Automatic detection of inertial sensor orientation for movement analysis in Parkinson's disease

Jennifer Maier, Samuel Reinfelder, Jens Barth, Jochen Klucken, Bjoern M. Eskofier

Digital Sports Group, Pattern Recognition Lab, Department of Computer Science
Friedrich-Alexander-University Erlangen-Nuremberg
Erlangen, Germany
jennifer.maier@medtech.stud.uni-erlangen.de

Abstract—**To analyze the gait of Parkinson's patients and to objectively rate their disease stage development all over the day, low cost inertial sensors can be used. A current challenge is to deal with the potentially unknown sensor position. When measurements are recorded over a longer time period and in an unsupervised environment, sensor position can vary. Therefore, it is necessary to detect the orientation of the sensors. In this study, an algorithm for the automatic orientation detection of an inertial sensor placed on the instep is presented. The detected orientation was used for a subsequent transformation of the sensor data to the coordinate system of recorded reference data. The systems showed a correlation of up to 0.99. Afterwards, a step segmentation of the recorded walking data was computed. Conclusively, the results of a sagittal angle comparison of the transformed gait data of different population groups are shown. With the presented methods, the feasibility of unsupervised long term monitoring of PD patients in daily life is increased.**

Keywords: long term monitoring, gait analysis, data transformation, sagittal angle.

I. INTRODUCTION

One of the most common neurologic diseases in modern times is Parkinson's disease (PD). PD is caused by slowly necrotizing neurons, causing motor symptoms like tremor, bradykinesia and rigidity. According to current medical knowledge, PD can not be cured. Therefore, the therapy mainly focuses on treating the symptoms of the patients with individual medication. This medication always has to be adjusted to the patient's current disease stage, which is mirrored in the physical symptoms.

To assess the continuous course of locomotive disorders in PD in everyday life over a longer time period, at present the only possibility is to rely on a patient's diary. In these diaries, the patient registers the current state of motion disorder at regular intervals. Since these ratings are always subjective, it is necessary to find a way to provide the physicians with objective parameters from long term studies.

In previous studies, such as [1], the patient's physical state was analyzed objectively in a controlled environment. An inertial sensor unit was attached to the lateral side of the shoe below the ankle. The movement of the patient was recorded during 10m walks and other standardized tests. Such a supervised data acquisition is not possible when data is recorded over a longer time period in daily life. To make an objective long term monitoring possible, it is therefore necessary to integrate the measurement in the patient's daily life.

Especially the sensor's position at the exterior of the foot may bother the patient in everyday activities, e.g. the patient's foot could get caught by a chair. Therefore, it is necessary to find a new position where the sensor does not handicap the patient. Placing the sensor on the instep and securing it with the straps of the shoe could be a possible solution. To use algorithms designed for the sensor on the exterior of the shoe, the new position of the sensor has to be detected and afterwards the recorded data has to be transformed. Furthermore, it is possible that the sensor is slightly moving during the recording of a day, therefore the position is not always the same.

For the gait analysis in [2], Bamberg et al. used the gravitational vector to compute the angles of inclination of two of the accelerometer axes. These angles were subsequently used for the computation of gait parameters. A similar approach was used in this paper. In contrast to [2], all three accelerometer axes were used for the detection of the sensor's orientation.

In [3], Wu et al. used a calibration method based on the motion signature to detect an inertial sensor's orientation and to correct for sensor misplacement. For the classification of the current disease stage of a patient, it is necessary to compare the recorded gait to healthy subjects. However, the walking movements of PD patients and healthy subjects are different and have a different motion signature. Therefore, this orientation detection can not be used for the presented problem. The method described in this paper does not rely on a motion signature and works for any subject.

In this paper, an algorithm to detect the orientation of the sensor on the instep is described. Afterwards, recorded walking data from the instep is transformed to match the data from the side of the foot. A step segmentation algorithm is applied to the transformed data and the sagittal angles of the extracted steps of different groups are compared.

II. METHODS

A. *Sensor Setup*

The inertial sensor produced by Shimmer [4] was mounted on the instep of the shoe secured by the shoestrings. For the validation of the developed algorithm, another sensor unit was placed on the exterior of the shoe below the ankle as seen in Fig. 1. Each sensor unit consisted of a three axial accelerometer and a three axial gyroscope with a range of ±6g and ±500°/s at a sampling frequency of 204.8Hz.

B. *Study Design*

For the validation of the algorithm, walking data of three groups was recorded. In the first group were Parkinson patients (aged 62.83 ± 8.11) with movement impairments rated with the Unified Parkinson's Disease Rating Scale – Part III (UPDRS motorscore) of 16.8 ± 2.10. The second group consisted of elderly healthy subjects (aged 66.33 ± 8.04), the third group of young healthy probands (aged 24.5 ± 1.89). Each group consisted of 6 subjects. The performed exercise was a 40m walk with a turn every 10m. The subjects had to stand still at the beginning of every recording to gather the data needed for the estimation of the orientation of the sensor on the instep.

C. *Orientation detection and data transformation*

The orientation detection of the sensor on the instep was used to transform the data recorded at this position to the coordinate system of the sensor on the exterior. For this purpose, it was assumed that one of the axes of the Shimmer unit placed on the exterior was normal to the horizontal plane and one axis was aligned with the walking direction (see vectors $\boldsymbol{n}$ and $\boldsymbol{w}$ in Fig. 1).

When the subject was standing still and the sensor on the instep was in its resting position, the accelerometer measured the earth's gravitation of 9,81m/s resp. 1g, split into its three axes. This gravitation vector $\boldsymbol{g}$ (see Fig. 1) was therefore used as a normal for the horizontal plane and as the first base vector for the new coordinate system. The two other axes were not defined by $\boldsymbol{g}$. By using only the gravitational vector $\boldsymbol{g}$, it was therefore not possible to calculate the direction that the sensor was pointing to. For this study, prior knowledge was used to address this problem. The x-axis of the sensor that was placed on the instep was assumed to be lying in the sagittal plane of the proband's body just as the axis of the

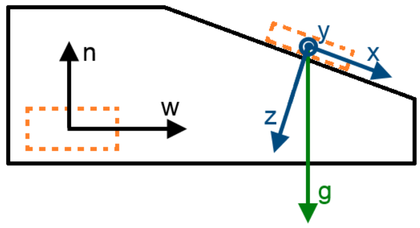

Figure 1: View on the sagittal plane of the foot symbolized by the black frame. Sensors are displayed as orange boxes with dashed lines. $\boldsymbol{n}$: normal vector of the horizontal plane, $\boldsymbol{w}$: walking direction, $\boldsymbol{x}$, $\boldsymbol{y}$, $\boldsymbol{z}$: axes of sensor on instep, $\boldsymbol{g}$: gravitation vector

sensor on the side of the shoe that was aligned with the walking direction (see Fig. 1). Hence, to get the second base vector $\boldsymbol{x'}$ of the new coordinate system, the x-part of the gravitation vector was orthogonally projected onto the horizontal plane defined by the gravitational vector:

$$\boldsymbol{x}' = \boldsymbol{x} - \frac{\langle \boldsymbol{g}, \boldsymbol{x} \rangle}{\|\boldsymbol{g}\|^2} \cdot \boldsymbol{g}$$

The third base vector for the new coordinate system was then computed as the cross product of $\boldsymbol{g}$ and $\boldsymbol{x'}$.

Once the base vectors of the target coordinate system were estimated, the recorded data could be transformed using a direction cosine matrix (DCM) [5]. This matrix allows the transformation of three dimensional data from an inertial coordinate system to a target coordinate system, if the base vectors of both systems are known.

In order to validate the presented data transformation, the correlation coefficient between the data recorded on the exterior of the foot and the transformed data from the instep was computed for each axis of the accelerometer and the gyroscope. First this correlation was calculated for every subject individually, then the mean value of the correlation for each group and the mean over all subjects were computed.

D. *Computation of the sagittal angles*

After the transformation of the data, the angle progression of the foot during a step was computed. The movement during walking is mainly taking place in the sagittal plane of the subject's body. Hence, the gyroscope axis that measures the angular velocity in the sagittal plane was used to compute the mean course of the sagittal angle of the three compared groups.

For this purpose, the pipeline shown in Fig. 2 was executed. First, the step sequences in the data were extracted using Subsequent Dynamic Time Warping (DTW) as described in [6] and [7]. Since the accuracy of this method is between 87% and 99% [6], the results were corrected manually.

The executed Subsequent DTW yields steps that start and end with a peak close to the toe off, where

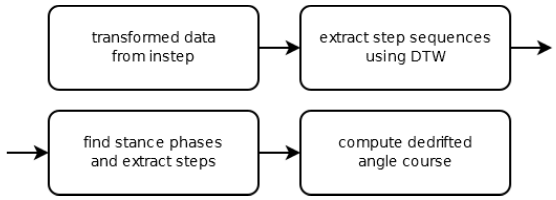

Figure 3: Pipeline for the computation of the sagittal angle course

the foot is leaving the ground. Gyroscope measurements are always afflicted with drift, which can only be subtracted out in phases where the angular velocity is zero. This is the case in stance phases, where the subject's whole foot touches the ground. In walking data, only at these stance phases the zero velocity update can be performed [8]. For that reason, the extracted steps of the DTW could not directly be used for the computation of the angle course. They could, however, be used to get the step sequences where the subject was walking on a straight line without turning. In these step sequences, the beginnings of the stance phases could be detected using a threshold based method. A step in an extracted step sequence was then defined as the movement between two beginnings of stance phases. For the subsequent comparison of the individual steps, each extracted step was linearly interpolated to a length of 250 samples.

By integrating the measured angular velocity of a step over time, its angle course was computed. To subtract the drift, a linear dedrifting was applied on the computed step angles.

This pipeline was executed for every subject of each group. Afterwards the mean angle course and the 95% confidence interval for each group over all angle courses of the group were computed.

III. RESULTS

A. *Accuracy of the data transformation*

The computed correlations over all subjects were 0.844, 0.297 and 0.548 for the accelerometer axes (AX, AY, AZ) and 0.843, 0.969 and 0.984 for the gyroscope axes (GX, GY, GZ). A detailed overview of the correlations for the three groups is shown in Table 1.

TABLE I. CORRELATION BETWEEN THE TRANSFORMED INSTEP DATA AND THE DATA FROM THE SIDE.

	Correlation coefficient					
	AX	***AY***	***AZ***	***GX***	***GY***	***GZ***
Patient	0.834	0.203	0.499	0.834	0.960	0.980
Elderly	0.845	0.305	0.546	0.884	0.975	0.990
Young	0.852	0.384	0.599	0.813	0.971	0.984
Overall	0.844	0.297	0.548	0.843	0.969	0.984

B. *Comparison of the sagittal angles*

The mean angle course and the 95% confidence interval for each group scaled to 250 samples per step are plotted in Fig. 3. The angle courses start with the stance phase, which lasts on average 68 samples for the patients, 75 for the elderly and 49 for the young subjects. The angle courses then have two extrema, a minimum when the foot was swung to the back and a maximum shortly before the foot was touching the ground again after the swing phase. These extrema have a mean amplitude of -68.6 and 18.4 degrees for the patients, -62.5 and 21.8 degrees for the elderly and -74.8 and 27.6 degrees for the young subjects.

IV. DISCUSSION AND OUTLOOK

Tab. 1 shows that, on average, the correlation coefficients for the gyroscope axes (0.81 to 0.99) are higher than the values for the accelerometer axes (0.20 to 0.85). This observation can be explained by considering the different positions of the two sensors. The foot is assumed to be a rigid body and the motion of the foot while walking mainly is a turning movement. The measured angular velocity is independent of the distance of the sensor from the turning axis and is thereby the same for every point on a rigid body. This does not apply to the acceleration, which is highly dependent on the distance from the turning axis because of the leverage. Since the sensor on the instep was not only differently oriented than the sensor on the exterior, but was also located at a different position, the two sensor units did not measure the same acceleration. Hence, the acceleration can not easily be transformed using the described method. A very low correlation is observed especially in the accelerometer y-axis, which is lying in the sagittal plane, where the main movement takes place while walking. During one step, the axis the foot is turning around in the sagittal plane moves from the toe over the midfoot to the heel. Therefore the distances of the two sensors to this axis also change during a step, leading to the recording of different accelerations.

The comparison of the sagittal angle courses of the different test groups shows that the stance phase of the young subjects lasts shorter than the one of the elderly subjects and patients. With progressing age, the human

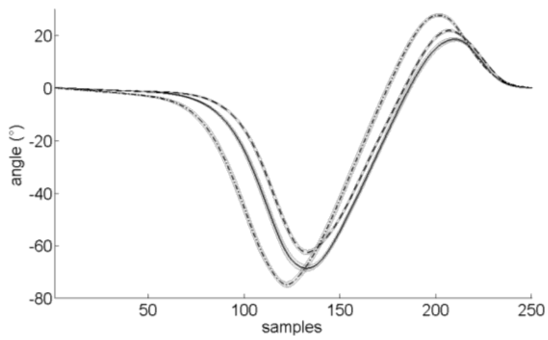

Figure 2: mean angle courses and their confidence intervals for the three groups. Patients: solid line, elderly healthy subjects: dashed line, young healthy subjects: line composed of dots and dashes

gait becomes more insecure and unbalanced [9], hence the elderly subjects and patients may have tried to leave the foot on the ground longer to better keep their balance. Furthermore the amplitude of the extrema is higher for the young subjects than for the other two groups. A reason for this could lie in the reduced mobility of elderly people [10] restraining them from swinging their feet further. Between the patients and elderly subjects the only significant differences are that the absolute value of the minimum angle is 6.1 degrees higher for the patients, whereas the maximum angle is 3.4 degrees higher for the elderly people. The fact that there is only a small difference between patients and elderly subjects can be explained by considering the UPDRS of the participating patients. On a scale from 0 (no disability) to 199 (total disability), the patients' scales lie on average at 16.8 corresponding to an early stage of PD.

In further studies, the problems with the transformation of the acceleration data should be addressed. It may be possible to detect the position of the turning axis to compute a transformation that makes the accelerations from different positions comparable. Furthermore the sensors mounted onto the shoes are not always aligned with the walking direction as assumed in this paper. An exact alignment using the recorded data could improve the results of the transformation algorithm. Moreover, the number of probands for upcoming studies should be increased to get more representative data.

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