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Technical note

Cross-sectional validation of inertial measurement units for estimating trunk flexion kinematics during treadmill disturbances

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ABSTRACT

Postural perturbation training has been shown to reduce falls. The key outcome measures are trunk flexion kinematics, at recovery step. Motion capture is typically used, but requires space and trained staff. Small, inexpensive, portable inertial measurement units (IMU) are preferred for routine clinical care. IMUs have been validated for trunk motion during walking and running on a treadmill, however treadmill fall prevention training generates higher accelerations. The purpose of this study was to validate the IMU estimate of trunk kinematics against motion capture during treadmill disturbances. Ten healthy young adults had an IMU with a retro-reflective marker triad placed on their sternum to estimate trunk kinematics. Disturbances, increasing in magnitude, were delivered until the harness supported at least 50% of the subject's weight. Equivalence testing ($\alpha = 0.05$) demonstrated the trunk angle (TA) and angular velocity (TAV) measured by the IMU and motion capture were equivalent. The 95% Confidence Intervals (TA: [-1, 1], TAV: [0, 17]) were within the equivalence interval (TA: [−2, 2], TAV: [−20, 20]) and the p-Values (TA: 0.005, TAV: 0.011) were less than alpha. This data confirms that IMUs provide a valid method for measuring trunk kinematics during treadmill perturbation training.

1. Introduction

Select populations have been shown to be at risk for falls. Highly vulnerable groups include older adults and individuals with lower limb amputation. Research has shown that over 50% of older aged women and lower limb amputees have fallen in the community [1–3] and in this group, over 25% self-reported an injury due to the fall [2,3]. These fall related injuries account for more than $50 billion in annual costs [2,4]. The Centers for Disease Control and Prevention has established an initiative to make fall prevention a routine part of clinical care. Falls in older aged women and individuals with lower extremity transtibial amputations have been reduced with task-specific fall prevention rehabilitation training programs [5,6]. Key outcome measures for fall prevention training and predictors for the probability of a fall are trunk flexion angle and velocity at recovery step [7,8]. Traditionally, motion capture systems have been used to estimate joint kinematics, such as trunk angles and velocities, in a controlled laboratory setting. However, this equipment is expensive, requires a large, dedicated space and is operated by dedicated staff; all of which aren't practical for a routine clinical setting. Inertial measurement units (IMUs) provide a better solution. They are commercially available, inexpensive, small, portable and easily attached to key body segments by tape or elastic straps.

IMUs contain a gyroscope, accelerometer and magnetometer. Many commercially available IMUs have proprietary software that implement a Kalman filter to fuse the output from the three sensors and generate an orientation matrix which can be used to estimate joint segment kinematics. IMU accuracy has been validated for trunk use in normal laboratory settings [9–12] and for estimating trunk kinematics during walking and running on a treadmill using custom algorithms for orientation estimates [13,14]. However, treadmill fall prevention training induces large, sudden changes in acceleration which are more energetic than simply waking or running on a treadmill and trunk flexion kinematics are key outcome variables. The accelerometer preforms poorly when there are large and abrupt changes in accelerations and it has been recommended that separate algorithms be used for low and high accelerations [15]. The custom algorithms previously developed for trunk kinematics during treadmill walking and running are not appropriate for treadmill fall prevention training programs, as one routine fuses only accelerometer and gyroscope data [14] and the other recommends output not provided by commercially available software [13]. For routine clinical applications, it is preferred to utilize commercially available filtering options rather than developing custom, situation specific post-processing routines. Mazza et al.

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specifically notes that further studies should be implemented for other motor tasks [14]. Therefore, it is necessary to verify the commercially available IMU orientation estimate during the rapid accelerations and decelerations implemented on the treadmill during a fall prevention training protocol. The purpose of this study was to validate IMU estimate of trunk flexion angles and velocities at recovery step using commercially available filtering routines against the gold-standard motion capture system during treadmill induced forward and backward disturbances.

2. Methods

Ten healthy young adults (5 female, age: 27.1 ± 4.2 years, BMI: 23.8 ± 2.7) participated in this study. This number is consistent with ranges established by previous upper extremity IMU and Motion Capture validation studies, where the sample sizes ranged from five to twenty-two (median: 13.5) [9–14]. Exclusion criteria included chronic or current musculoskeletal pain. Participants were also excluded if they had medical conditions or previous injuries/trauma/surgeries that negatively affected their balance, mobility, or strength or precluded safe participation in the protocol at the investigator or study staff's discretion, used an assistive device such as a cane or walker, or were knowingly pregnant. Twelve subjects were recruited, but two potential subjects had lower extremity pain, which precluded their participation. Informed, written consent was obtained from each subject prior to data collection. Each data collection occurred in a single session and lasted less than two hours.

A commercial IMU (OPAL, APDM, INC., Portland, OR) was secured to the sternum of each subject. Custom code, using static standing and forward bends for each subject, was implemented to define the subject-specific IMU trunk coordinate system [16] and estimate trunk flexion angles and velocities at the recovery step from the IMU orientation matrix (Motion Studio, APDM, INC., Portland, OR). The IMU orientation matrix was generated using all three sensors or using just the accelerometer and gyroscope (Motion Studio, APDM, INC., Portland, OR). A 10-camera (Raptor-12, Motion Analysis Corporation, Santa Rosa, CA) motion analysis system was utilized to acquire kinematic parameters. Retro-reflective, 3D markers were secured to the right and left acromion processes and the right and left anterior superior iliac spines (ASIS) to construct the proximal and distal borders of the trunk segment. A local anatomical coordinate system was defined by these landmarks, following the right hand rule. A plane passing through these four landmarks defined the y-z plane of the segment. The origin of the thorax segment was defined at the midpoint of the left and right acromion markers. The z-axis connected the midpoint of the ASIS markers and midpoint of the acromion markers. The x-axis was orthogonal to the y-z plane, pointing anteriorly. The y-axis was the cross-product of the x- and z-axes, pointing laterally to the subject's left. A triad of markers was also placed directly on the front of the IMU to track the motion of the trunk segment. A commercial software program (Visual3D v6.01.07, C-Motion, Inc., Germantown, MD) used the 3D coordinates of the triad markers, in the trunk anatomical coordinate frame, as input to estimate the trunk flexion angles and velocities at the recovery step and assess equivalence between the IMU and the motion capture system estimates.

The participants donned a safety vest and stood, facing the front of the treadmill, in the middle of the treadmill (DBCEEWI-Instrumented Treadmill, AMTI, Inc., Watertown, MA). The vest was clipped securely to a harness, which was attached to a ceiling mounted track with the freedom to move in the same direction as the treadmill belts. Subject-specific adjustments were made to the harness to ensure that when a fall occurred, the subject had enough range of motion to sufficiently recover from the disturbance on their own without contacting the treadmill. Prior to each trial, the subject was asked to assume an upright static, neutral posture. All of the subjects naturally kept their arms at their sides. A small, lightweight level was attached to the top IMU to gauge the subject's position.

Custom code (LabVIEW 2012, National Instruments, Austin, TX) was used to generate a series of forward and backward postural disturbances, which the participants experienced as simulated trips and slips, respectively. The participants were instructed to maintain an upright posture and to take a step, if necessary. A fall occurred if the harness clearly supported the participant's body weight after the given disturbance [17]. The disturbance direction was randomized to prevent the subjects from anticipating the disturbance and the magnitude incrementally increased each time the participant did not fall. When a fall occurred three times in one direction, the magnitude was decreased for that direction and testing resumed until the participant logged three falls in the opposite direction.

Trunk flexion angle and angular velocity were estimated at recovery step for both the motion capture system and the IMU with the orientation calculated with and without the magnetometer data. The Root Mean Square Error between the motion capture estimates and IMU estimates were calculated for both IMU orientation conditions. Equivalence testing was performed to demonstrate that the two methods were equivalent, with alpha set to 0.05 [18]. A Bland Altman plot was also prepared to demonstrate agreement between the two methods [19]. The results have clinical significance, based on pre versus post perturbation training results [5], if the trunk flexion angles are equivalent within 5° and angular velocity are equivalent within 44°/s.

3. Results

The number of disturbances for each subject ranged from 27 to 49 trials (average: 36±6), where the number of forward disturbances ranged from 11 to 23 trials (average: 18±3) and the number of backward disturbances ranged from 11 to 26 trials (average: 18±4). The forward disturbances magnitude ranged from $18\,m/s^2$ to $27\,m/s^2$ (average: $21\,m/s^2 \pm 3\,m/s^2$) and the backward disturbances magnitude ranged from $12\,m/s^2$ to $18\,m/s^2$ (average: $14\,m/s^2 \pm 2\,m/s^2$). The trunk angles estimated by IMU and Motion Capture at recovery step ranged from −25° to 75° and −17° to 63°, respectively, where negative angles were extension and positive were flexion. These values were consistent with physiological ranges of motion at the trunk. The trunk angular velocities estimated by IMU and Motion Capture at recovery step ranged from −198°/sec to 250°/sec and −161°/sec to 202°/sec, respectively, where again negative values were extension and positive values were flexion.

The Root Mean Square Error (RMSE) between the motion capture estimate and the IMU estimate with the orientation data from all three sensors for trunk flexion angle and trunk flexion angular velocity were, respectively, 5° (±5°) and 44°/s (44±°/s). While

Table 1
Equivalence Results at foot contact across all trials for all 10 subjects.

Variable	Equivalence Interval	95% CI for Equivalence	p-Value
Trunk Flexion Angle (degree)	[−2, 2]	[−1, 1]	0.005
Trunk Flexion Angular Velocity (degree/sec)	[−20, 20]	[0, 17]	0.011

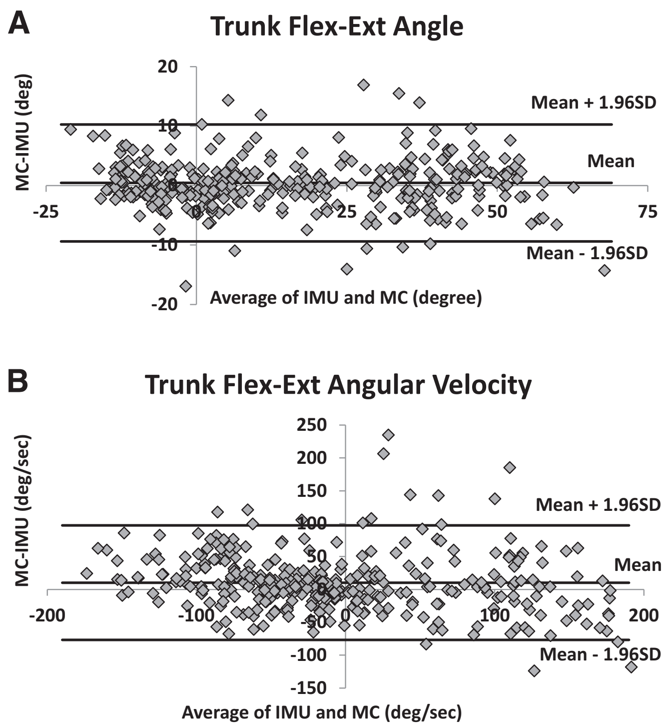

Fig. 1. Bland Altman plots for Motion Capture estimates compared to Inertial Measurement Unit estimates for trunk flexion-extension angle (A) and trunk flexion-extension angular velocity (B) across all trials for all 10 subjects. Each marker represents the value at recovery step for a single trial. The middle horizontal line represents the mean data value and the top and bottom horizontal lines represent the limits of agreement. Positive angles are flexion.

the RMSE for the two methods using the IMU orientation estimate from only the gyroscope and the accelerometer for trunk flexion angle and trunk flexion angular velocity were, respectively, 53° (±49°) and 60°/s (±60°/s). Given the large discrepancy between the two IMU orientation estimates, the remaining analysis was conducted using the orientation estimate from all three sensors. The Equivalence Test (Table 1) demonstrated for both the trunk flexion angle and angular velocity at foot contact that the methods were equivalent for the IMU orientation data generated using all three sensors. The angle and angular velocity equivalence intervals were within the criteria established for clinical significance. Additionally, the 95% Confidence Intervals were within the equivalence interval and the p-Values were less than alpha. The Bland-Altman plots show each individual trial for all 10 subjects (Fig. 1). A comparison of the mean trunk angle and mean trunk angular velocity for both the IMU and motion capture across all 10 subjects demonstrates that the methods are equivalent (Fig. 2). The data points in the figures represent the values for trunk flexion angle and angular velocity at recovery step. Each subject utilized a unique, postural recovery strategy, which led to a difference between subjects, but within each subject the trunk flexion kinematics were equivalent for both estimates.

4. Discussion

The maximum magnitude of the disturbances implemented during this study was 27 m/s^2. The maximum speeds testing in previous treadmill validation studies were 2.4 m/s [13] and 1.9 m/s [14]. While the accelerations were not reported for these previous studies, a similar study with comparable steady state velocities reported trunk flexion accelerations of 3.9 m/s^2 [20]. The accelerations implemented in this study were an order of magnitude

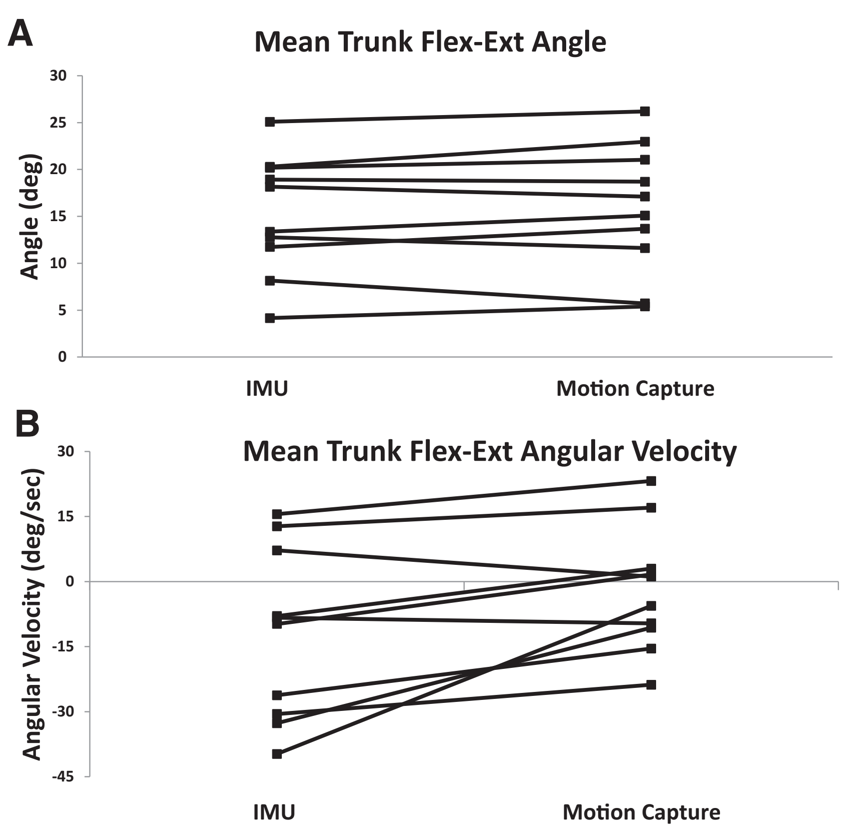

Fig. 2. Mean trunk angle (A) and trunk angular velocity (B) at recovery step across all trials for each subject for IMU and motion capture measurements. Each line represents a single subject. Positive angles are flexion.

greater than previous validation studies, which demonstrates that they were worthy of investigation.

The trunk flexion angle equivalence interval was comparable to results reported previously. Previous studies have reported trunk flexion angle accuracies ranging from $1.6 \pm 1.1°$ to $11 \pm 1°$ [9–14]. The trunk flexion angular velocity equivalence was also comparable to previous studies, where errors of $6 \pm 13°/s$ have been reported [10]. Additionally, other treadmill fall prevention training programs with success (recover)/fail (fall) outcomes reported a mean difference of 14° with 95% CI [8,19] for trunk angle and a mean difference of 84°/s with 95% CI [55, 113] for trunk angular velocity between outcomes [5,8,17,21,22]. The equivalence intervals for both the trunk angle and the angular velocity at foot contact were considerably less than the confidence intervals for these studies proving that differences between falls and recovery can be measured with both motion capture and IMUs. While the Bland Altman limits of agreement were greater than the Equivalence Test equivalence intervals, they were consistent with accuracy reported previously for fast walking [14]. Based on the results of this study, IMUs using orientation estimates from a commercially available Kalman filter with all three sensors have been shown to be a valid method for measuring trunk flexion angle and angular velocities during treadmill induced disturbances.

This study had limitations. While the same IMU was used for all subjects and specific attention was used to place the IMU in the same location and orientation on all the subjects, it is possible that there were differences with IMU placement across subjects. However, these differences were taken into account with the subject-specific calibration routine. In this study, the triad of tracking markers was taped to the IMU and the IMU was taped to the subject's sternum. It is plausible that the motion of the IMU and the triad were different than the true trunk motion. Additionally, while each subject was asked to assume a neutral posture for each trial, the position was subject specific and varied across subjects. The neutral position was only visually confirmed by the level on the IMU. While other researchers have reported drift error with IMU use [23], it doesn't apply to the current study. In this study each disturbance was 400 milliseconds and the IMU data collection lasted three seconds. The other studies had substantially longer data collections. The equivalence criteria established for this study was specific to clinical treadmill fall prevention rehabilitation training. Other applications may have different specifications and require tighter equivalence bands.

IMUs using a commercially available Kalman filter with input from all three sensors provide a valid method for measuring trunk flexion angles and angular velocity during forward and backward treadmill disturbances in a clinical setting. This instrumentation is appropriate for clinical fall prevention rehabilitation training applications.

Conflict of interest

The authors declare that they have no competing interests.

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Institutional review

Approval for human subject research was authorized under the Mayo Clinic Institutional Review Board, IRB 16–000754.

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