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Taking balance measurement out of the laboratory and into the home: discriminatory capability of novel centre of pressure measurement in fallers and non-fallers

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Abstract – **We investigated three methods for estimating centre of pressure excursions, as measured using a portable pressure sensor matrix, in order to deploy similar technology into the homes of older adults for longitudinal monitoring of postural control and falls risk. We explored the utility of these three methods as markers of falls risk in a cohort of 120 community dwelling older adults with and without a history of falls (65 fallers, 55 non-fallers). A number of standard quantitative balance parameters were derived using each centre of pressure estimation method. Rank sum tests were used to test for significant differences between fallers and non-fallers while intra-class correlation coefficients were also calculated to determine the reliability of each method. A method based on estimating the changes in the magnitude of pressure exerted on the pressure sensor matrix was found to be the most reliable and discriminative. Our future work will implement this method for home-based balance measurement.**

I. INTRODUCTION

Previous falls history and poor balance capabilities have been identified as important risk factors in predicting future falls [1, 2]. Falling without serious injury increases the risk of being placed in a nursing home 3-fold after accounting for cognitive, psychological, social, functional and medical factors. A serious injury increases the risk 10-fold [3]. Falls often result in limitations in everyday activities due to an acquired fear of falling [4]. It is therefore important to identify balance problems at an early stage as the first fall can expose older persons to a cascade of negative physical and psychological consequences. Accordingly, geriatric evaluations generally include balance and mobility assessments, such as the Berg Balance or timed-up-and-go test. However, such functional balance measures usually lack the ability to capture balance impairment in its early stages when no obvious balance problems are manifest [2]. These tests are known to be prone to floor and ceiling effects [5]. Additionally, the subjective evaluation of a person's balance raises obvious reliability issues. Objective balance measurements, based on force-platform technology, that are sensitive enough to reveal subtle deterioration in balance control have therefore been developed for the clinic setting.

Force-platform based measures are considered to be the "gold standard" for objective balance measurement. Centre of pressure (COP) based metrics capture the integrated functioning of the balance control systems. Pajala et al. [2] demonstrated that older people with seemingly intact balance may suffer from early deterioration in their ability to control posture, as measured using COP data. Their study prospectively linked force platform measures with incidence of indoor falls. However, despite this evidence, force-platform technology used to evaluate balance control through measures of postural sway, is not commonly used in a clinical setting. This is mostly due to the prohibitive cost of force platforms. In response to this, recent advances in wearable sensors have resulted in a growing body of evidence that supports the use of on-body accelerometers for quantitative balance assessments. The prospect of clinicians administering an objective balance test based on postural sway, using low-cost inertial sensors is now very much a reality.

However, we propose that there may be some problems with this approach, if not implemented rigorously. The vast number of possible sway-related measures that characterize balance performance have exhibited questionable reliability. Santos et al. [6] have shown poor to moderate reliability of COP measures in young healthy adults. The majority of the variance was attributed to Subject (2%–76%), Subject x Day (0%–24%) and Subject x Day x Trial (16%–79%) variance components depending on the summary measure and condition. The authors suggested that reliability could be improved by averaging measurements between-days than by increasing the number of trials during 1 day. Similarly, de Bruin *et al.*[7] reported poor correlations between two consecutive measurement days for dynamic and static movement, highlighting the necessity of recording further days to assess activity in the geriatric population. This raises a critical question: is a once-off clinic-based measurement of balance truly representative of the state of an individual's balance control system? Falling is referred to as one of the geriatric syndromes because of its complex etiology resulting not only from one or more discrete diseases but also from

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accumulated effects of impairments in multiple systems [8]. It is highly likely therefore, that balance performance from one day to the next is variable, due to the various interactions of intrinsic falls risk factors throughout the day. Duchene and Hewson [9] recently adapted a bathroom scales for the long-term day-to-day evaluation of balance measurement in elderly adults, over a one-year period. Their results showed high variability in day-to-day measurement. However, their study also showed that long-term monitoring of balance in the home could possibly provide a means for early detection of declining balance performance that may relate to future falls. We would suggest that day-to-day variability of balance measures may reveal important insights into neuromuscular control that could be harnessed for effective falls prevention strategies.

This study investigates three different methods for estimating COP excursions measured using a portable pressure sensor matrix, for the calculation of a variety of postural sway parameters. With a long-term view towards longitudinal balance measurement in the homes of older adults, we first needed to determine which method of the three best discriminated a group of fallers and non-fallers. The most discriminatory and reliable method that emerges from this study will be implemented in a future home-based study that investigates the fluctuations of balance measures over time, and how this relates to falling.

II. Methods

A. Data Sample

A convenience sample of 120 community dwelling older adults (57 male and 63 female, mean age: 73.7±5.8 years) were recruited for this study. All participants were community-dwelling, aged ≥60 years, medically stable, able to walk independently (with or without aids), and able to provide written informed consent. The data were acquired from 65 subjects with a self-reported history of falling in the past 5 years, i.e. 'fallers', and 55 'non-fallers'. Non-fallers were participants with no history of falls in the previous 5 years. Fallers were defined as participants who experienced 2 or more falls in the previous 5 years or experienced 1 fall with any of the following criteria: unexplained fall; experienced syncope, presyncope or loss of consciousness; pre-fall symptoms of dizziness/light-headedness; injuries resulting in fracture or major injury; chronic pain/fear of falling (FES-I)/physical disability/depression following the fall [10]. Falling was defined as an unexpected loss of balance resulting in coming to rest on the ground, or an object below the knee level [11].

B. Protocol

Each participant was asked to complete two standing balance tasks: a semi-tandem stance (i.e. the first metatarsal head of one foot placed beside the heel of the other foot), arms by their sides, eyes open, looking straight ahead (EO condition); and a narrow stance, arms by their sides, eyes closed (EC condition). Each task lasted 30 seconds and was repeated 3 times as follows: EO condition, approximately one minute rest, EC condition, seated rest for approximately 3 minutes (x3). For practical reasons this was not a randomized process, however the tasks were alternated as opposed to performed in blocks to avoid any learning or fatigue effects. While many permutations of eyes open, eyes closed and various stances have been used to examine static postural control, these conditions were specifically selected for their reported ability to discriminate between older adults with and without balance impairments [12, 13]. Two nurses stood beside each participant in case of loss of balance. Participants removed their own footwear and performed the tests in non-slip disposable foam slippers.

C. Data Collection

Pressure sensor data for each subject were obtained using a Tactex S4 high density pressure mat (Tactex Controls Inc., Victoria, Canada). The sensor measured 0.915 m x 0.610 m and contained an evenly distributed grid of 72 x 48 plastic optical fibre pressure sensors (Kinotex®) embedded in foam. A 650nm LED shines light through a 'transmit' fibre to a termination point where the amount of light is detected by a 'receive' fibre. The termination point of the send and receive fibre is called a 'taxel'. For each sample of data collected, a frame of data described the pressure applied to the pressure sensing grid at that instant. Changes in pressure detected by sensors resulted in an update to the matrix reflecting the current pressure applied to the mat. A sample and hold algorithm interpolated the data resulting in a constant sampling rate of 10 Hz. All sensor data (streamed via Bluetooth) were synchronously acquired using a custom BioMOBIUS application (http://www.biomobius.org). Data for each test were then exported to text format for subsequent offline analysis in Matlab version 7.11 (Mathworks, Natick, VA, USA).

D. Data Analysis

Three methods were used to estimate the centre of pressure (COP) excursions for each trial. The first two methods have been validated using a force plate in a previous study [14]. The first method, known as centre of all active sensors (CAAS), used an empirical threshold to define an individual sensor as being active and then calculated the absolute centre of all active sensors as an estimate of the COP location per frame. The second method, known as the centroid of heel and toe points (CHAT) calculates the overall centre of the four (automatically detected) heel and toe points, to estimate the COP location for a given data frame [14]. The centroid of each of these areas was then defined as the heel and toe points for each foot respectively. The third method used the individual sensor pressure values along with sensor location to estimate the vertical projection of the centre of mass, centred at the origin. We refer to this method as magnitude of pressure (MOP).

Quantitative balance parameters were estimated using each of the three methods: mean COP distance, root mean squared (RMS) COP distance, sway length, mean sway velocity, mean sway frequency, mean sway frequency (medio-lateral (ML) and anterior-posterior (AP) direction). Prieto et al. referred to the mean sway frequency measures as being proportional to the ratio of the mean sway velocity to the mean sway distance [15]. The average of three trials was used for hypothesis testing.

TABLE I: RESULTS OF EXPLORATORY CROSS-SECTIONAL ANALYSIS OF EACH DERIVED PARAMETER. PARAMETERS OBTAINED FROM EYES OPEN (EO) AND EYES CLOSED (EC) CONDITIONS ARE TABULATED SEPARATELY (MEAN±SD) SIGNIFICANT DIFFERENCES (P<0.05) IN EO AND EC ARE DENOTED BY ^ AND * RESPECTIVELY. ICCS (95% LOWER-UPPER CONFIDENCE INTERVALS) ARE PRESENTED FOR EACH PARAMETER CALCULATED ACROSS ALL PARTICIPANTS. NOTE THAT THESE PARAMETERS ARE DERIVED USING THE NUMERICAL PRESSURE VALUES OF EACH TAXEL, AND ARE THEREFORE ESTIMATES OF THE TRUE PARAMETERS.

Variable	Faller	Non-faller	ICC	Faller	Non-faller	ICC (95% CI)
CAAS		EO			EC	
Mean COP dist	41.52±29.24	41.10±28.88	0.40(0.19-0.56)	34.42±28.91	37.70±24.17	0.56(0.40-0.68)
RMS COP dist	50.31±32.58	50.12±31.49	0.43(0.23-0.59)	41.70±31.40	44.60±26.80	0.58(0.43-0.69)
Sway length	878.14±1028.51	858.49±1165.96	0.46(0.27-0.61)	945.70±1236.88	1261.44±1573.11	0.47(0.28-0.61)
Mean sway velocity	43.91±51.43	42.92±58.30	0.46(0.27- 0.61)	47.28±61.84	63.07±78.66	0.47(0.28-0.61)
Mean sway freq	0.17±0.13	0.20±0.18	0.25(-0.02- 0.45)	0.21±0.16	0.24±0.19	0.50(0.33-0.64)
Mean ML sway freq	0.22±0.16	0.24±0.20	0.21(-0.07- 0.43)	0.23±0.20	0.27±0.26	0.54(0.38-0.67)
Mean AP sway freq	0.20±0.14	0.23±0.24	0.19(-0.10- 0.41)	0.25±0.17	0.28±0.19	0.38(0.17-0.55)
CHAT						
Mean COP dist	3.96±1.35	4.11±1.73	0.50(0.32-0.64)	3.00±1.32	2.69±0.82	0.68(0.56-0.77)
RMS COP dist	4.64±1.73	4.82±2.53	0.48(0.29-0.62)	3.44±1.55	3.06±0.88	0.68(0.56-0.76
Sway length	216.48±123.42	255.29±183.72	0.61(0.47-0.71)	134.07±102.04	178.94±115.24	0.81(0.74-0.86)
Mean sway velocity	10.82±6.17	12.76±9.19	0.61(0.47-0.71)	6.70±5.10	8.95±5.76	0.81(0.74-0.86)
Mean sway freq *	0.44±0.22	0.52±0.32	0.60(0.46-0.71)	0.35±0.22	0.50±0.31	0.82(0.75-0.89)
Mean ML sway freq *	0.55±0.29	0.66±0.45	0.56(0.40-0.68)	0.44±0.30	0.65±0.41	0.78(0.71-0.84)
Mean AP sway freq*	0.46±0.25	0.54±0.33	0.58(0.43-0.70)	0.38±0.23	0.52±0.30	0.73(0.64-0.81)
MOP						
Mean COP dist	14.29±5.29	13.92±4.27	0.78(0.71-0.84)	13.84±4.24	13.05±4.68	0.56(0.40-0.68)
RMS COP dist	16.35±6.01	16.01±5.05	0.79(0.71-0.85)	15.59±4.83	14.78±5.34	0.59(0.45-0.70)
Mean sway length *	632.55±341.47	786.98±485.72	0.83(0.77-0.88)	577.54±386.90	691.56±454.09	0.81(0.75-0.87)
Mean sway velocity*	31.63±17.07	39.35±24.29	0.83(0.77-0.88	28.88±19.34	34.58±22.70	0.81(0.75-0.87)
Mean sway freq	0.35±0.17	0.45±0.28	0.78(0.70-0.84)	0.33±0.19	0.43±0.26	0.84(0.79-0.89)
Mean ML sway freq*	0.38±0.19	0.49±0.32	0.73(0.64-0.81)	0.36±0.20	0.49±0.34	0.85(0.79-0.89)
Mean AP sway freq *^	0.42±0.19	0.55±0.32	0.73(0.63-0.80)	0.41±0.25	0.51±0.31	0.80(0.74-0.86)

E. Statistical Analysis

The Mann-Whitney version of the Wilcoxon rank sum test was used to test for significant differences in each parameter between fallers and non-fallers, to examine the utility of those parameters in assessing falls risk. Intra-class correlation coefficients (ICC(2, k)) were then calculated to assess the reliability of each quantitative balance parameter across the three trials.

III. RESULTS

No parameter significantly discriminated fallers from non-fallers using the CAAS method (Table 1). The CHAT and MOP methods were more discriminative, with significant differences occurring mostly in the EC condition (denoted by * in table 1). Similarly, the CHAT and MOP methods demonstrated the best reliability, with the MOP method demonstrating mostly excellent reliability (i.e ICC>0.75 [16]) for both EO and EC parameters, and the CHAT method demonstrating good to excellent reliability for the EC parameters.

IV. DISCUSSION

The validity and reliability of potential markers of falls risk were investigated in this study using three methods to estimate COP excursions using a floor sensor matrix. A method based on the vertical projection of the centre of mass (MOP) provided the best discrimination between fallers and non-fallers, and the best reliability. A limitation of the CAAS and CHAT methods is that these methods do not account for pressure magnitude, hence are subject to error due to the empirically tuned threshold used. The MOP method used the pressure magnitude at each active sensor, hence it is likely to have provided a more accurate representation of the body's centre of pressure.

The fact that the MOP method discriminated fallers and non-fallers on the basis of sway distance and velocity, and a derived parameter (mean ML sway frequency) in the eyes closed conditions is consistent with a large body of literature [17]. However, it was somewhat unexpected that only one parameter (mean AP sway frequency) differentiated groups in the EO semi-tandem stance condition, despite previous research suggesting that such a task can differentiate fallers and non-fallers [2, 12]. Our data show good to excellent reliability in the EO condition for the MOP method, which may raise a question about the sensitivity of the pressure sensor mat, rather than the accuracy of the measure per se.

The development of methodologies that provide reliable information about a person's balance capabilities for use in home monitoring is an emerging area of research. While previously we have relied on one-off balance measurements - usually acquired in a laboratory setting – to make inferences about the health of a person's postural control system, we now have the opportunity to delve deeper into the day-to-day patterns in balance performance that evolve over time. For now we can only speculate as to how valuable this new information will be, but clearly the current innovations in long-term remote monitoring will become more pervasive in the coming years. Demonstrating the clinical validity and reliability of novel measures as we have done here is an essential first step in this process.

This study has focused on linear measures of postural control. However, it is also possible for the time series produced by each of the three methods to be analysed using nonlinear approaches. Nonlinear analyses have been used to examine complexity in many physiological systems. Previous research by Harbourne et al. [18] used principal component analysis to demonstrate that nonlinear measures provided additional information about postural control in infants that was not captured using standard linear measures. Our future work will explore the reliability and validity of nonlinear measures as potential markers of falls risk in older adults.

Many novel technologies exist for extra-laboratory measurement of balance. Najafi et al. [19] assessed the clinical validity of a wearable sensor system by comparing balance control of healthy subjects with a group of diabetes patients suffering from peripheral neuropathy. The technology enabled screening of balance impairment in these patients in both EO and EC conditions. While wearable sensors represent a low cost, light-weight objective means to evaluate balance, the use of such technologies in long-term home monitoring may pose certain problems due to the burden they may present for the individual e.g. battery charging, attaching the device. Pressure sensitive mats, on the other hand, could fit unobtrusively into a person's home, requiring that person to simply stand still on a pre-determine spot for a few moments every day. Their postural sway data could be recorded and monitored automatically, raising red-flags if a sustained deterioration were to occur. The progress in sophisticated gaming technologies presents the perfect opportunity for low-cost home deployment of balance-monitoring technologies. The Nintendo Wii Balance Board – approximately $100 per unit – has been used for assessment of standing balance and postural control asymmetries [20, 21]. The most robust method that emerged from this investigation (method 3, MOP) could easily be implemented on low cost pressure sensitive hardware. Future work will explore this possibility with a view to deploying cost-effective, reliable and clinically valid balance monitoring tools into the home, as well as developing the potential for individualized balance biofeedback training for the prevention of falls.

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