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Enhancing clinical measures of postural stability with wearable sensors*

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***Abstract*— About 30% of individuals over the age of 65, and 50% over age 80, fall at least once per year [1]. Fall-related injuries cost the Canadian health care system $2.8 billion annually [2]. Risk for falls in older adults is commonly assessed in the clinical environment using tools such as the Short Physical Performance Battery (SPPB) [3], which include subjective assessments of postural sway while standing under various sensory conditions. This study uses wearable accelerometers and a force plate to quantify measures of postural stability during these tasks. Four participants were asked to maintain quiet stance in six different conditions, while their center of pressure (COP) and accelerations from six accelerometers were recorded. Standard deviations in signals were used as measures of postural sway. The sway observed in all sensors increased with the difficulty of the stance condition. Manipulation of vision and surface stiffness caused greater changes in sway in the AP than ML direction, while changes in stance configuration were more evident in the ML direction. Furthermore, the ankle sensor was the most sensitive in registering changes in sway when manipulating vision and surface stiffness (showing an increase of 236% over baseline values in AP sway with eyes closed and standing on foam), while the thigh was most sensitive to changes in stance width (showing an increase of 336% over baseline values in ML sway in the tandem stance condition). This study contributes in establishing the utility of wearable sensors for quantifying postural stability under various stance configurations in future studies with high-risk older adults.**

I. INTRODUCTION

Falls are a major cause of injury in older adults. An individual's risk for falls associates with their postural stability during daily activities. In the clinical environment, postural stability is often assessed using tools such as the Short Physical Performance Battery (SPPB) [3] the timed Get-Up-and-Go [4], and the Physiological Balance Profile [5]. However, such tools rely on subjective classifications of performance.

In the laboratory environment, postural stability is commonly assessed by measuring the variability in the location of the centre-of-pressure (COP) between the feet and ground from a force plate [6]. Miniature wearable sensors represent a lower-cost alternative to force plates for quantifying postural stability both within and outside of the clinical environment [7]. However, previous studies have not validated sensor-based measures of postural stability through comparison with force plate data.

Accordingly, the primary goal of this study was to compare COP-based measures of postural stability to those acquired by miniature inertial sensors worn at various body sites. A second goal was to examine the effect on these measures of alterations in vision, floor stiffness and stance configuration (manipulations typically incorporated into clinical tests). Our results illustrate how wearable sensors provide information on postural stability that correlates with COP measures, is sensitive to task conditions, and conveys underlying postural control mechanisms. These results support the value of this portable technology as an attractive option to force plates in quantifying postural stability during stance.

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II. METHOD

A. Study participants

Study participants consisted of 4 healthy young individuals (1 male, 3 female), all of whom provided informed consent.

B. Stance Conditions

Participants were instructed to stand as quietly as possible while adopting different stance configurations, with eyes open and closed, and while standing on a firm surface or foam. The conditions were as follows:

1) Normal stance (feet shoulder width apart), eyes open, on rigid surface;

2) Semi-tandem (ST) stance (right foot in front of left, with narrow stance width), eyes open, on rigid surface;

3) Tandem (T) stance (right foot in front of left, with zero stance width), eyes open, on rigid surface;

4) Normal stance, eyes closed, on rigid surface;

5) Normal stance, eyes open, on compliant surface (10 cm thick foam pad);

6) Normal stance, eyes closed, on compliant surface.

7) Normal stance, eyes open, rigid surface, exaggerated anterior-posterior (heel-toe) rocking.

C. Data collection and analysis

In each trial, data were collected from miniature wireless sensors (tri-axial accelerometers ±6g, Opals, APDM Inc.) secured at six body sites (Fig. 1): sternum, waist (posterior aspect), right and left thighs and right and left ankles (lateral aspects). An additional sensor was placed on a mallet drop synchronization system. Sensor data were sampled at 128 Hz (the maximum sampling frequency offered by the sensors).

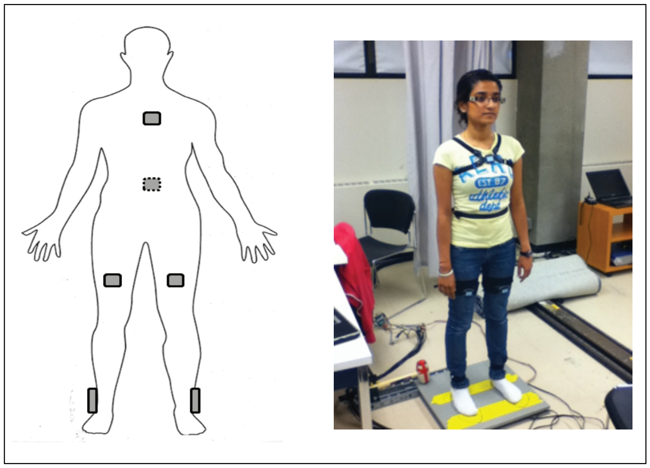

Fig. 1. Location of inertial sensors on the body.

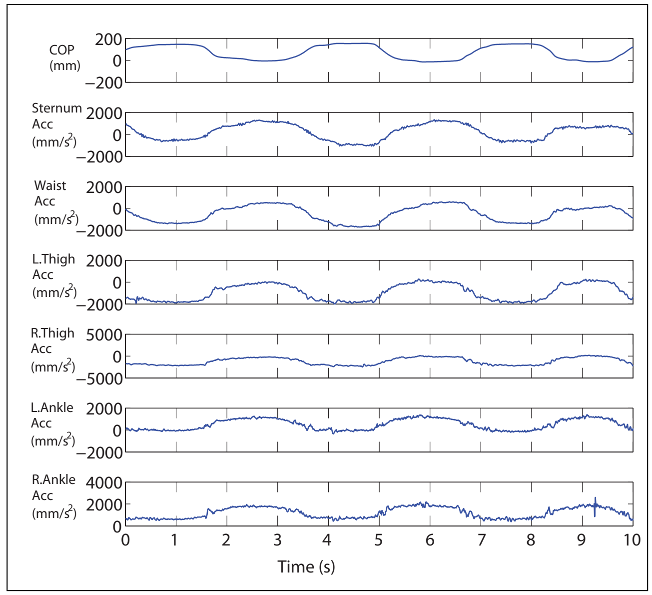

Fig. 2. COP and acceleration traces in the AP direction (7) - exaggerated heel-toe rocking.

We also acquired foot reaction forces and moments from a force plate (Accusway, AMTI) at 1280 Hz via LabVIEW. COP in the x and y directions was calculated from ground reaction forces (Fx, Fy, Fz) and moments (Mx, My, Mz) as follows: COP_x_direction $= \frac{-My}{Fz} \times 1000$ and COP_y_direction $= \frac{Mx}{Fz} \times 1000$.

The APDM data collection software allowed for synchronized measures from each of the 7 sensors. To synchronize these sensor data with data from force plate, a hinged mallet, having a sensor attached to its head, was raised a fixed height and released to strike the force plate, providing a distinct time stamp for synchronization of sensor and force data. Force plate and sensor data were filtered using a fourth-order low-pass Butterworth filter, having a 10Hz cut-off frequency [8]. Fig. 2 illustrates an example of COP and acceleration data acquired under condition (7), exaggerated heel-toe rocking.

In each trial, we characterized postural sway as the standard deviation (SD) in acceleration (from the sensors) or COP (from the force plate) in both the anterior-posterior (AP)

TABLE I
EFFECT OF CONDITION ON POSTURAL SWAY

	Sternum	Waist	Thigh	Ankle	COP
AP Direction					
Normal stance	54.1 (20.6)	43.9 (13.3)	40.6 (11.6)	22.5 (8.1)	2.3 (0.8)
Semi tandem	56.9 (13.4)	50.4 (6.3)	46.8 (5.8)	30.7 (5.7)	2.7 (0.4)
Tandem	73.2 (11.2)	72.7 (19.4)	84.9 (16.0)	78.8 (22.0)	3.5 (0.9)
Eyes closed, rigid surface	75.3 (11.6)	70.7 (21.1)	55.3 (15.7)	41.2 (11.6)	4.2 (0.9)
Eyes open, foam surface	84.3 (26.8)	70.1 (22.2)	65.6 (17.6)	59.4 (19.9)	4.9 (1.5)
Eyes closed, foam surface	91.3 (19.0)	86.1 (21.0)	86.1 (25.6)	74.4 (23.4)	6.6 (1.9)
Heel-toe-rocking	860.3 (234.5)	792.9 (275.4)	616.8 (169.3)	414.0 (21.7)	58.4 (8.5)
ML Direction					
Normal stance	33.5 (10.7)	19.8 (2.6)	23.4 (2.8)	12.7 (3.6)	1.0 (0.2)
Semi tandem	45.7 (9.9)	34.7 (5.3)	32.7 (2.3)	31.1 (10.2)	2.6 (0.3)
Tandem	110.6 (18.6)	68.6 (9.7)	103.3 (22.0)	82.8 (24.8)	5.5 (1.2)
Eyes closed, rigid surface	38.4 (9.7)	21.7 (4.1)	26.5 (6.6)	20.9 (8.6)	1.2 (0.3)
Eyes open, foam surface	42.5 (7.6)	32.3 (8.1)	34.2 (6.5)	32.7 (7.0)	2.3 (0.6)
Eyes closed, foam surface	42.1 (7.4)	34.2 (6.7)	40.8 (8.8)	43.4 (14.6)	2.9 (0.8)
Heel-toe-rocking	122.5 (51.8)	98.2 (37.1)	141.2 (16.2)	129.3 (53.5)	6.9 (2.4)

Notes: Cell entries show mean values of the standard deviations (SD's) in acceleration and COP location, averaged over all subjects, across right and left sides for thigh and ankle sensors. SD's of the SD's are shown in parentheses. Units are mm/s^2 for sensor accelerations and mm for COP position.

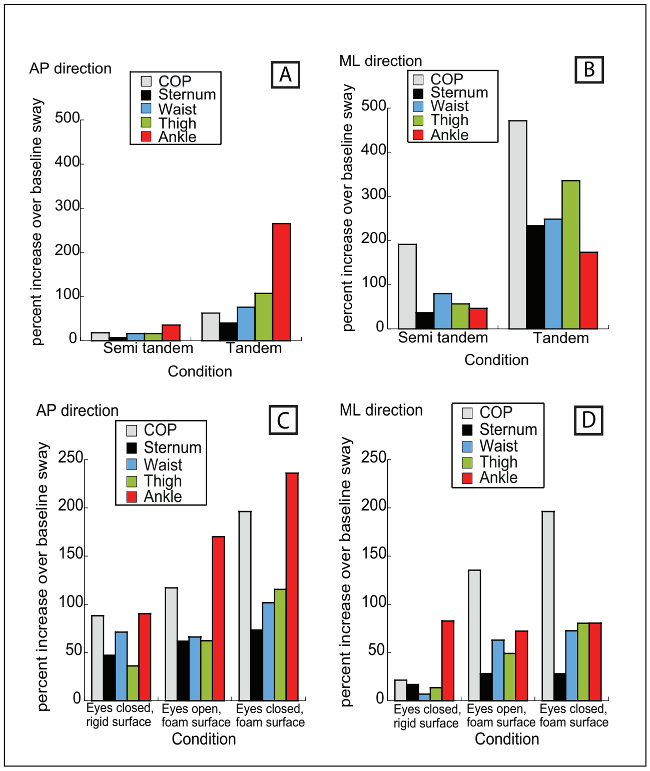

Fig. 3. Percent change in postural sway (signal SD) over baseline (condition (1)) condition following manipulations in stance configurations (A and B) or vision and surface stiffness (C and D).

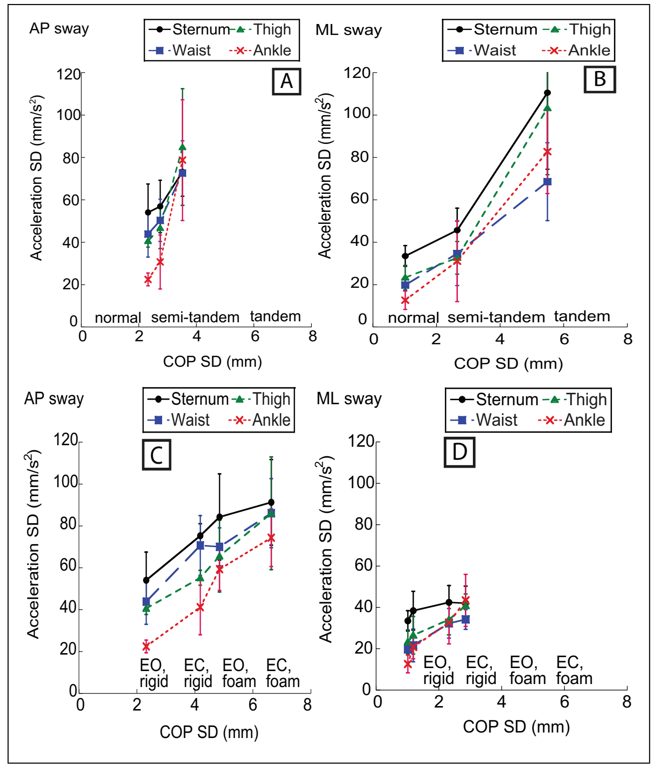

Fig. 4. Combinations of COP and acceleration for various sensor locations following manipulations in stance configurations (A and B) or vision and surface stiffness (C and D).

and medial-lateral (ML) directions, over 10 seconds. We also examined the correlation between sensor accelerations and COP, after down-sampling the latter (from 1280 to 128 Hz) using a shape-preserving piecewise cubic interpolation.

III. RESULTS

As the difficulty of the task increased (from 1 to 6), there was an increase in the SD's of both COP and acceleration signals (Table 1). In the AP direction, the largest SD's occurred in condition 6 (eyes closed, compliant surface), except for the ankle sensor, where the largest SD's occurred in condition 3 (tandem stance). In the ML direction, the largest SD's in COP and acceleration occurred in condition 3 (tandem stance).

Alterations in stance configuration caused larger changes in sway in the ML than AP direction, with the largest changes observed at the thigh and waist sensors (Figs. 3A and 3B). When compared to baseline conditions, tandem stance (condition 3) involved increases in ML sway of 336% at the thigh, and 249% at the waist. Conversely, changes in vision and surface stiffness yielded larger changes in sway in the AP than ML direction (Figs. 3C and 3D), with the largest changes observed at the ankle (increases of 90%, 170% and 236% over baseline values for conditions 4, 5 and 6, respectively).

IV. DISCUSSION

In this study, we examined the utility of wearable sensors in characterizing postural sway under various clinical testing conditions. We found that the variance (standard deviation) from all sensors increased as the base of support decreased, as vision was removed, or when moving from a rigid to compliant ground. We also found that the ankle sensor was most sensitive in registering changes in sway when manipulating vision and surface stiffness, while the waist and thigh sensors were most sensitive to changes in stance width.

Furthermore, manipulation in vision and surface stiffness caused greater changes in sway in the AP than ML direction, while changes in stance configuration had a larger effect on sway in the ML than AP direction. These results guide the design of a minimum sensor array for future clinical use.

They also illustrate the value of wearable sensors in providing insight on the postural control strategies (e.g., hip versus ankle strategy) used under various sensory and support conditions [9,10]. An important limitation of this study is that our participants were young healthy individuals, and an essential next step is to repeat the experiment with older adults.

V. CONCLUSION

We instructed human participants to stand as quietly as possible under various sensory and support conditions, and compared measures of postural sway from miniature accelerometers mounted at various body locations, to those acquired with a force plate. Of all the signals we examined, AP sway at the ankle was most sensitive to alternations in vision and surface stiffness, while ML sway at the waist or thigh was most sensitive to changes in stance width.

While inertial sensors have previously been used to assess postural stability, to our knowledge, there have been no previous studies that compare COP measures to information provided through wearable sensors across a wide range of static task conditions (that not only include normal quiet stance, but also alter the base of support, vision and somatosensory input). Furthermore, we employed a novel approach by using sensors to identify the relationship between various task conditions and direction (AP vs. ML) of greatest instability. Lastly, study findings can guide in the identification of a minimum sensor array system to help understand underlying postural control mechanisms. Overall, the study results contribute to the development of a cost effective wearable sensor system for providing accurate and meaningful measures of postural stability.

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