an ambulatory monitoring of the subjects.
- Altogether, the results from Sections IV-A and B support the instrumental method described in this study as a sensitive tool to assess subtle changes of gait parameters over the time and to distinguish physiological from Parkinsonian gait, even in optimally STN-DBS-treated PD patients where the gait is frequently reported as normal.

The method presented here provides a simple and effective way of ambulatory gait analysis in PD patients. Moreover, we have already shown that symptoms related to PD such as tremor and bradykinesia can be quantified using forearms sensors [26]. We, therefore, believe that by increasing the memory capacity and the battery stamina, the proposed system can be used in long-term monitoring to assess gait in PD patients in their daily life along with other PD symptoms.

APPENDIX

A. Filtering and Drift Cancellation

To cancel possible offset and drift of the sensors, resulting from changes in temperature and also possible variations in supply voltage, a high-pass IIR filter was used. The transfer function of this filter was

$$H(z) = \frac{1 - z^{-1}}{1 - \alpha \cdot z^{-1}}. \tag{11}$$

To obtain precisely zero-phase distortion, the filter was applied to the input data twice. After filtering in the forward direction, the filtered sequence was reversed and run back through the filter. With $\alpha = 0.995$ the cutoff frequency of the filter was $f_c \approx 0.25$ Hz.

B. Gait Model

A double pendulum model for swing and an inverse double pendulum model for stance were used [Fig. 6(a)]. In this figure, L1 and L2 are length of the thigh and shank, respectively. Having the time of IC and TC events, swing and stance phases could be detected. The stride length was broken into three different segments, d_1 to d_3. The value of $d_1 + d_2$ was estimated during swing phase and d_3 was estimated during the stance phase. To estimate the stride length for the right foot, $d_1 + d_2$ was calculated by calculating α, range of right thigh rotation

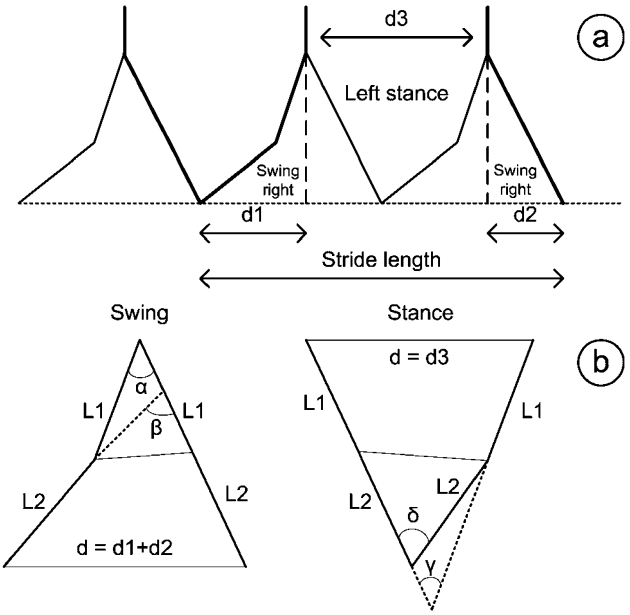

Fig. 6. Double pendulum model used to estimate gait parameters. (a) Gait cycle starts with right foot's initial contact. Right foot leaves the ground at TC and hits the ground at IC. (b) Stride length can be estimated by solving two separate geometries for swing and stance phases.

during swing phase, and β is range of right shank rotation during swing phase. For the same gait cycle, d_3 was estimated by calculating γ, range of left shank rotation during stance and δ, range of left thigh rotation during stance. Details of the calculation have been already reported [12]. To validate the results a prerecorded database of gait cycles was used and estimated stride length and velocity were compared to those estimated by reference system.

C. Prerecorded Database of Gait Cycles With a Reference System

For validation purposes, a prerecorded database of gait cycles has been used [16]. This database included 229 gait cycles of a group of eight normal subjects, seven coxarthorosis subjects, and six hip-arthoroplasty patients. Each subject had up to eight walking trials for each of the right and left legs and gait cycles were recorded by Physilog, a force plate, and a camera-based system (Elite). Recorded data was verified for potential technical problems like incorrect synchronization between the three devices or cases when subject did not walk over the force plate or both of his feet touched the force plate. Finally, 229 gait cycles recorded correctly by all three systems were selected.

ACKNOWLEDGMENT

The authors would like to thank P. Morel and J. Gramiger for the design of Physilog system and sensors.

REFERENCES

[1] J. Parkinson, *An Essay on the Shaking Palsy*. London, U.K.: Sherwood: Neely and Jones, 1817.

[2] T. A. Larsen, P. A. LeWitt, and D. B. Calne, "Theoritical and practical issues in assessment of deficits and therapy in parkinsonism," in *Lisuride and Other Dopamine Agonists*, D. B. Calne, R. J. McDonald, R. Horowski, and W. Wuttke, Eds. New York: Raven, 1983, pp. 363–373.

[3] R. L. Watts and A. S. Mandir, "Quantitative methods of evaluating parkinson's disease," in *The Scientific Basis for the Treatment of Parkinson's Disease*, C. W. Olanow and A. N. Lieberman, Eds. Park Ridge: Parthenon, 1992, pp. 13–32.

[4] S. Fahn and R. L. Elton, "Unified parkinson's disease rating scale," in *Recent Developments in Parkinson's Disease*, S. Fahn, C. D. Marsden, D. Calne, and M. Goldstein, Eds. Florham Park, NJ: McMillan Health Care, 1987, vol. II, pp. 153–163.

[5] S. K. Patrick, A. A. Denington, M. J. A. Gauthier, D. M. Gillard, and A. Prochazka, "Quantification of the UPDRS rigidity scale," *IEEE Trans. Neural Syst. Rehab. Eng.*, vol. 9, pp. 31–41, Mar. 2001.

[6] Z. Pataky, L. Faravel, J. Da Silva, and J.-P. Assal, "A new ambulatory foot pressure device for patients with sensory impairment. A system for continuous measurement of plantar pressure and a feed-back alarm," *J. Biomechan.*, vol. 33, pp. 1135–1138, 2000.

[7] I. Lackovic, V. Bilas, A. Santic, and V. Nikolic, "Measurement of gait parameters from free moving subjects," *Measurement*, vol. 27, pp. 121–131, 2000.

[8] M. Sekine, T. Tamura, M. Akay, T. Fujimoto, T. Togawa, and Y. Fukui, "Discrimination of walking patterns using wavelet-based fractal analysis," *IEEE Trans. Neural Syst. Rehab. Eng.*, vol. 10, pp. 188–196, Sept. 2002.

[9] A. T. M. Willemsen, F. Bloemhof, and H. B. K. Boom, "Automatic stance-swing phase detection from accelerometer data for peroneal nerve stimulation," *IEEE Trans. Biomed. Eng.*, vol. 37, pp. 1201–1208, Dec. 1990.

[10] K. Aminian, K. Rezakhanlou, E. De Andres, C. Fritsch, P. F. Leyvraz, and P. Robert, "Temporal feature estimation during walking using miniature accelerometers: An analysis of gait improvement after hip arthroplasty," *Med. Biol. Eng. Computing*, vol. 37, pp. 686–691, 1999.

[11] W. Zijlstra and A. L. Hof, "Assessment of spatio-temporal gait parameters from trunk accelerations during human walking," *Gait Posture*, vol. 18, pp. 1–10, Oct. 2003.

[12] K. Aminian, B. Najafi, C. Bula, P.-F. Leyvraz, and P. Robert, "Spatio-temporal parameters of gait measured by an ambulatory system using miniature gyroscopes," *J. Biomechan.*, vol. 35, pp. 689–699, 2002.

[13] F. J. G. Vingerhoets, J.-G. Villemure, P. Temperli, C. Pollo, E. Pralong, and J. Ghika, "Subthalamic DBS replaces levodopa in parkinson's disease: Two-year follow-up," *Neurology*, vol. 58, pp. 396–401, 2002.

[14] A. L. Benabid, A. Benazzouz, D. Hoffmann, P. Limousin, P. Krack, and P. Pollak, "Long-term electrical inhibition of deep brain targets in movement disorders," *Movement Disorders: Official J. Movement Disorder Soc.*, vol. 13, pp. 119–125, 1998.

[15] P. Temperli, J. Ghika, J. G. Villemure, P. R. Burkhard, J. Bogousslavsky, and F. J. G. Vingerhoets, "How do parkinsonian signs return after discontinuation of subthalamic DBS?," *Neurology*, vol. 60, pp. 78–81, 2003.

[16] K. Aminian, C. Trevisan, B. Najafi, H. Dejnabadi, C. Frigo, E. Pavan, A. Telonio, F. Cerati, E. C. Marinoni, P. Robert, and P.-F. Leyvraz, "Evaluation of an ambulatory system for gait analysis in hip osteoarthritis and total replaced patients," Gait Posture, to be published.

[17] F. Wicoxon, "Individual comparisons by ranking methods," *Biometrics Bull.*, vol. 1, pp. 80–83, 1945.

[18] G. G. Judge, R. C. Hill, W. E. Griffiths, H. Lütkepohl, and T.-C. Lee, *Introduction to the Theory and Practice of Econometrics*, 2nd ed. New York: Wiley, 1988.

[19] P. Krystkowiak, L. J. P. Defebure, J. L. Blatt, J. L. Bourriez, S. Blond, J. D. Guieu, and A. Destee, "Influence of subthalamic nucleus stimulation on gait in parkinson's disease: A study using the optoelectronic VICON system," *Neurology*, vol. 54, pp. A446–A446, 2000.

[20] M. E. Melnick, S. A. Radtka, N. Sierra, W. J. Marks, and P. A. Starr, "Effects of deep brain stimulation on gait parameters using 3- dimensional gait analysis," *Neurology*, vol. 54, pp. A457–A457, 2000.

[21] M. Ferrarin, L. Lopiano, M. Rizzone, M. Lanotte, B. Bergamasco, M. Recalcati, and A. Pedotti, "Quantitative analysis of gait in Parkinson's disease: A pilot study on the effects of bilateral sub-thalamic stimulation," *Gait Posture*, vol. 16, pp. 135–148, 2002.

[22] M. E. Morris, J. McGinley, F. Huxham, J. Collier, and R. Iansek, "Constraints on the kinetic, kinematic and spatiotemporal parameters of gait in parkinson's disease," *Human Movement Sci.*, vol. 18, pp. 461–483, 1999.

[23] J. C. M. Zijlmans, P. J. E. Poels, J. Duysens, J. vanderStraaten, T. Thien, M. A. vantHof, H. O. M. Thijssen, and M. Horstink, "Quantitative gait analysis in patients with vascular parkinsonism," *Movement Disorders*, vol. 11, pp. 501–508, 1996.

[24] B. Efron and R. J. Tibshirani, *An Introduction to the Bootstrap*. New York: Chapman & Hall, 1993.

[25] K. Y. Tong and M. H. Granat, "A practical gait analysis system using gyroscopes," *Med. Eng. Phys.*, vol. 21, pp. 87–94, 1999.
[26] A. Salarian, H. Russmann, F. J. G. Vingerhoets, P. R. Burkhard, Y. Blanc, C. Dehollain, and K. Amininan, "An ambulatory system to quantify bradykinesia and tremor in parkinson's disease," in *Proc. 4th Int. IEEE EBMS Special Topic Conf. Inform. Technology Applications Biomedicine*, Birmingham, U.K., Apr. 2003, pp. 35–38.

Arash Salarian (M'04) was born in Isfahan, Iran, in 1972. He received the B.Sc. degree in computer engineering from Isfahan Iniversity of Technology (IUT), Isfahan, Iran, in 1993 and the M.Sc. degree in computer architecture from Shairf University of Technology (SUT), Tehran, Iran, in 1997. He is currently working toward the Ph.D. degree in biomedical engineering at the Laboratory of Movement Analysis and Measurement, Swiss Federal Institute of Technology, Lausanne.

His research interests include biomedical signal processing, ambulatory systems, and movement analysis.

Heike Russmann received the M.D. degree from the University of Freiburg, Germany, in 1985 and the FMH for the specialization in neurology in March 2003.

She now holds the position of Junior Registrar in the Unit of Neurodegenerative Diseases, Department of Neurology, University Clinics of Lausanne, Lausanne, Switzerland. Her research interests include movement disorders and their neurophysiology.

François J. G. Vingerhoets was born in Neuchâtel, Switzerland, in 1958. He received the M.D. degree in 1982 and the medical doctorate (DM) from the University Clinics of Lausanne, Lausanne, Switzerland, in 1987. He completed his training as Neurologist in Lausanne in 1991.

From 1992 to 1995, he worked as a Clinical and Research Fellow at the Neurodegenerative Disorders Center, University of British Columbia, Vancouver, Canada. He joined the Neurology Department, University Hospital of Geneva, Geneva, Switzerland, in 1995. He has been a Privat Docent at Geneva University since 2000. In 1998, he joined the Neurology Department, University Hospital of Lausanne, where he is Head of the Neurodegenerative Disorders Unit. He has been an Associate Professor in Neurology at the University of Lausanne since 2003. He is the author of nearly 100 scientific publications comprising 70 peer-review papers.

Catherine Dehollain (M'93) was born in Paris, France, in 1957. She received the M.Sc. degree in electrical engineering in 1982 and the Ph.D. degree in 1995, both from the Swiss Federal Institute of Technology (EPFL), Lausanne, Switzerland.

From 1982 to 1984, she was a Research Assistant at the Electronics Laboratories (LEG) of EPFL. In 1984, she joined the Motorola European Center for Research and Development, Geneva, Switzerland, as a Design Engineer. In 1990, she returned to EPFL as a Senior Assistant at the Circuits and Systems Laboratory where she did her Ph.D. on impedance broadband matching. In 1995, she joined the LEG where she is currently the Group Leader of the RF design team. In particular, she has been the Technical Project Manager of the Esprit European project SUPREGE and Partner of the ISTEuropean project SODERA. She currently participates to the IST European project WIDE-RF and to the IST European project MuMoR. Since 1998, she has been a Lecturer at EPFL, in the area of HF and UHF circuits and techniques. Her technical interests include low-power analogue circuit techniques and electrical filters. She is author or co-author of 30 scientific publications and of three scientific books in the field of electrical filters and RF circuit design.

Yves Blanc received the physical therapy diploma in 1968, the teacher diploma in 1971, the M.Sc. degree in kinesiology and biomechanics in 1981, and the Ph.D. degree in rehabilitation in 1995. from the Catholic University of Louvain (UCL), Bleguim.

He has been teaching functional anatomy, kinesiology, and biomechanics at graduate and post graduate levels. Since 1977, he has been in charge of the Laboratoire de Cinésiologie, Hôpitaux Universitaires de Genève, Genève, Switzerland.

Pierre R. Burkhard received the M.D. degree from the Swiss Confederation in 1982, the medical doctorate from the University of Geneva, Geneva, Switzerland, and the FMH certification in neurology in 1994.

He trained in Geneva and was a Research Fellow at the Parkinson's Institute, CA, from 1995 to 1997. As an Associate Physician, he is currently running the Movement Disorders Clinic of the Neurology Department, University Hospital of Geneva.

Kamiar Aminian (M'89) received the electrical engineering degree in 1982 and the Ph.D. degree in biomedical engineering in 1989 from the Swiss Federal Institute of Technology—Lausanne (EPFL).

He worked as a Research Associate in the Metrology Laboratory, EPFL and as an Assistant Professor at Sharif University of Technology, Tehran, Iran. In January 2002, he joined the Institute for Biomedical Engineering of EPFL where he is the Head of the Laboratory of Movement Analysis and Measurement. He teaches in the areas of sensors and medical instrumentation. His research interests include transducers, movement analysis, ambulatory systems, and biomedical signal processing.