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Quantifying performance and effects of load carriage during a challenging balancing task using an array of wireless inertial sensors

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ABSTRACT

We utilize an array of wireless inertial measurement units (IMUs) to measure the movements of subjects ($n = 30$) traversing an outdoor balance beam (zigzag and sloping) as quickly as possible both with and without load (20.5 kg). Our objectives are: (1) to use IMU array data to calculate metrics that quantify performance (speed and stability) and (2) to investigate the effects of load on performance. We hypothesize that added load significantly decreases subject speed yet results in increased stability of subject movements. We propose and evaluate five performance metrics: (1) time to cross beam (less time = more speed), (2) percentage of total time spent in double support (more double support time = more stable), (3) stride duration (longer stride duration = more stable), (4) ratio of sacrum M–L to A–P acceleration (lower ratio = less lateral balance corrections = more stable), and (5) M–L torso range of motion (smaller range of motion = less balance corrections = more stable). We find that the total time to cross the beam increases with load ($t = 4.85$, $p < 0.001$). Stability metrics also change significantly with load, all indicating increased stability. In particular, double support time increases ($t = 6.04$, $p < 0.001$), stride duration increases ($t = 3.436$, $p = 0.002$), the ratio of sacrum acceleration RMS decreases ($t = -5.56$, $p < 0.001$), and the M–L torso lean range of motion decreases ($t = -2.82$, $p = 0.009$). Overall, the IMU array successfully measures subject movement and gait parameters that reveal the trade-off between speed and stability in this highly dynamic balance task.

1. Introduction

Understanding how human movement is affected by load is critical for designing equipment that warfighters and emergency responders carry during missions or rescues. Of critical importance is the warfighter's or emergency responder's ability to maintain balance during a wide range of tasks, which include but are not limited to standing, walking, and running. Maintaining balance requires maintaining a specific position and velocity relationship between the center of mass (COM) and the base of support [1,2]. Balance can be controlled in simple standing tasks via ankle torque [1], whereas challenging balance tasks require the use of the entire body [3]. During walking and running, balance and lateral stability are preserved through lateral step placement and arm swinging [4–6].

Human balance is commonly evaluated in a laboratory using a force plate to examine the center of pressure (COP) sway/trajectory [7]. COP motion is correlated to balance performance and degradation [7–10]. Dynamic stability tests that utilize a force plate quantify balance performance in dynamic balance tasks [11]. However, dynamic tests are specific to the test scenario and may not predict performance in other tasks [12]. Balance during walking and running can be evaluated via step parameters; step width provides insight into lateral stability with higher step width variability correlated with less-stable gait [13].

One approach for evaluating balance outside of the laboratory is through functional balance tests [14,15]. Functional balance tests typically utilize balance beams that subjects are instructed to traverse as quickly as possible. When traversing a beam, step width is constrained and therefore subjects must employ other balance

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strategies [14]. Typically, the total time to cross the beam and the number of steps off the beam constitute measures of balance performance. Despite their simplicity, functional balance tests can provide a means to quantify how balance performance is altered by footwear [14], age [16], and added load and equipment [17], and are also useful as a method to predict the work ability of firefighters [18].

Similar to balance tests for firefighters [18], balance tests for warfighters are embedded in obstacle courses that incorporate numerous other functional tests [19]. One such obstacle course is the Marine Corps Load Effects Assessment Program (MC-LEAP), which contains a zigzag and sloped balance beam to assess balance performance and the attendant effects of load and equipment. However, as in the studies above, the sole balance performance metric used on this obstacle is the time to traverse the beam. While traverse time is certainly indicative of performance, it does not reveal the underlying biomechanical movements that govern performance. A major goal of our study is to measure those movements and to probe how they are altered by load. We do so by exploiting wireless inertial measurement unit (IMU) technology that enables unobtrusive measurements of body segment kinematics in real-world environments [20–22]. While IMUs have been employed to measure balance during postural tasks and walking [20,22–26], they have not been utilized to study balance during functional balance beam tasks.

The objectives of this study are: (1) to use kinematic data recorded by a body-worn IMU array to calculate metrics that quantify performance (speed and stability) and (2) to investigate the effects of load on performance. We hypothesize that adding mass to subjects performing a maximal speed balance-challenging mobility task will result in poorer performance (decrease subject speed) yet increase the cautiousness of subject movements. To this end, we propose five metrics of balance beam performance, namely: (1) time to traverse the beam, (2) percentage of total time spent in double support, (3) stride duration, (4) ratio of sacrum medial–lateral (M–L) to anterior–posterior (A–P) root mean squared (RMS) acceleration, and (5) M–L torso range of motion. Shorter traverse times indicate higher speeds, and we therefore expect that added load will increase traverse time (i.e., added load will decrease speed performance). For walking, double support time increases with decreasing speed [27] and increases with load carriage [28]. We therefore expect that for balance beam walking, added load will similarly increase time spent in double support. Increases in stride duration are correlated to slower walking speeds as well as increased double support [27]. Therefore, we expect that added load will result in an increase in average stride duration. The ratio of RMS M–L acceleration to RMS A–P acceleration measures the magnitude of lateral balance corrections relative to fore-aft accelerations (due to push-offs and foot-strikes). Elderly walkers walk with reduced M–L accelerations relative to healthy young walkers [29], likely because they walk with greater caution. Therefore, we expect that added load will decrease the ratio of M–L to A–P acceleration at the sacrum. The range of M–L torso motion quantifies the maximum lean angles exhibited by a subject crossing the beam. We expect that subjects will be more cautious when carrying a load and therefore will exhibit less M–L torso range of motion.

2. Methods

We tested 30 subjects (11 females, 19 males; age = 20.8 ± 2.6 years; body mass = 75.4 ± 11.1 kg; height = 1.75 ± 0.08 m; mean ± standard deviation). The University of Michigan IRB approved the study, and all subjects gave informed consent. Subjects completed an outdoor obstacle course that included a balance beam obstacle (Fig. 1) both with and without a 20.5 kg load (standard mass of body armor and tactical assault panel that all warfighters must carry in a combat situation, regardless of body mass; Fig. 2); the order of trials (load or no load) was random. Subjects also carried a mock-rifle (Fig. 2) in both the load and no-load conditions; the mock-rifle was designed to replicate the weight (3.4 kg) and length of an M4 rifle. The balance beam obstacle (Fig. 1) is composed of five elevated aluminum planks (0.15 m wide, 3.05 m long). The first plank is level, whereas the other planks alternately slope up or down by 9 degrees. The junction between the first and second planks is straight, whereas the other junctions are alternating right and left 90 degree turns. The design of the balance beam obstacle incorporates the major features of the balance beam obstacle in the aforementioned MC-LEAP (Marine Corps Load Effects Assessment Program), an obstacle course designed to assess the effects of load on warfighter mobility. Subjects are instructed to traverse the beam as quickly as possible, while remaining safe and in control. Prior to testing, subjects are permitted as much time as they need to familiarize themselves with the balance beam obstacle, both with and without load. Most subjects only made a single practice run across the beam in each condition, with some subjects making two practice runs in the loaded condition. Subjects that step off the beam return to the start and repeat the obstacle until successful (i.e., no steps off the beam). In our testing, the maximum times that any subject repeated the obstacle was once. Only 8 out of 60 trials required an additional attempt (6 unloaded trials, 2 loaded trials). We used the first successful attempt in each trial for analysis.

We secure eight wireless IMUs (Opal, APDM, Inc.) to each subject's feet, shanks, thighs, sacrum, and torso (Fig. 2). IMUs are secured with elastic straps and reinforced with self-adhering elastic athletic tape. Each IMU (128 Hz sample rate) contains a 3-axis accelerometer (±6g range), angular rate gyro (±2000 deg/s

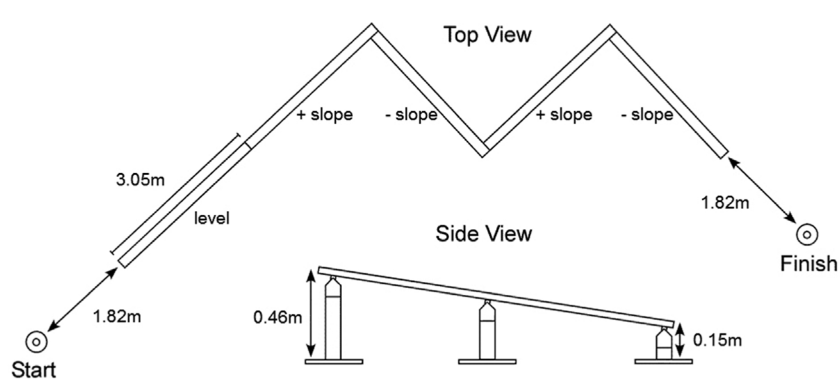

Fig. 1. Schematic and dimensions of the balance beam obstacle in top and side views.

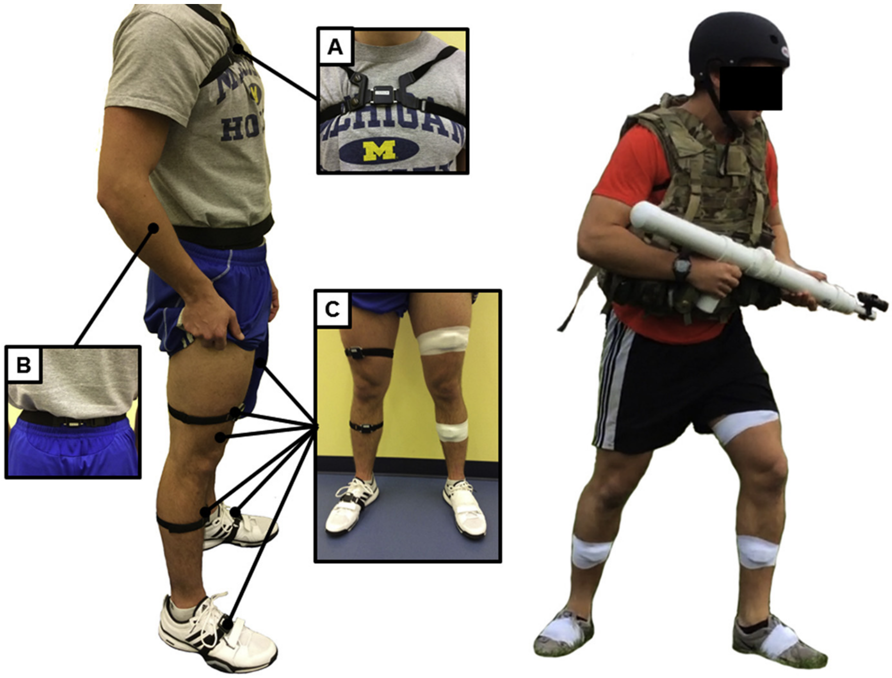

Fig. 2. Placement of the IMUs on a subject (left) and subject with body armor, tactical assault panel, and mock-rifle (right). Boxes illustrate the locations of the torso IMU (box A), the sacrum IMU (box B), and the lower extremity IMUs (box C). IMUs are held in place using a combination of elastic straps and athletic tape, as depicted in box C.

range), and magnetometer ($\pm$6 Gauss). In addition to the raw measurements of acceleration, angular velocity, and magnetic field, a proprietary Kalman filter (APDM, Inc.) estimates the orientation of the IMU relative to an inertial frame defined by the directions of gravity and magnetic north.

Subjects perform three specific calibration movements prior to testing: (1) quiet standing (5 s), (2) straight walking (four strides), and (3) toe-touches (four). Data from these movements allows us to resolve IMU measurements into anatomically relevant frames of reference. Measurements from each IMU are relative to a sensor-fixed frame that rotates with the IMU. For each IMU, we also introduce: (1) a segment-fixed frame with axes aligned with anatomical directions, (2) a tilt-corrected frame (for sacrum and torso mounted IMUs only) that has one axis aligned with gravity (vertical) and two axes defined by M–L and A–P projections onto the horizontal plane, and (3) an inertial frame, which is fixed relative to gravity and magnetic north. The construction of these reference frames from IMU data is detailed in Supplement 1.

We use measurements in each of the frames (sensor, segment, inertial, and tilt) to compute five proposed metrics of balance performance. These five performance metrics include: (1) time to traverse the beam, (2) percentage of the time to traverse the beam spent in double support, (3) average stride duration, (4) ratio of root-mean squared (RMS) M–L acceleration to RMS A–P acceleration, and (5) range of motion of torso M–L flexion.

The direction cosine matrix $R_{\text{interial}|\text{sensor}}$, which describes the transformation from the sensor-fixed frame to the inertial frame (Supplement 1), is employed to obtain the vertical component of the sacrum acceleration. That result is then integrated to obtain the vertical velocity of the sacrum using the initial condition that the sacrum starts from rest at the start of the balance beam. The resulting velocity estimate is polluted by drift error which is then corrected by requiring that the sacrum return to rest at the end of the balance beam. The velocity drift error is modeled as a linear function of time from the start to the end of the balance beam.

The time difference between the step off the beam and the step onto the beam defines the total time to traverse the balance beam obstacle. The maximum and minimum vertical sacral velocities readily reveal when the subject steps onto and off the elevated balance beam obstacle, respectively; see Fig. 3. Note that the traverse time is automatically extracted from this data—a stopwatch, timing gates, or other forms of timing equipment are not required.

The percentage of time to traverse the beam spent in double support and average stride duration are calculated after identifying the foot-strikes and push-offs for each foot. Foot-strikes and push-offs are identified using a wavelet analysis, which reveals when the acceleration measured by a foot-mounted IMUs contains a threshold level of high frequency content. Given the times of the foot strikes and push-offs for each foot, we calculate swing, single support, and double support phases of gait as well as stride durations. We consider all gait events beginning with the last push-off prior to the step onto the beam and ending with the first foot strike after the step off of the beam.

Corrections for balance may manifest as rapid lateral movements of the body in order to reposition the center of mass above the base of support. To quantify this effect, we consider the acceleration of the sacrum in the medial–lateral (M–L) direction as well as the anterior–posterior (A–P) direction. We define the sacrum M–L acceleration as the acceleration component along the $\vec{X}_{\text{tilt}}$ axis (horizontal projection of the pelvis M–L axis) and the A–P acceleration as the acceleration component along the $\vec{Y}_{\text{tilt}}$ axis (horizontal projection of the pelvis A–P axis). The sacrum accelerations along both axes are resolved along these axes using the direction cosine matrix $R_{\text{tilt}|\text{sensor}}$, which describes the transformation from the sensor-fixed frame to the tilt-corrected frame (Supplement 1). The ratio of the root-mean-square (RMS) M–L sacrum acceleration to that of the A–P acceleration component measures the magnitude of the M–L balance corrections relative to the A–P acceleration needed to negotiate the balance beam at high speed.

Lateral movement of the body may also manifest in medial–lateral lean of the torso. The M–L torso lean angle ($\theta_{\text{torso ML}}$) is computed from the components of $\vec{Z}_{\text{segment}}$ along the $\vec{X}_{\text{tilt}}$ (M–L) and $\vec{Z}_{\text{tilt}}$ (vertical) directions (Supplement 1).

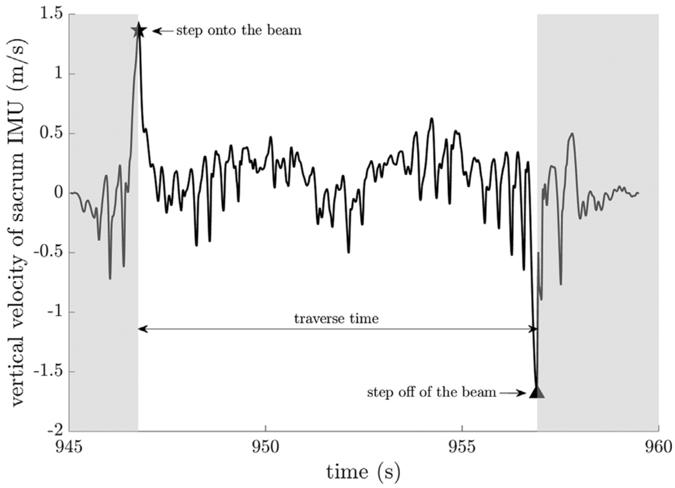

Fig. 3. Drift-corrected vertical velocity of the sacrum-mounted IMU for a sample trial. The black star indicates the maximum vertical velocity, which corresponds to the subject stepping up onto the balance beam obstacle. The black triangle indicates the minimum vertical velocity, which corresponds to the subject stepping off of the balance beam obstacle. The traverse time is the time interval between these extrema.

We use paired *t*-tests with a Bonferroni correction ($\alpha = 0.05/5 = 0.01$) to test for significant differences in our five performance metrics between loaded and unloaded conditions.

3. Results

Table 1 reports the average difference in all five balance beam performance metrics between the loaded and unloaded conditions across all subjects; all data are available in Supplement 2. The speed decreases significantly with load, as quantified by an increase in the traverse time (mean difference = 1.28 s, $t = 4.85$, $p < 0.001$). The four "stability metrics" also exhibit statistically significant load effects, suggesting increased stability in the presence of additional load. In particular, the double support time increases (mean difference = 4.64%, $t = 6.04$, $p < 0.001$), the average stride duration increases with load (mean difference = 0.04 s, $t = 3.436$, $p = 0.002$), the ratio of sacrum RMS M–L to A–P acceleration decreases (mean difference = −0.19, $t = -5.56$, $p < 0.001$), and the M–L torso range of motion decreases (mean difference = −5.41 degrees, $t = -2.82$, $p = 0.009$). We analyzed the effects of possible confounders by checking for significant correlations of subject weight, height, and body mass index (BMI) with traverse time (in both loaded and unloaded conditions) and difference in the traverse time between loaded and unloaded conditions; no significant correlations were found.

Table 1
Summary of the differences between loaded and unloaded conditions.

Metric	Difference, loaded–unloaded (mean ± standard deviation)	t	p
Time to traverse beam (s)	1.276 ± 1.442	4.847	<0.001
Percentage of total time in double support (%)	4.64 ± 2.33	6.044	<0.001
Average stride duration (s)	0.040 ± 0.064	3.436	0.002
Sacrum M–L/A–P acceleration RMS ratio	−0.190 ± 0.187	−5.564	<0.001
Range of motion of torso M–L lean angle (deg)	−5.42 ± 10.53	−2.819	0.009

4. Discussion

Data from the body-worn array of IMUs reveals the underlying biomechanical motions that determine performance on the balance beam task well beyond a simple timing measure. The kinematic data for the major body segments are employed to quantify gait and balance parameters that collectively determine performance and how that performance is altered by load carriage. Results demonstrate that, under load, subjects choose a more conservative gait pattern, as evidenced by the increase in double support, increased stride duration (decreased stride frequency), and decreased gait speed (increased traverse time). Our results are consistent with other studies that demonstrate that load carriage decreases the speed of subjects during balance tasks [17] and other obstacle course tasks [30].

The more conservative gait strategy accompanies significant reductions in balance corrections. In particular, results under load demonstrate significant reduction in the lateral acceleration of the sacrum (relative to the fore-aft acceleration of the sacrum). In concert, the range of motion of the M–L torso lean angle also decreases, suggesting that subjects are less willing to rely on large lateral torso movements for balance correction when carrying a load. The more conservative gait strategy suggests that balance control is worse with load; subjects are less willing to rely on large balance corrections and prefer a larger movement safety margin.

In general, more conservative (stable) gait strategies come at the cost of decreased speeds. Fig. 4 illustrates the overall trade-off between speed (traverse time) and stability (percentage of total time spent in double support). A linear fit to the data ($R^2 = 0.42$, Fig. 4) reveals a positive slope (0.239) significantly different from zero, indicating that spending more time in double support results in slower speeds. However, an example comparison of the two subject data points A and B (Fig. 4) reveals that some subjects (B) are able to maximize speed while simultaneously maintaining high stability, illustrating that the trade-off is not absolute for all subjects.

The dynamic balance test used in this study (or a similar balance test, such as that in MC-LEAP) will be used as part of a larger obstacle course to assess the effects of different loads, load configurations, and equipment on warfighter performance. The potential performance metrics that we identified above will be useful for understanding how load and equipment choices affect the underlying biomechanical movements that govern performance.

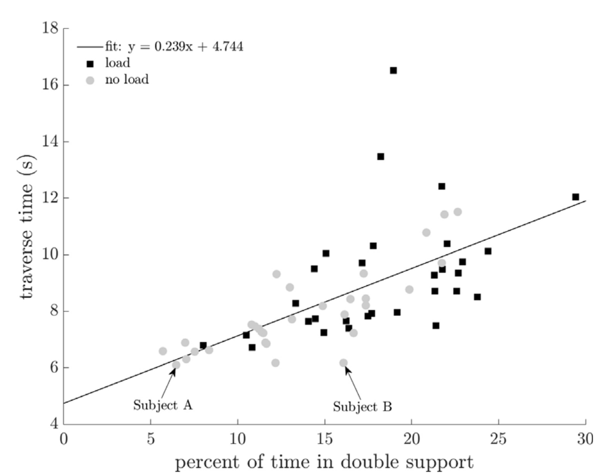

Fig. 4. Traverse time versus double support time. Subjects A and B traverse the beam in a similar amount of time (6.11 versus 6.16 s). However, subject B spends much more time in double support (16.1%) than subject A (6.5%), indicating that subject B maximizes both speed and stability.

5. Conclusion

Data from an array of IMUs enables quantification of performance on a challenging balance beam task that delves deeper than the simple timing measure used in prior studies. The kinematic data for the major body segments reveals corrections to balance and changes in gait that underlie the observed performance changes with added load, which is quantified using: (1) time to traverse the beam, (2) percentage of total time spent in double support, (3) stride duration, (4) ratio of sacrum medial–lateral (M–L) to anterior–posterior (A–P) root mean squared (RMS) acceleration, and (5) M–L torso range of motion. Load carriage has a significant effect on all five performance metrics. Results reveal that, under load, subjects on average decrease their speed while increasing their stability (increasing caution). Future work will explore additional performance metrics (e.g., foot trajectories) and quantify differences in technique, for example, between subjects that are pre-stratified as high and low performing or fit and unfit.

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Conflict of interest statement

The authors do not have any financial and personal relationships with other people or organizations that could inappropriately influence (bias) their work.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2015.10.022.

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