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Reliability and practical clinical application of an accelerometer-based dual-task gait balance control assessment

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ABSTRACT

Background: Gait balance control assessment using whole body center of mass (COM) kinematic measures in concussed individuals reveals persistent balance deficits up to two months post-injury. A reliable and clinically practical gait balance control assessment leveraging similar kinematic measures is necessary to improve concussion assessment and management.

Research question: Can peak accelerations collected during a dual-task (DT) gait assessment from a single low back placed accelerometer be measured reliably on different days, by different raters, in different environments, and be practically applied in a Division One (D1) athletics program?

Methods: A single accelerometer placed on the low back over the L5 vertebra was utilized with a DT gait analysis protocol. Twenty (10 F) healthy participants performed the assessment in a laboratory and non-laboratory environment, on two separate days, and with two different raters. Eight gait event specific peak accelerations along three orthogonal axes were collected. In addition, data were collected from a cohort of 14 D1 female soccer players during a single assessment to explore the practical clinical application.

Results: Cronbach's α values for the eight metrics ranged from 0.881 to 0.980 and ICC values from 0.868 to 0.987. Average assessment time for the 14 D1 female athletes was 8.50 ± 0.58 min, and significant differences between walking conditions were identified for Vert Accel 1 ($p < .01$), Vert Accel 2 ($p = .01$), and A-P Accel ($p < .01$).

Significance: High Cronbach's α and ICC values coupled with a short assessment time and sensitivity to differences in gait balance control indicate our testing apparatus and protocol are both reliable and clinically practical. Additionally, gait event specific peak accelerations from a single accelerometer can detect subtle changes in gait balance control and may facilitate improvements in sport-related concussion diagnosis and return to activity decision making.

1. Introduction

Assessment of dynamic gait balance control is increasingly important in concussion evaluation for diagnosing injury, evaluating rehabilitation progress, and informing return to activity (RTA) decision making [1–3]. Concussion induced gait balance impairments persisting as long as two months post-injury have been detected in laboratory studies using sensitive biomechanical markers obtained by sophisticated camera-based motion analysis systems [4–7]. This suggests that despite earlier resolution of subjective symptoms, static balance, and neurocognitive function, used in current RTA guidelines, recovery may not be complete prior to RTA.

Many studies reporting prolonged deficits in gait balance control in acutely concussed individuals employ a dual-task (DT) assessment paradigm [8–10]. The paradigm pairs a gait task with a concurrent cognitive task, more closely resembling the demands of athletic, occupational, and daily activities [11,12]. It is particularly sensitive to prolonged gait imbalance as it requires simultaneous allocation of attentional resources to sensory processing for motor planning and cognitive task completion. This challenge to attentional function creates a competition for limited processing resources resulting in a reduction in cognitive and/or motor performance [13]. Often utilized cognitive tasks include the auditory Stroop test [14] and a more complex Question and Answer (Q&A) battery [15,16].

Many whole body center of mass (COM) kinematic measures, including the total medial-lateral (M-L) displacement, peak M-L velocity, and maximum M-L horizontal separation between the COM and center of pressure (COP) [5,10,17], were identified as capable of detecting

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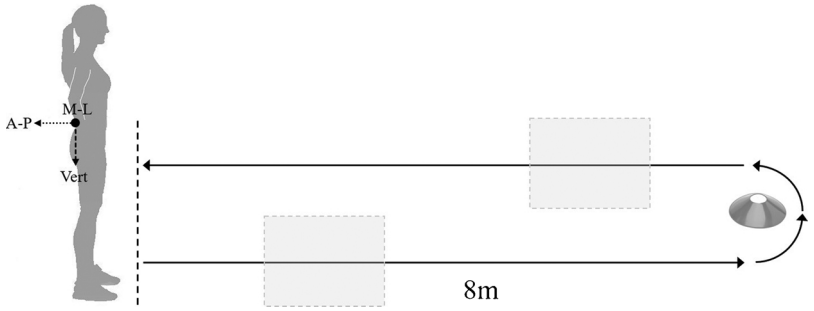

Fig. 1. Placement of the L5 sensor and orientation of the three orthogonal axes. Graphical illustration of the walking task including straight level walking along an 8 m path, a 180° counter clockwise turn around a cone, and a return straight line walk. Shadowed boxes represent the two gait cycles trimmed for analysis.

persistent gait balance control deficits in concussed individuals despite earlier normalization of temporal distance gait parameters. This altered COM position and velocity may be the result of poor momentum control due to deficits in regulating COM acceleration, ultimately compromising an individual's ability to successfully execute the physical demands of daily and sport-related activities. It has been reported that individuals with and without functional limitations demonstrated differences in COM accelerations [18], supporting this conclusion and suggesting measurement of COM acceleration may help illuminate balance control strategies during various activities.

Whole body COM is traditionally calculated as a weighted sum of linked rigid body segments captured with a whole body reflective marker set and camera-based motion analysis equipment. Fortunately, recent advances in wearable motion sensor technologies offer an opportunity to translate previous findings that employed whole body video motion capture into the clinical environment with closely related measures of gait balance control. These inertial measurement units (IMU) combine accelerometers, gyroscopes, and magnetometers into a single sensor. As an emerging technology, the use of wearable sensors to identify gait and balance impairments has grown rapidly. Many studies demonstrated reliability and utility in assessing static balance control [19–21] and estimating gait temporal distance parameters [20,22,23] in both healthy and neurologically impaired (concussion and Parkinson's) individuals; while others used a low back placed sensor to estimate COM kinematics [24,25]. Given the close proximity between the fifth lumbar vertebra (L5) and the whole body COM during level walking, it could serve as a feasible location for IMU placement [26]. In a recent investigation using a single accelerometer placed on L5, the peak M-L acceleration occurring during the transition from single- to double-support was able to differentiate concussed from healthy subjects up to two weeks post injury. This suggests directly measured accelerations from the L5 may have a similar sensitivity to persistent gait imbalance as whole body COM kinematics [7].

While promising, measures of gait balance control using directly measured accelerations from a single IMU over the L5 vertebra are not well established or validated. Furthermore, sensitivity to changes in gait balance control and the consistency outside of the laboratory and across time of these measures is unknown. Without such investigations, the application of an IMU-based gait balance control assessment tool for use in rehabilitation clinics, athletic training rooms, or austere field environments cannot be accomplished.

Therefore, the purposes of this study were to 1) determine the consistency of a DT wearable sensor gait balance assessment protocol in a non-laboratory setting and across time, 2) determine the assessments inter-rater reliability, and 3) demonstrate its ability to be practically administered in a Division One (D1) collegiate sports medicine clinic. It was hypothesized that 1) acceleration-based kinematic metrics collected from a single sensor placed over the L5 vertebra would have high internal consistency in single- and dual-task walking conditions measured on two different testing days, in a laboratory and non-laboratory environment, 2) and by two different raters. It was further hypothesized that 3) the assessment protocol could be practically administered in a real-world Division One (D1) sports medicine clinic in minimal time by sports medicine personnel.

2. Methods

Study participants consisted of healthy non-athlete young adults and D1 female soccer players from the University community. Participants were healthy, able to walk over level ground without an assistive device, and had normal hearing. Individuals with lower extremity deficiencies, an injury affecting normal gait, a history of cognitive deficiencies such as permanent memory loss or concentration abnormalities, attention deficit hyperactivity disorder, three or more previous concussions, or a concussion within the past year were excluded. The study was approved by the Institutional Review Board and all participants provided written informed consent prior to enrollment.

The testing apparatus consisted of a single laptop, a three-sensor OPAL motion analysis system operated by Motion Studio software (APDM, Inc., Portland, OR, USA), a single ear Bluetooth synced wireless headset and boom microphone (Blue Tiger USA, Stafford, TX, USA), and Superlab 5 software (Cedrus Corp., San Pedro, CA, USA). Sensors were attached with elastic straps over both lateral ankles for gait event detection and over the L5 vertebrae on the back [7]. The axes of the L5 sensor were oriented such that the positive vertical (Vert) axis pointed inferiorly, the positive medial-lateral (M-L) axis pointed to the right, and the positive anterior-posterior (A-P) axis pointed posteriorly (Fig. 1). The DT assessment protocol was automated in a custom Superlab 5 program. All verbal commands were administered through the wireless headset.

Participants were instructed to walk at a self-selected comfortable pace from a feet together standing position over an eight meter level path, perform a 180° counter clockwise turn around a cone, and return to a stop at the start position (Fig. 1). Trials were initiated with an automated verbal command instructing participants to look straight ahead, followed by an auditory beep. The two concurrent cognitive tasks utilized during DT walking were the auditory Stroop test and a Q&A task. The auditory Stroop test consisted of four auditory stimuli, the words "high" or "low" spoken in either a high or a low pitch [14]. The stimuli were either congruent (pitch of the voice matched the meaning of the word) or incongruent (pitch of the voice did not match the meaning of the word). Participants were instructed to correctly identify the pitch of the voice rather than the word spoken. Three randomly presented stimuli were triggered manually by the rater to begin on the third heel strike during the walkout, one step prior to the 180° turn, and on the third heel strike during the return walk (Fig. 1). The Q&A consisted of either spelling a five letter word backwards or subtracting from a given number by sixes or sevens. The first question was queued immediately upon walking initiation and participants responded continuously throughout the trial.

This study consisted of data collections from two separate cohorts of participants. The first cohort included 20 male and female healthy,

young-adult, non-athletes (10 females; age 22.2 ± 2.8 yrs.; height 175.8 ± 8.1 cm; weight 71.0 ± 12.0 kg). They were assessed at two data collection sessions separated by one to two weeks, in two different environments, and by two different raters for the purpose of examining the reliability of the assessment. At each session participants completed four trials in a single-task (ST) and two DT walking conditions: ST (walking only), DT Stroop (walking while performing a concurrent auditory Stroop), and DT Q&A (walking while responding to Q&A). The three walking conditions were repeated in two different environments and by two different raters. The testing environments were a quiet laboratory and a hallway in an adjacent building. Occasional visual and auditory distractions occurred in the hallway, similar to those expected in a medical clinic. Raters consisted of the primary author and an undergraduate student research assistant, minimally trained in biomechanical analysis. The order of environment, rater, and walking condition was randomized for each individual at each session. Participants performed four trials of the Stroop and Q&A tasks in a seated position prior to initiating the walking trials. Participants wore normal athletic clothing including a t-shirt, shorts (or tight fitting exercise pants), and athletic shoes.

The second cohort of 14 female athletes from the University D1 female soccer team (age 19.3 ± 1.3 yrs.; height 168.4 ± 7.1 cm; weight 64.6 ± 4.0 kg) was included for the purpose of assessing the practical clinical application of the assessment. Data was collected in this cohort during a separate session and examined independent of the first cohort. Athletes performed a single assessment with three trials in each of the three randomly presented walking conditions. The assessment was performed in a quiet workout recovery room in the university athletic medicine facility. The walls of the room were transparent glass allowing for occasional visual distractions. Participants completed the same walking task, received the same verbal instructions, and wore the same apparel as participants in the first cohort. Prior to completion of the three walking conditions, three trials of the Stroop and Q&A tasks were performed in a seated position, followed by two familiarization ST walking trials. A certified athletic trainer minimally trained in biomechanical analysis performed all athlete assessments. Total assessment time, including sensor placement was also recorded.

IMU sensor data was sampled at 128 Hz and streamed in real-time to a wireless hub. Raw data was filtered with a 2nd order low-pass Butterworth filter with a 12 Hz cutoff frequency and analyzed with a custom Labview program (National Instruments, Austin, TX, USA). Ankle pitch angular velocity collected from the two ankle IMUs was used to identify heel strikes for trimming individual gait cycles. In this study, data from straight line walking were examined. Two gait cycles were analyzed in each trial; the gait cycles beginning on the fifth heel strike during the walk out and walk back. Acceleration measures from the L5 sensor were referenced to a single left heel strike initiation gait cycle. A tilt correction was performed to obtain acceleration values with respect to the anatomic frame of reference. Eight peak accelerations corresponding to specific gait events along the M-L, A-P, and Vertical axes were identified and obtained for analysis (Fig. 2). Gait events included right terminal swing (M-L acceleration 1), left terminal stance (A-P acceleration), right heel strike (Vertical acceleration 1, M-L acceleration 2, and A-P deceleration), right loading response (M-L acceleration 3 and Vertical acceleration 2), and right mid-stance (Vertical acceleration 3).

An eight item Cronbach's α was calculated on data collected from the first cohort for each of the eight peak accelerations in each of the three walking conditions to establish the internal consistency of the assessment across two testing sessions, two environments, and two raters. An Intra-class Correlation Coefficient (ICC) with a 95% Confidence Interval (CI) was also calculated to determine inter-rater reliability. Cronbach's α values greater than .9 and ICC values greater than .75 were considered to indicate very high internal consistency and excellent interrater reliability.

A one-way, repeated-measures, Analysis of Variance (ANOVA) with an alpha level of .05 was performed on peak acceleration data collected from the second cohort (D1 athletes) to determine if there were differences in peak acceleration values between the three walking conditions. Total assessment time was also recorded. Data from the first and second cohorts were not compared to one another. All analyses were performed on Statistical Product and Service Solutions (SPSS version 24) software.

3. Results

The two testing sessions from cohort one were separated by 8.7 ± 1.7 d. Cronbach's α values for all eight metrics in each of the three walking conditions had a range of .839–.989 (Table 1) indicating high to very high internal consistency. Inter-rater reliability was excellent as demonstrated by ICC values of .935–.989 (Table 1).

Total assessment time for the second cohort of female athletes including sensor placement and verbal instructions was 8.50 ± 0.58 min. Significant differences between the three walking conditions for Vert Accel 1 ($p < 0.01$, $\eta_p^2 = 0.49$), Vert Accel 2 ($p = 0.01$, $\eta_p^2 = 0.30$), and A-P Accel ($p < 0.01$, $\eta_p^2 = 0.49$) were identified with the repeated-measures ANOVA (Fig. 3). While not reaching a significant level, Vert Accel 3, M-L Accel 2, and M-L Accel 3, demonstrated potential to differentiate between walking conditions with p-values of 0.12, 0.14, and 0.06, respectively.

4. Discussion

This study established the reliability of a DT gait balance control assessment utilizing multiple peak accelerations recorded from a single IMU placed over the L5 vertebra. Our high Cronbach's α values, a statistic utilized for its ability to measure both the correlation and agreement among multiple measures, demonstrate the reproducibility of the kinematic measures in a non-laboratory environment. They further establish the temporal consistency of the outcome measures in repeated assessments separated by one to two weeks inferring a minimal learning effect. High ICC values further suggest the test may be reliably performed by different, minimally trained raters.

We also demonstrated the ability to perform the assessment in a D1 collegiate sports medicine clinical environment. Our average assessment time of 8.50 min indicate it may be performed within a reasonable time with minimal impact to athlete training schedules. The assessment was also conducted by a team athletic trainer, untrained in biomechanical analysis, in a pre-existing space in the athletic medicine facility. The length, ease of administration, and ability to perform the assessment in preexisting clinical space, suggest it may be practically applied in this setting. Furthermore, analysis of athlete data revealed multiple peak accelerations capable of differentiating between different walking conditions, indicating their sensitivity to subtle changes in gait balance control.

Traditionally, whole body COM is computed from a full body reflective marker set captured with sophisticated camera-based motion analysis systems. A previous studies using this method identified whole body COM kinematic variables (total M-L displacement and peak M-L velocity) capable of distinguishing healthy and concussed individuals across a two month post injury period with large effect sizes [5]. Recent advancements in wearable IMU technology offer an opportunity to translate these findings into a clinical biomechanical assessment by using a single IMU placed over the low back. A recent pilot investigation using a similar single IMU method identified an acceleration metric (peak M-L acceleration during the transition from single- to double-support) also capable of differentiating concussed from healthy individuals with a similarly large effect size for up to two weeks post injury [7]. Low power likely contributed to the reduced temporal sensitivity, however, the comparison of camera and IMU measures suggest the IMU-based technique may be a viable alternative for application of these measures in clinical environments.

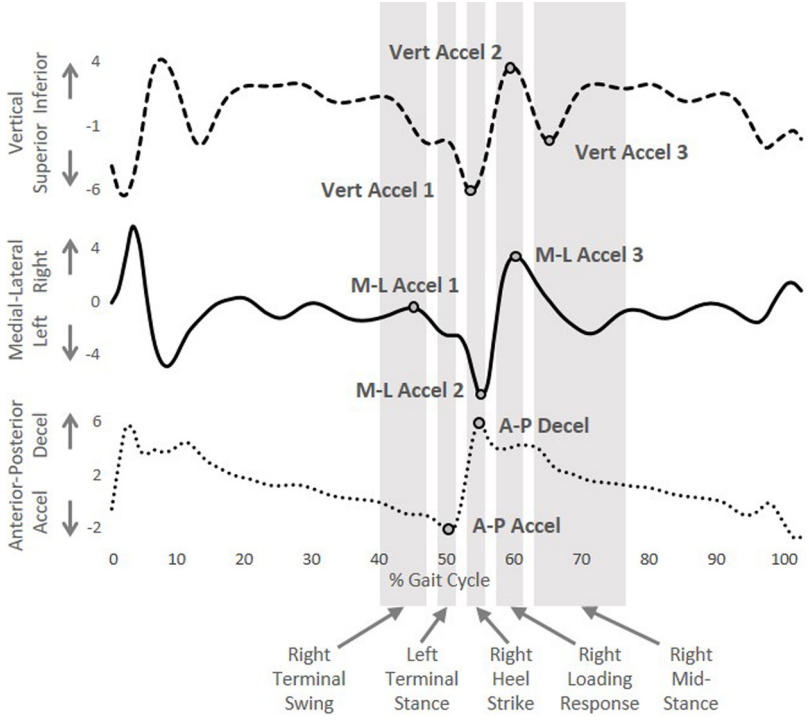

Fig. 2. Representative acceleration profiles for vertical, medial-lateral, and anterior-posterior accelerations (m/s^2) and selected acceleration peaks. Gait events corresponding to peak accelerations are listed below and highlighted with gray vertical bars.

Table 1
Cronbach's α values for each of the eight acceleration metrics for all three walking conditions calculated. The eight items of the Cronbach's α consisted of all combinations of the levels of the three independent variables, environment, time, and rater. Inter-rater reliability determined by an Intraclass Correlation Coefficient and 95% Confidence Interval calculated for each of the eight metrics.

	Cronbach's α			ICC	95% CI
	ST	Stroop	Q&A		
Vert Accel 1	.967	.975	.974	.989	.972–.996
Vert Accel 2	.954	.975	.966	.988	.970–.995
Vert Accel 3	.946	.959	.944	.983	.957–.993
M-L Accel 1	.932	.942	.965	.964	.913–.986
M-L Accel 2	.955	.961	.961	.975	.937–.990
M-L Accel 3	.900	.910	.948	.935	.844–.974
A-P Decel	.940	.932	.929	.987	.968–.995
A-P Accel	.859	.839	.871	.966	.917–.987

In this study eight gait event specific peak accelerations along three orthogonal axes were investigated, the most consistent of which occurred at (in order of occurrence) terminal swing, terminal stance, heel strike, the loading response during the transition into single-support, and mid-stance. During the propulsive phase at terminal stance the COM is propelled forward and medially away from the base of support. A change in acceleration magnitude in the A-P or M-L directions may indicate adoption of a conservative gait strategy or excessive force output. Conversely, at heel strike the COM is moving toward the heel striking foot, requiring the neuromuscular system to attenuate the body's momentum. This continues into the early phase of single support where the entire momentum of the body is loaded onto the single limb and must be controlled or risk breaching the base of support. Intuitively, changes in acceleration magnitudes during these gait events may denote altered gait balance control capability.

This study was not without limitations. The generalizability of our testing protocol may be affected by our relatively homogenous samples. Possible confounding variables present in athletic populations such as motivation (shamming), peer pressure, and other psychosocial variables may affect these results as well as the high level of dynamic balance control of elite athletes compared to non-athletes. There is also a high degree of between subjects variability in peak acceleration values, however, it is similar in magnitude to whole body COM position and velocity data obtained from video motion capture. Additionally, all facilities may not have the required space to perform such assessments nor be capable of accommodating the additional time required to add it to their existing assessment protocol. Finally, the small sample size of the D1 athlete group limited the ability to identify peak accelerations capable of detecting changes in balance control as evidenced by an observed power between .40 and .55 for the three metrics demonstrating non-significant trends.

5. Conclusion

Our results indicate analysis of accelerations collected with a wearable IMU over L5 under a DT testing paradigm, is a reliable measure of gait balance control during normal gait outside of the laboratory, is consistent over different testing days, and may be conducted by minimally trained individuals. Furthermore, it is able to detect subtle differences in DT gait balance control and can be performed in a reasonably short time by sports medicine staff in pre-existing clinical facilities, supporting its practical application in a D1 sports medicine environment. It is therefore reasonable to expect our clinical assessment may be successfully applied to acutely concussed individuals to accurately and rapidly measure deficits in dynamic gait balance control. It may further be capable of detecting improvements in balance control over the post-injury recovery period resulting in a better informed RTA decision.

Conflicts of interest

The authors declare no conflicts or financial interests with any organization or entity in the subject area covered in this manuscript.

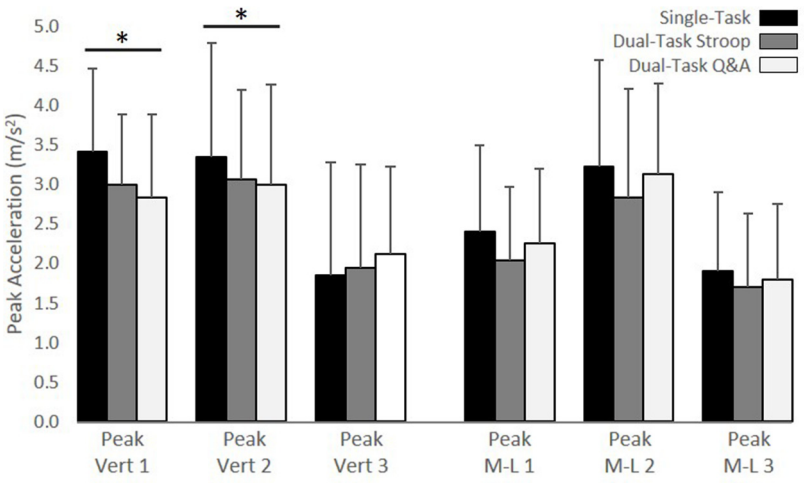

Fig. 3. Vertical and medial-lateral axis peak acceleration metrics in each of the three walking conditions for female D1 soccer players presented as mean ± SD. * indicates $p \leq 0.01$.

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