Wearable Static Posturography Solution Using a Novel Pressure Sensor Sole

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***Abstract*— Static posturography is an important measurement in the diagnostic workup for patients with postural instability. New wearable sensor technologies enable researchers to use in-shoe pressure soles in the home environment and outdoor applications. In this study a newly developed in-shoe pressure sole was used for calculating the sway path and 95 % confidence ellipse area as the standard parameters of typical static posturography. Insole posturography was validated on 24 subjects by a state of the art pressure plate assessment during three static posturography conditions (*eyes open*, *eyes closed* and *barefoot*). The adaptive low pass filtered data resulted in an overall correlation of 0.63 to 0.78 for the sway path and 0.66 to 0.79 for the 95 % confidence ellipse area. Individual correlations of up to 0.97 for the sway path and 0.99 for the 95 % confidence ellipse area could be obtained. Future applications could utilize the mobile advantage of in-shoe pressure soles and measure static and dynamic posturography in clinical and home environments.**

I. INTRODUCTION

Static posturography is an important balance control tool and is frequently used for patients with postural instability or movement disorders like Parkinson's disease [1], somatosensory vertigo [2] or other diseases affecting the balance control. The gold standard for static posturography is currently the use of a pressure or force plate. Although these systems are easy to use in a clinical environment, they are stationary and not usable in mobile applications. Wearable in-shoe pressure soles overcome this problem but are technically more complicated because each foot is measured independently and only pressure on distinct areas of the sole is considered [3]. Several studies presented or compared in-shoe pressure soles regarding their technical abilities, limitations and application in common research problems [3], [4], [5]. These systems were validated with technical parameters.

In the scope of a broader project [6] an in-shoe pressure sole was used for static posturography measurements (IEE, Contern, Luxembourg). Not only technical parameters were measured but standard clinical postural analysis parameters like the sway path and the 95 % confidence ellipse area were calculated and validated by a pressure plate (zebris Medical GmbH, Isny im Allgäu, Germany). The focus of this work was to evaluate an optimal Butterworth low pass filter for the raw data and investigate the influences of different postural measurement conditions on the resulting postural parameters.

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II. METHODS

A. Data Collection

24 test subjects free of medical conditions associated with postural imbalance (35.8 $\pm$ 8.0 years, Table I) were recruited after obtaining written informed consent. The approval from the ethical committee was received (Re.-No. 4208, 21.04.2010, IRB, Medical Faculty, University of Erlangen-Nürnberg, Germany).

Three conditions were chosen for the validation of the in-shoe pressure sole based postural parameters. In the first condition (CON1), the participant was advised to stand still for $30\,s$. The shoes were placed parallel and touched each other on the medial side [7]. Meanwhile, the participant focused on a cross $3\,m$ in front of her/him at approximately the height of the eyes. Each trial was conducted three times with intermediate resting phases of one minute. The second condition (CON2) was similar except that the participants had their eyes closed which is widely used in literature for omitting balance control influences of the visual cortex. For analysis of the impact of the shoes on the measurement, a third condition (CON3) was conducted. The procedure was the same as for CON2 whereas the participants were barefooted. The used measurement conditions were chosen carefully regarding their validity and are in compliance to the proposed parameters of Scoppa et al. [8].

For CON1 and CON2 all 24 participants were recorded which led to 72 measurements each. CON3 was recorded with 11 participants which resulted in 33 measurements.

TABLE I
CHARACTERISTICS OF TEST SUBJECTS.

Subjects			
Sex (m/f)	14:10		
Age (years)	35.8	$\pm$	8.0
Height (cm)	167.4	$\pm$	10.1
Weight (kg)	78.0	$\pm$	19.1
BMI	25.1	$\pm$	6.2

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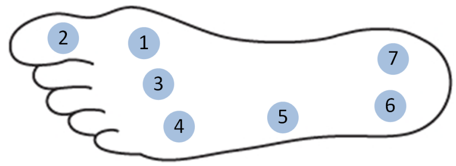

Fig. 1. Schematic representation of the left sensor insole; the right insole is designed mirror inverted.

B. Measurement Systems

The in-shoe pressure soles, provided by IEE (IEE, Contern, Luxembourg), were connected to the recording system from WalkinSense (Kinematix, Porto, Portugal). Shoe insoles with UK sizes 6, 8 and 11 were used which contained each seven measure points (Fig. 1). The data were transmitted wirelessly via Bluetooth to a laptop.
A pressure plate from zebris (zebris Medical GmbH, Isny im Allgäu, Germany) was chosen as the gold standard system (Fig. 2). The technical specifications of both systems are listed in Table II.

TABLE II
CHARACTERISTICS OF THE SENSOR SYSTEMS.

	Sensor Systems	
	WalkinSense®	zebris®
Sampling Rate (Hz)	100	50
Measuring Unit	kg/cm^2	N/cm^2
PC-Interface	Bluetooth	USB
Sensor Type	Piezoresistive	Capacitive

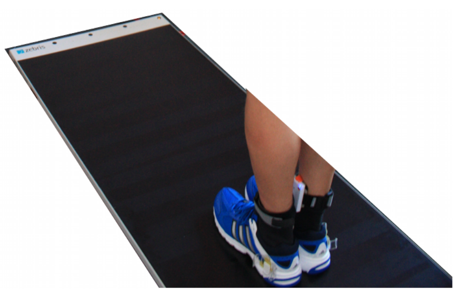

Fig. 2. Test subject standing on the zebris ® pressure plate.

C. Parameters

For a quantitative evaluation of the ability of a participant to maintain balance, the displacement of the center of gravity (COG) is a meaningful parameter [9]. The COG is the projection of a person's center of mass onto the base of support [9]. If the COG is placed outside the supporting area of a resting person's feet, the person will fall without further intervention [10]. The direct measurement of the COG is, due to permanent balance correction movements of the human body, not possible. For static conditions, the center of pressure (COP) varies around the COG position with a higher magnitude and frequency as the COG [11] and can be used instead.
The COP in the media-lateral (ML) and anterior-posterior (AP) direction can be calculated as the weighted sum of all 14 pressure sensors from both sensor soles with respect to the geometrical placement of the sensors (Fig. 1) and the distance between the shoes:

$$COP_{ML} = \frac{1}{P}\sum_{i=1}^{14} P_i(ML)_i \quad (1)$$

$$COP_{AP} = \frac{1}{P}\sum_{i=1}^{14} P_i(AP)_i \quad (2)$$

P_i is the pressure of sensor i and P is the summarized pressure of all sensors. $(ML)_i$ and $(AP)_i$ are the spatial positions of sensor i. The geometrical dimensions were adjusted to the used shoe and therefore insole sizes. The resulting COP over time in the transverse plane is therefore (COP_{ML}, COP_{AP}).
Based on the COP several parameters can be calculated. In this work two parameters are considered more closely:

1) The sway path (SP) is the accumulated distance of the COP in a specified time interval. For N sampling points, the sway path calculates to:

$$SP = \sum_{i=2}^{N} \sqrt{(ML_{i-1} - ML_i)^2 + (AP_{i-1} - AP_i)^2} \quad (3)$$

2) The 95 % confidence ellipse area (EA) is the area of an ellipse that is computed to approximately enclose 95 % of the COP data points (Fig. 3). It incorporates not only an area of movement but also the dominant direction of movement. For the approximation a bivariate Fisher-Snedecor distribution is used [12]:

$$EA = 2\pi F_{.05[2,N-2]} \sqrt{SD_{ML}^2 SD_{AP}^2 - COV_{ML,AP}^2} \quad (4)$$

SD_{ML} and SD_{AP} are the media-lateral and anterior-posterior standard deviations of the COP. $COV_{ML,AP}$ is the covariance between the media-lateral and anterior-posterior data. $F_{.05[2,N-2]}$ is the F statistic for a bivariate distribution with a 95 % confidence level for N sampling points. $F_{.05[2,N-2]}$ can be approximated to 3.00 for large sample sizes ($N > 120$) so that the equation can be simplified [13]:

$$EA = 6\pi \sqrt{SD_{ML}^2 SD_{AP}^2 - COV_{ML,AP}^2} \quad (5)$$

For stable values of the EA, it is advised to set the time interval for the measurement not less than 30 seconds [8].

D. Filter Optimization

The COP trajectory of the in-shoe pressure sole showed noise and high frequency characteristics (Fig. 3). For this reason, a Butterworth low pass filter was applied on the COP data of both systems. The use of low pass Butterworth filters is common in literature:

- Baratto et al. [14]: 2nd order, cutoff frequency of 10 Hz
- Prieto et al. [13] and Betker et al. [15]: 4th order, cutoff frequency of 5 Hz
- Andreasen et al. [16]: 3rd order, cutoff frequency of 5 Hz
- Adkin et al. [17]: 2nd order, cutoff frequency of 5 Hz
- Gage et al. [18]: 4th order, cutoff frequency of 3 Hz

Summarizing the mentioned literature, the order of the Butterworth filter should be between 2 and 4, the cutoff frequency between 3 and 10 Hz. Considering the recorded data, even lower cutoff frequencies seemed relevant. To avoid overfitting, a leave-one-out (LOO) cross-validation for the 24 participants was conducted for obtaining optimal Butterworth filter parameters. The unweighted sum of the SP and EA was used as optimization criteria. The chosen values for the grid search are listed in Table III.

It was technically not possible to start both sensor systems simultaneously and therefore a non-constant time delay of less than 0.5 s occurred. The signals were temporal cross-correlated in the LOO cross-validation. For the impact of the time delay, the results are given also without this correction.

TABLE III

PARAMETERS USED IN THE LOO.

Parameters and Ranges		
Parameter	Range	Step Size
Order N	2-6	2
Cutoff Frequency ω_c	1-6 Hz	0.5 Hz

III. RESULTS

The optimal Butterworth cutoff frequency was 1.50 $\pm$ 0.15 Hz with and without time correction. The optimal mean order was 4 $\pm$ 0 with time correction and 4.08 $\pm$ 0.41 without time correction. The data was therefore filtered with a 4th order Butterworth filter with a cutoff frequency of 1.50 Hz.

The overall Pearson correlations between both systems for the sway path and the 95 % confidence ellipse area are listed for all conditions in Table IV. Time corrected individual correlations combined from all three conditions were up to 0.97 for the sway path and up to 0.99 for the 95 % confidence ellipse area. For CON1, the correlations for the SP and EA between both systems are plotted in Fig. 4. The corresponding Bland-Altman plots are in Fig. 5.

IV. DISCUSSION

The results demonstrated good correlations between the in-shoe sensor sole and the zebris pressure plate for static posturography measurements. Mean correlations of all conditions of 0.73 for the sway path and 0.78 for the 95 % confidence ellipse area could be obtained. This makes the system usable for clinical assessments. Nevertheless, several

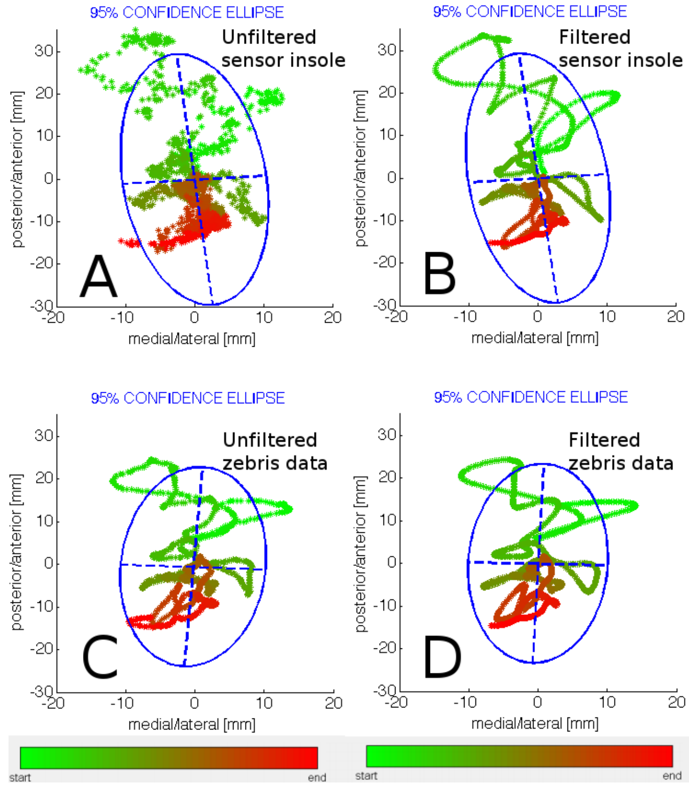

Fig. 3. Representative representation of the COP measured with the insole sensor and the zebris pressure plate. A & C is unfiltered, B & D is low pass filtered (Butterworth, 3rd order, 1.75 Hz cutoff frequency).

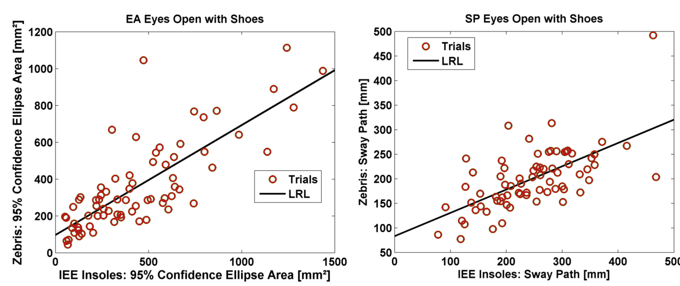

Fig. 4. Correlation of the EA and SP between insoles and zebris pressure plate. LRL is the linear regression line.

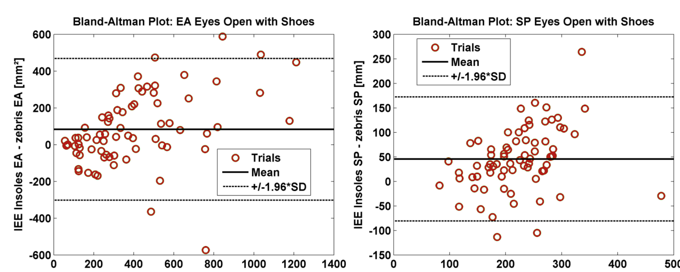

Fig. 5. Bland-Altman plot of the EA and SP between insoles and zebris pressure plate.

TABLE IV

PEARSON CORRELATION COEFFICIENT BETWEEN SENSOR SOLE AND ZEBRIS PRESSURE PLATE.

Pearson Correlation Coefficient with LOO				
Time Correction	Yes		No	
Data Set	r of EA	r of SP	r of EA	r of SP
CON1:	0.79	0.63	0.78	0.63
CON2:	0.77	0.77	0.77	0.76
CON3:	0.66	0.71	0.67	0.72
Complete Data:	0.73	0.78	0.73	0.78

outliers are degrading the result (Fig. 4 and Fig. 5).

One possible reason is the design and number of the used sensors. It was assumed that pressure measured by a sensor was equally distributed over the whole sensor area. Considering the size of the sensors this might be a major cause for errors. Another reason might be that not all pressure is measured by the sensors because pressure was directly distributed to the shoe on areas not covered by a sensor [3].

The results improved considerably with the Butterworth low pass filter. The parameters of the filter were stable and did not change for different subgroups. Nevertheless, groups with balance disorders may have a different frequency characteristic of the COP and the LOO cross-validation should be conducted again.

No considerable difference could be seen between condition 2 and condition 3. This indicates that the effect of the deformation of the sensor sole and other effects of the shoe did not influence the results. Better results can therefore not be obtained by using more flat shoes. The temporal cross-correlation of both measurement systems did not improve the results.

The cross-validated Butterworth filter improved the calculated parameters considerably without overfitting. Future applications could utilize the mobile advantage of pressure insoles and transfer the clinical workup setting to a home environment for measuring postural parameters of static and dynamic posturography.

V. CONCLUSION

The current study validated standard postural analysis parameters of a state-of-the-art in-shoe pressure sole with a pressure plate. A LOO cross-validation method was used for obtaining the best parameters for the used Butterworth low pass filter. For future applications, static and dynamic posturography in a home environment should be considered.

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REFERENCES

[1] R. Marchese, M. Bove, and G. Abbruzzese, "Effect of cognitive and motor tasks on postural stability in parkinson's disease: a posturographic study," *Movement Disorders*, vol. 18, no. 6, pp. 652–658, 2003.

[2] L. Baratto, R. Capral, P. Morassol, P. Serpero, and G. Spada, "Clinical application of posturography for the treatment of somatosensory vertigo," *Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, vol. 18, pp. 571–572, 1996.

[3] A. H. A. Razak, A. Zayegh, R. K. Begg, and Y. Wahab, "Foot plantar pressure measurement system: A review," *Sensors*, vol. 12, no. 7, pp. 9884–9912, 2012.

[4] F. Neaga, D. Moga, D. Petreus, M. Munteanu, and N. Stroia, "A wireless system for monitoring the progressive loading of lower limb in post-traumatic rehabilitation," in *International Conference on Advancements of Medicine and Health Care through Technology*. Springer, 2011, pp. 54–59.

[5] K. J. Chesnin, L. Selby-Silverstein, and M. P. Besser, "Comparison of an in-shoe pressure measurement device to a force plate: concurrent validity of center of pressure measurements," *Gait & Posture*, vol. 12, no. 2, pp. 128–133, 2000.

[6] J. Klucken, J. Barth, P. Kugler, J. Schlachetzki, T. Henze, F. Marxreiter, Z. Kohl, R. Steidl, J. Hornegger, B. Eskofier *et al.*, "Unbiased and mobile gait analysis detects motor impairment in parkinson's disease," *PloS one*, vol. 8, no. 2, p. e56956, 2013.

[7] F. Scoppa, R. Capra, M. Gallamini, R. Schiffer, and S. D'Ottavio, "Feet position in the stabilometric assessment and postural stability," Faculty of Medicine and Dental Surgery, Sapienza University of Rome, Rome, Italy, Tech. Rep., 2013.

[8] F. Scoppa, R. Capra, M. Gallamini, and R. Schiffer, "Clinical stabilometry standardization: basic definitions–acquisition interval–sampling frequency," *Archives of Physical Medicine and Rehabilitation*, vol. 37, no. 2, pp. 290–292, 2013.

[9] R. J. Schilling, E. M. Bollt, G. D. Fulk, J. D. Skufca, and C. J. Robinson, "A quiet standing index for testing the postural sway of healthy and diabetic adults across a range of ages," *Biomedical Engineering, IEEE*, vol. 56, no. 2, pp. 292–302, 2009.

[10] Y.-C. Pai and J. Patton, "Center of mass velocity-position predictions for balance control," *Journal of biomechanics*, vol. 30, no. 4, pp. 347–354, 1997.

[11] T. E. Prieto, J. B. Myklebust, and B. M. Myklebust, "Characterization and modeling of postural steadiness in the elderly: a review," *IEEE Transactions on Rehabilitation Engineering*, vol. 1, no. 1, pp. 26–34, 1993.

[12] H. Suarez, D. Geisinger, A. Suarez, X. Carrera, P. Spiller, and V. Lapilover, "Postural strategies in normal subjects and in patients with instability due to central nervous system diseases after sudden changes in the visual flow," *Acta Oto-Laryngologica*, vol. 128, no. 4, pp. 398–403, 2008.

[13] T. E. Prieto, J. B. Myklebust, R. G. Hoffmann, E. G. Lovett, and B. M. Myklebust, "Measures of postural steadiness: differences between healthy young and elderly adults," *IEEE Transactions on Biomedical Engineering*, vol. 43, no. 9, pp. 956–966, 1996.

[14] L. Baratto, P. G. Morasso, C. Re, and G. Spada, "A new look at posturographic analysis in the clinical context: sway-density vs. other parameterization techniques," *Motor Control*, vol. 6, no. 3, pp. 246–270, 2002.

[15] A. Betker, T. Szturm, and Z. Moussavi, "Application of feedforward backpropagation neural network to center of mass estimation for use in a clinical environment," *Proceedings of the 25th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (IEEE Cat. No.03CH37439)*, pp. 2714–2717, 2003.

[16] L. N. S. Andreasen, W. Jensen, P. H. Veitink, and J. J. Struijk, "Natural sensory feedback for control of standing," *18th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, pp. 441–442, 96.

[17] A. L. Adkin, J. S. Frank, M. G. Carpenter, and G. W. Peysar, "Fear of falling modifies anticipatory postural control," *Proceedings of the 25th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (IEEE Cat. No.03CH37439)*, vol. 143, no. 2, pp. 160–170, 2002.

[18] W. H. Gage, D. A. Winter, J. S. Frank, and A. L. Adkin, "Kinematic and kinetic validity of the inverted pendulum model in quiet standing," *Gait & Posture*, vol. 19, no. 2, pp. 124–132, 2004.