Short communication

Comparison of overhead harness configurations for measuring trunk kinematics during treadmill disturbances

Emily J. Miller, Kenton R. Kaufman*

Motion Analysis Lab, Department of Orthopedic Surgery, Mayo Clinic, Rochester, MN, USA

ABSTRACT

Background: Research has shown that postural perturbation training reduces falls. Trunk kinematics, at recovery step, are key outcome measures. Fall prevention training programs are being developed for routine clinical care. In these programs, the subject is positioned on a treadmill and subjected to forward and backward disturbances which simulate trips and slips. The patient wears a safety vest attached to an overhead harness to create a secure environment. Motion capture is used to measure trunk kinematics.

Research question: It is important to verify that trunk kinematics are independent of harness configuration. The purpose of this study was to compare the trunk flexion angle and angular velocity at recovery step after forward and backward disturbances on a treadmill measured by motion capture with the harness in both a fixed and free position.

Methods: Ten healthy young adults (5/10 female, age: 29.2 ± 6.3 years, BMI: 24.2 ± 2.4) participated in this study. The subjects had retro-reflective markers placed on key anatomical landmarks to measure trunk kinematics. The participants experienced forward and backward disturbances, which incrementally increased in intensity until the harness clearly supported the subjects for three disturbances in both directions. The order of harness configurations was randomized across subjects and each subject participated in two consecutive rounds of disturbances, one round for each harness configuration. Equivalence testing was performed to demonstrate that the harness configurations were equivalent ($\alpha = 0.05$).

Results and significance: The Equivalence Test demonstrated that the trunk angle (TA) and angular velocity (TAV) were equivalent for the different harness configurations. The 95% Confidence Intervals (TA: $[-2, 1]$, TAV: $[-18, 16]$) were within the equivalence interval (TA: $[-3, 3]$, TAV: $[-20, 20]$) and the p-Values (TA: 0.008, TAV: 0.034) were less than alpha. Trunk kinematics are independent of overheard harness configurations during treadmill induced disturbances for clinical postural perturbation training.

1. Introduction

Older adults and individuals with lower limb amputation are at high risk for falling. It has been reported that over 50% of lower limb amputees have fallen in the community [1,2] and out of these individuals over 25% self-reported an injury due to the fall [1]. Research has shown task-specific fall prevention rehabilitation training reduces falls in older aged women and individuals with lower extremity transtibial amputations [3,4]. Perturbation-based balance training is a reactive task-specific program aimed at mimicking real-life exposure to slips and trips in a safe, controlled environment [5]. Trunk flexion and velocity at recovery step determine the probability of a fall and are key outcome measures for fall prevention training [6,7]. In these training programs, the subject is positioned on a treadmill and subjected to forward and

backward disturbances which simulate trips and slips, respectively. The participant wears a safety vest that is attached to an overhead harness to provide a secure training environment and trunk kinematics are measured to track the participant's progress. Harness configurations differ across sites. The harness may be a fixed harness system [3,8] or have the freedom to slide along a track in the ceiling as the participant takes recovery steps [7,9,10]. For data comparison, it is important to validate that trunk kinematics are independent of the harness configuration. The purpose of this study was to compare the trunk flexion angle and angular velocity at recovery step after forward and backward disturbances on a treadmill measured by motion capture with the harness in both a fixed and free position.

* Corresponding author at: Mayo Clinic Rochester, 200 First Street SW, DAHLC 4, Rochester, MN, 55905, USA.
E-mail address: kaufman.kenton@mayo.edu (K.R. Kaufman).

2. Methods

Ten healthy young adults (5/10 female, 27.1 ± 4.2 years, BMI: 23.8 ± 2.7) were recruited for this study. Informed written consent was obtained from each subject prior to data collection. A 10-camera (Raptor-12, Motion Analysis Corporation, Santa Rosa, CA) motion analysis system was utilized to acquire kinematics. Retro-reflective, 3D markers were placed bilaterally on the acromion processes and the anterior superior iliac spines (ASIS) to define the proximal and distal ends of the trunk segment. A local anatomical coordinate system was constructed using these landmarks, following the right hand rule. The markers on the acromion processes and a marker on the scapula were used to track the trunk segment. The 3D coordinates of the marker data were used as input to a commercial software program (Visual3D v6.01.07, C-Motion, Inc., Germantown, MD) to calculate the trunk flexion angles and velocities at the recovery step as measured by the motion capture system.

The participants stood in the center of the treadmill (DBCEEWI-Instrumented Treadmill, AMTI, Inc., Watertown, MA) facing the front of the treadmill. The participants wore a safety vest that was secured to a harness, which was attached to a track in the ceiling 2.9 m above the base of the treadmill. The harness was adjusted for each subject to ensure that the subject did not contact the treadmill if a fall occurred, but also had enough range of motion to sufficiently recover from the disturbance on their own ability. Two harness configurations were compared in this study. In the free position, the harness freely moved forward and backward along the track in the same direction as the treadmill belts. In the fixed position, custom stops were locked in place around the harness-ceiling connection directly above the subject's head to create a fixed-single point harness configuration. Each subject participated in two consecutive rounds of disturbances. The order of harness configurations was randomized across subjects. Data collections occurred during a single visit and lasted two hours. Participants were given a ten minute break between rounds.

The participants experienced a series of forward and backward postural disturbances, generated using custom software code (LabVIEW 2012, National Instruments, Austin, TX), which caused trips and slips, respectively. The participants were instructed to take a step, if necessary, but to remain upright. A trial was considered a fall if the harness clearly supported the participant after the induced disturbance [11]. The direction of the disturbances was randomized to prevent the subjects from anticipating the disturbance and the magnitude increased each time the participant did not fall. When the participant fell three times in a given direction, it was considered a fail. The intensity was decreased for the fail direction. Testing continued and the disturbance magnitude was increased for the opposite direction until the participant fell three times in the opposite direction.

Trunk flexion angle and angular velocity were calculated at recovery step using the motion capture system for both the free and fixed harness configurations. Equivalence testing was performed to demonstrate that the two harness configurations were equivalent, with alpha set to 0.05 [12]. The results will have clinical significance, based on pre versus post perturbation training results [3], if the trunk flexion angles and angular velocity are equivalent within 5° and 60°/s, respectively.

3. Results and discussion

The total number of disturbances ranged from 45 to 77 trials (average: 60 ± 10), while the magnitude of the forward disturbances ranged from 14 m/s^2 to 22 m/s^2 (average: 18 m/s^2 ± 2 m/s^2) and the magnitude of the backward disturbances ranged from 10 m/s^2 to 18 m/s^2 (average: 13 m/s^2 ± 3 m/s^2). The trunk flexion angle at foot contact ranged from 36° extension to 45° flexion and the trunk angular velocity at foot contact ranged from 230°/s extension to 217°/s flexion. A previous trip study with healthy subjects reported maximum trunk flexion angles of 34° ± 11° and maximum trunk flexion velocity of

Table 1
Equivalence Results.

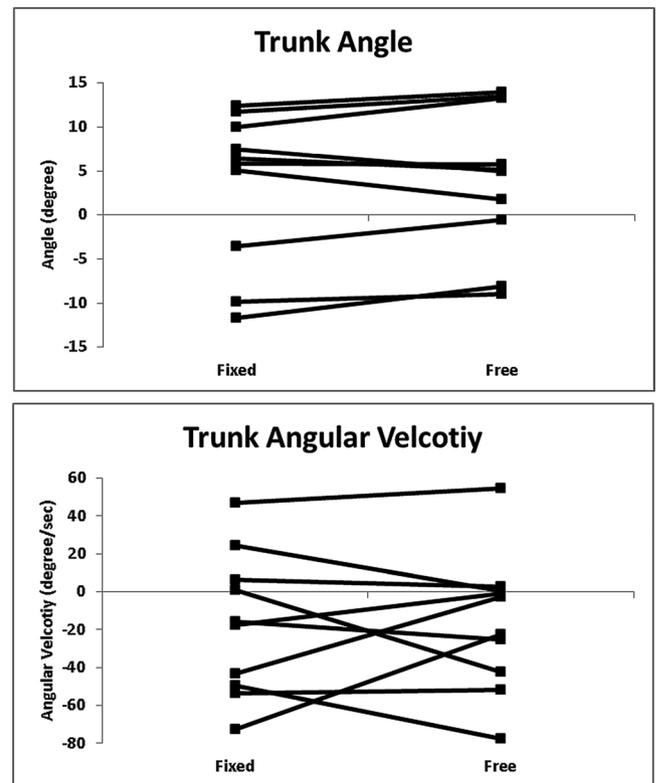| Variable	Equivalence Interval	95% CI for Equivalence	P-Value
Angle (deg)	[−3, 3]	[−2, 1]	0.008
Angular Velocity (deg/s)	[−20, 20]	[−18 16]	0.034

Fig. 1. Mean trunk angle (A) and trunk angular velocity (B) for fixed and free harness configurations. Each line represents a single subject. Flexion angles and velocities are positive.

139°/s ± 46°/s [13]. A previous slip study with healthy subjects reported maximum trunk extension angles of 40° ± 13° and maximum trunk extension velocity of 145°/s ± 23°/s [14].

The results of the Equivalence Test (Table 1) demonstrated that the harness configurations were equivalent for both the trunk angle and angular velocity. The 95% Confidence Intervals were within the equivalence interval and the P-Values were less than alpha. Additionally, the equivalence intervals were less than the predetermined ranges established for clinical significance. A comparison of the mean trunk angle and mean trunk angular velocity for both the free and fixed harness configurations across all 10 subjects demonstrated the equivalence of the harness configuration (Fig. 1). Each subject assumed a unique, subject-specific postural recovery strategy, which led to a spread between subjects, but the kinematics were equivalent for both harness configurations within each subject.

There were limitations associated with this study. While care was taken to place the reflective markers on key anatomical locations, it is possible that the motion of the markers was not the same as the motion of the trunk. Prior to each trial, the subjects were instructed to stand in an upright posture, but this posture was not measured and standardized across subjects.

4. Conclusions

Trunk kinematics are independent of overheard harness configurations during treadmill induced disturbances for clinical fall prevention rehabilitation.

Conflict of interest statement

The authors declare that they have no conflicts of interest.

Acknowledgements

References

[1] J.R. Crenshaw, et al., The circumstances, orientations, and impact locations of falls in community-dwelling older women, Arch. Gerontol. Geriatr. 73 (2017) 240–247.
[2] W.C. Miller, M. Speechley, B. Deathe, The prevalence and risk factors of falling and fear of falling among lower extremity amputees, Arch. Phys. Med. Rehabil. 82 (2001) 1031–1037.
[3] K.R. Kaufman, et al., Task-specific fall prevention training is effective for warfighters with transtibial amputations, Clin. Orthopaed. Relat. Res. 472 (2014) 3076–3084.
[4] M.D. Grabiner, et al., Trunk kinematics and fall risk of older adults: translating biomechanical results to the clinic, J. Electromyogr. Kinesiol. 18 (2) (2008) 197–204.
[5] M.H.G. Gerards, et al., Perturbation-based balance training for falls reduction among older adults: current evidence and implications for clinical practice, Geriatr. Gerontol. Int. 17 (2017) 2294–2303.
[6] J.R. Crenshaw, K.R. Kaufman, M.D. Grabiner, Compensatory-step training of healthy, mobile people with unilateral, transfemoral or knee disarticulation amputations: A potential intervention for trip-related falls, Gait Posture 38 (2013) 500–506.
[7] T.M. Owings, M.J. Pavol, M.D. Grabiner, Mechanisms of failed recovery following postural perturbations on a motorized treadmill mimic those associated with an actual forward trip, Clin. Biomech. 16 (2001) 813–819.
[8] E. Protas, et al., Gait and step training to reduce falls in Parkinson's disease, NeuroRehabilitation 20 (3) (2005) 183–190.
[9] W.-N. Bair, et al., Kinematic and behavioral analyses of protective stepping strategies and risk for falls among community living older adults, Clin. Biomech. 36 (2016) 74–82.
[10] J.R. Crenshaw, K.R. Kaufman, The intra-rater reliability and agreement of compensatory stepping thresholds of healthy subjects, Gait Posture 39 (2014) 810–815.
[11] C.F. Honeycutt, M. Nevisipour, M.D. Grabiner, Characteristics and adaptive strategies linked with falls in stroke survivors from analysis of laboratory-induced falls, J. Biomech. 49 (2016) 3313–3319.
[12] E. Walker, A.S. Nowacki, Understanding equivalence and noninferiority testing, J. Gen. Intern. Med. 26 (2) (2011) 192–196.
[13] M.D. Grabiner, J.W. Feuerbach, D.W. Jahnigent, Measures of paraspinal muscle performance do not predict initial trunk kinematics after tripping, J. Biomech. 29 (6) (1996) 735–744.
[14] P. Parijat, T.E. Lockhart, J. Liu, EMG and kinematic responses to unexpected slips after slip training in virtual reality, IEEE Trans. Biomed. Eng. 62 (2) (2015) 593–599.