Available online at www.sciencedirect.com
SCIENCE DIRECT®
Clinical Biomechanics 18 (2003) 737–744
CLINICAL BIOMECHANICS
www.elsevier.com/locate/clinbiomech

Can motion of individual body segments identify dynamic instability in the elderly?

Michael E. Hahn, Li-Shan Chou *

Department of Exercise and Movement Science, Biomechanics Laboratory, University of Oregon, Eugene, Oregon 97403, USA

Received 26 November 2002; accepted 17 June 2003

Abstract

Objective. To determine if medio-lateral motion of the head, trunk, or pelvis demonstrates dynamic stability as well as whole-body center of mass during obstructed walking.

Design. Group comparison of two elderly populations using whole-body motion analysis.

Background. Detection of imbalance through analysis of center of mass motion is commonly adopted, requiring three-dimensional reconstruction of a multi-link biomechanical model. It would be advantageous clinically if similar detection could be made by analyzing segmental displacements of the pelvis, trunk, or head.

Methods. Healthy elderly adults and elderly patients with balance disorders walked over level ground and crossed obstacles of height ranging from 2.5% to 15% of body height. Whole-body center of mass was calculated as the weighted sum of segmental centers of mass. Group differences in medio-lateral displacements and peak velocities of head, trunk, pelvis, and the center of mass were analyzed using a two-way ANOVA with repeated measures for obstacle height.

Results. Elderly patients with balance disorders exhibited greater medio-lateral displacement and peak velocities of all segments. However, significant group differences were only detected in the center of mass displacement and peak velocity.

Conclusion. Whole-body center of mass motion distinguishes elderly patients with balance disorders from healthy peers more consistently than markers representing the head, trunk, or pelvis. Large variation of individual segment motion makes dynamic stability difficult to assess. This study demonstrates that center of mass motion allows more sensitive detection of dynamic instability.

Relevance

Detection of dynamic instability in at-risk individuals before falls occur will allow preventative interventions, preserving quality of life in the elderly population.

Keywords: Dynamic instability; Center of mass; Elderly; Balance disorders; Gait analysis; Obstacle crossing

1. Introduction

A very serious problem faced by the growing elderly population is an increased susceptibility to falls due to age-related declines in balance control. It has been estimated that one third of the elderly population (>65 years) experiences at least one fall per year (Tinetti et al., 1988; Campbell et al., 1990; Sattin, 1992). Half of these falls are reported to occur during locomotion (Ashley et al., 1977; Prudham and Evans, 1981). One of the determinant risk factors reported for traumatic hip fracture is falls to the side (Greenspan et al., 1998). Traumatic falls resulting in fracture account for a great portion of the disability, death, and medical costs in this growing population (Tinetti et al., 1988; Campbell et al., 1981; Hayes et al., 1996). Fall prevention strategies are most effective if people at risk can be identified before they are injured. Therefore, biomechanical research is needed to identify measures that can sensitively detect and quantify dynamic instability in elderly adults. Given that imbalance and tripping over obstacles during gait were reported as two of the most common causes of falls in the elderly (Campbell et al., 1990; Overstall, 1977;

* Corresponding author.
E-mail address: chou@uoregon.edu (L.-S. Chou).

0268-0033/$ - see front matter © 2003 Elsevier Ltd. All rights reserved.
doi:10.1016/S0268-0033(03)00139-6

Tinetti and Speechley, 1989), the use of obstacle crossing as a laboratory testing paradigm has been documented in several clinical and biomechanical studies (Chen et al., 1991; Patla and Riedyk, 1993; Means, 1996; Chou and Draganich, 1997; Begg and Sparrow, 2000; McFadyen and Prince, 2002).

The dynamic balance control system is known to respond to extrinsic risks encountered during locomotion, modulating appropriate responses and adjustments to the motor plan according to the intrinsic abilities of the sensory systems (somatosensory, vestibular, vision) and neuromuscular response functions. As balance maintenance is perturbed, the balance control system is thought to apply reactive feed-forward corrections through the musculoskeletal system to control whole body center of mass (CoM) motion, by relocating the center of pressure (CoP). This general model of balance control has been well established (Murray et al., 1967; Prieto et al., 1993; Maki and McIlroy, 1996; Winter et al., 1998) and demonstrates that the CoP is continually adjusted to reach beyond the CoM sway affecting a direction change, thus returning the CoM back to a more stable position. The aging process is thought to impose degenerative changes in sensory systems which compromise an individual's ability to modulate appropriate interactions between the CoM and CoP.

Many studies have used whole-body CoM motion and its interaction with the CoP as indicators to examine an individual's dynamic stability, demonstrating consistent motion of the CoM, tightly regulated to pass between the alternating CoP of each supporting foot (Jian et al., 1993; MacKinnon and Winter, 1993; Prince et al., 1994; Winter, 1995; Kaya et al., 1998; Krebs et al., 1998; Polcyn et al., 1998; Tucker et al., 1998; Chou et al., 2001). In particular, MacKinnon and Winter (1993) demonstrated that differences in CoM and CoP locations in the frontal plane provide a valuable means for defining dynamic stability. Recent studies have demonstrated that examination of the CoM motion could better distinguish subjects with balance disorders from healthy subjects when compared to the temporal-distance parameters (Tucker et al., 1998; Chou et al., in press). These findings suggest that there is relatively high variability in selecting different movement strategies to enhance dynamic stability (i.e., control of CoM motion) during walking among older adults or patients with balance disorders. Furthermore, with the addition of an obstacle, frontal plane CoM motion could be a more sensitive measurement of dynamic stability (Chou et al., in press).

Accurate estimation of the whole body CoM requires three-dimensional reconstruction of a multiple segment biomechanical model. This technical requirement may restrict the applicability of assessing dynamic instability via the whole body CoM motion. One potential alternative would be to track individual body segments that may indicate instability by displaying changes in segmental motion. It was reported that similar vertical CoM displacement could be estimated with a sacral marker when compared to a multi-segmental model (Saini et al., 1998). There was a lack of information on their agreement in the medio-lateral (M-L) direction however. Also, in a clinical setting, the head, trunk and pelvis appear to be intuitive choices due to their clear visibility during locomotion. A number of studies have suggested that upper body motion, especially head movement, is a valid reference for dynamic equilibrium (Berthoz and Pozzo, 1988; Pozzo et al., 1990; Mouchnino et al., 1990; Alexander et al., 1992; Pozzo et al., 1995; Wu, 1998). However, the testing protocols of these studies involved perturbations of quiet stance, and did not include dynamic tasks such as level walking or obstacle crossing. Currently, it is not known whether motion of the head, trunk, or pelvis segments may be used as dynamic stability references during gait.

Therefore, the purpose of this study was to investigate whether the M-L motion of an individual body segment while negotiating an obstacle during gait could be used to differentiate between older adults with and without balance disorders. It was hypothesized that whole-body CoM motion during obstacle crossing is more sensitive in detecting dynamic instability than individual segment motions; as demonstrated by significant differences between samples of healthy elderly and those with balance impairment. Secondly, if required to monitor whole-body dynamic stability without CoM capability, we sought to determine which body segment's motion would best distinguish healthy elderly from those with balance disorders. Due to close proximity with the whole-body CoM it was hypothesized that motion of the pelvis would be the 'next-best' indicator of dynamic stability, second to the CoM.

2. Methods

Nine healthy elderly adults, 7 male and 2 female; 72 years (SD, 6.4 years), 172.3 cm (SD, 10.3 cm), 75 kg (SD, 15.7 kg), and six elderly patients with balance disorders, 1 male, 5 female; 76 years (SD, 3.9 years), 162.1 cm (SD, 7.8 cm), 70 kg (SD, 15.1 kg) were recruited for this study. The experimental protocol was approved by the Institutional Review Board and experimental procedures were explained to all subjects prior to testing, with verbal and written consent obtained.

Healthy elderly subjects were recruited from the surrounding community and had to be free of vertigo, lightheadedness, unsteadiness, or a history of falls, with no history of significant head trauma, neurologic disease or musculoskeletal impairments. Elderly patients with balance problems were recruited from the Vestibular/

Balance Laboratory at the Mayo Clinic (Rochester, MN, USA), where the patients were referred for evaluation due to their complaints of "dizziness" or "unsteadiness" during walking. Three of the patients were diagnosed with either unilateral or bilateral vestibular weakness. More detailed information of their clinical evaluations was reported previously (Chou et al., in press). All patients were community dwelling and able to walk more than 100 m without the use of a gait aid at the time of testing. The mental status of all healthy subjects and patients was assessed using the Folstein Mini-Mental test and all were required to have a score of 24 or higher (Folstein et al., 1975).

The experimental protocol included level unobstructed walking and crossing of obstacles set to a height equal to 2.5%, 5%, 10%, or 15% of each individual's body height. In this way, the difficulty of obstacle crossing was normalized to individual body height, accounting for variation within the sample. The lowest height ($\sim$5 cm) represents that of a door threshold, and the greatest height ($\sim$25 cm) represents the height of a high curb or stair. Subjects performed all trials while barefoot. They were asked to walk at a self-selected pace along a 6-m walkway, stepping over the obstacle and to continue at a normal pace after crossing. Each subject was allowed to select his/her preferred limb for leading over the obstacle. Subjects were allowed at least three complete steps prior to the obstacle to ensure that a comfortable pace was reached before encountering the obstacle. Crossing stride was defined from the heel-strike of the trailing limb before the obstacle to the heel-strike of the same limb after crossing the obstacle. The obstacle was made of two adjustable upright standards and a padded crossbar resulting in a diameter of 2.5 cm, and a length of 2 m. The crossbar rested loosely on the standards and would dislodge easily if foot contact were made. Unobstructed walking trials were performed first, followed by obstacle-crossing trials. For the obstacle conditions, obstacle height was randomly selected for each trial. Three trials were analyzed for each obstacle condition.

Whole body motion analysis was performed, using a 6-camera ExpertVision™ system (Motion Analysis Corp., Santa Rosa, CA, USA). Twenty-seven reflective markers were placed on bony landmarks of each subject. The three-dimensional marker trajectory data were collected at 60 Hz and low-pass filtered using a fourth-order Butterworth filter with cutoff frequency of 8 Hz. The descriptive temporal-distance parameters common in gait analysis were also determined in all conditions.

M-L excursion of markers on the superior aspect of the head, right acromion process (trunk), and right anterior superior iliac spine (pelvis) were targeted as possible descriptors of M-L stability. M-L displacement of each marker was defined as the M-L range (maximum–minimum) of that marker's trajectory during the obstacle-crossing stride. Additionally, M-L excursion of the whole-body CoM was examined as a descriptor of M-L stability. The position of the whole body CoM was computed as the weighted sum of each body segment's CoM using a 13-link biomechanical model (Jian et al., 1993; Meglan, 1991; Chou et al., 2001). Virtual markers were created in EVa software (Version 6.0, Motion Analysis Corp.) to estimate internal segment endpoints from the external markers, and the relative positions of segmental CoMs. Anthropometric reference data for this model were adapted from Dempster's initial work (Winter, 1990). Linear velocities were calculated using the generalized, cross-validated spline algorithm (Woltring, 1986).

The dependent variables of this study consisted of temporal-distance gait parameters, displacement and peak velocity of the CoM in the M-L direction, and M-L displacement and peak velocity of the head, trunk, and pelvis during the obstacle-crossing stride. Effects of subject group and obstacle height condition (including level, unobstructed gait) on the dependent variables were assessed using a two-way ANOVA with repeated measures of obstacle height. Significance level was set at $\alpha = 0.05$. For those dependent variables showing significant differences for obstacle height, a polynomial test was performed to determine the trend (linear, quadratic, or cubic). Post hoc analysis was also carried out for obstacle height effect using matched pairs *t*-tests with Bonferroni adjustment, allowing multiple comparisons. All statistical analyses were conducted with SYSTAT (Version 9, SPSS Inc.).

3. Results

No significant differences were detected between subject groups for any of the temporal-distance parameters investigated in this study. Temporal-distance data for all conditions are presented in Table 1. No incidents of tripping occurred for any of the obstacle height conditions in either subject group.

Patterns of M-L displacement trajectories of the head, trunk, pelvis and whole body CoM were similar in both subject groups. Typical M-L displacement trajectories of the head, trunk, pelvis and whole body CoM during the crossing stride of a representative trial, with obstacle height set at 15% body height, are shown for a healthy control subject and a patient with imbalance in Fig. 1. Additionally, representative M-L velocity profiles of the head, trunk, pelvis and CoM during the crossing stride for a healthy subject and a patient with imbalance are presented in Fig. 2. No significant group differences were detected for the M-L displacement of the head, trunk, or pelvis marker (Table 2). The effect of increasing obstacle height resulted in significant increases in M-L head and pelvis displacements ($P = 0.008$ and

Table 1
Gait temporal-distance measurements for both groups during the crossing stride; group mean (SD)

Obstacle height	None		2.5%		5%		10%		15%		Effect
Parameters	Control	Patients	Control	Patients	Control	Patients	Control	Patients	Control	Patients	
Gait velocity (m/s)	1.179 (0.145)	1.018 (0.131)	1.112 (0.173)	0.990 (0.125)	1.114 (0.178)	0.911 (0.153)	1.009 (0.194)	0.865 (0.157)	0.974 (0.147)	0.810 (0.169)	**
Stride time (s)	1.092 (0.090)	1.123 (0.093)	1.170 (0.171)	1.270 (0.169)	1.219 (0.213)	1.309 (0.157)	1.306 (0.222)	1.368 (0.125)	1.327 (0.212)	1.454 (0.145)	**
Stride length/ body height	0.744 (0.063)	0.700 (0.056)	0.744 (0.061)	0.745 (0.065)	0.761 (0.057)	0.726 (0.077)	0.746 (0.064)	0.722 (0.089)	0.737 (0.069)	0.717 (0.100)	
Step width/inter-ASIS distance	0.363 (0.086)	0.371 (0.093)	0.386 (0.100)	0.532 (0.210)	0.382 (0.100)	0.492 (0.164)	0.414 (0.052)	0.526 (0.178)	0.434 (0.053)	0.600 (0.275)	**

** Significant obstacle height effect ($P < 0.01$).

$P = 0.02$, respectively). However, elderly patients with balance disorders demonstrated significantly greater M-L CoM displacement ($P = 0.03$) across all obstacle conditions when compared to the healthy elderly group (Table 2). The M-L displacement of the CoM was found to significantly increase with obstacle height ($P = 0.039$).

Similarly, no significant group differences were detected for the peak M-L velocity of the head, trunk, or pelvis marker (Table 3), although differences in peak velocity of the pelvis approached significance ($P = 0.054$). Stepping over a higher obstacle also resulted in significant increases in the peak M-L head velocity ($P = 0.005$) and peak M-L trunk velocity ($P = 0.003$). The peak M-L CoM velocity was significantly greater in the elderly patients with balance disorders ($P = 0.026$; Table 3) and was found to significantly increase with obstacle height ($P = 0.041$). Post hoc analysis revealed that a significant effect of obstacle height existed between the 5% and 15% of body height obstacle conditions for both M-L CoM displacement and peak velocity.

4. Discussion

This study sought to determine whether an older individual with dynamic instability could be detected by examining the M-L motion of the head, trunk and pelvis, or the whole body CoM. Inability to adequately control whole body motion in the frontal plane may lead to loss of balance resulting in a sideways fall. This has been reported as one of the most important risk factors for hip fractures among frail elderly nursing home fallers (Greenspan et al., 1998). It is reasonable to expect that M-L motion of the whole body and/or its major segments may serve as a functional indicator, identifying persons at greater risk of sideways falls. Results of this study support our primary hypothesis by demonstrating that M-L motion of the head, trunk, or pelvis does not provide the same level of sensitivity as the whole body CoM to distinguish older patients with balance disorders from healthy older adults. Of these segment motions, only differences in the peak M-L velocity of the pelvis approached significance ($P = 0.054$). This finding did not directly support our secondary hypothesis but suggested that the pelvic motion might be the next-best indicator of dynamic imbalance when motion of the whole body CoM is not available.

Another common method for deriving information about CoM movement has been the use of force platform ground reaction force (GRF) data (Crowe et al., 1995; Saini et al., 1998). The advantage of this technique is the ability to determine whole-body motion without requiring kinematic information and anthropometric estimation models. Two integrations are necessary to derive CoM position from the acceleration term of the GRF vector. During each integration, a time-integral constant is required and would make the computation of CoM position highly individualized to the initial temporal-distance conditions of each trial. Such a requirement might limit its ability to accurately calculate CoM position for non-cyclic activities, such as obstacle crossing. Both methods, kinematic collection with anthropometric models and GRF double integration, provide certain benefits as well as limitations inherent in the assumptions belonging to each.

Previous studies have reported that control of head motion may be a viable reference for dynamic equilibrium (Berthoz and Pozzo, 1988; Pozzo et al., 1990; Mouchnino et al., 1990; Alexander et al., 1992; Pozzo et al., 1995; Wu, 1998). The present study measured segment and whole-body motion during the dynamic tasks of gait and obstacle crossing, whereas previous studies examined control of segment motion during perturbation of quiet stance and the subsequent recovery of balance. Increased variability is therefore possible in the subjects' response to the perturbations, exhibited by self-selected crossing strategies. In this way, patterns of stability seen during quiet stance may not be exhibited in dynamic gait. It is therefore difficult to compare results

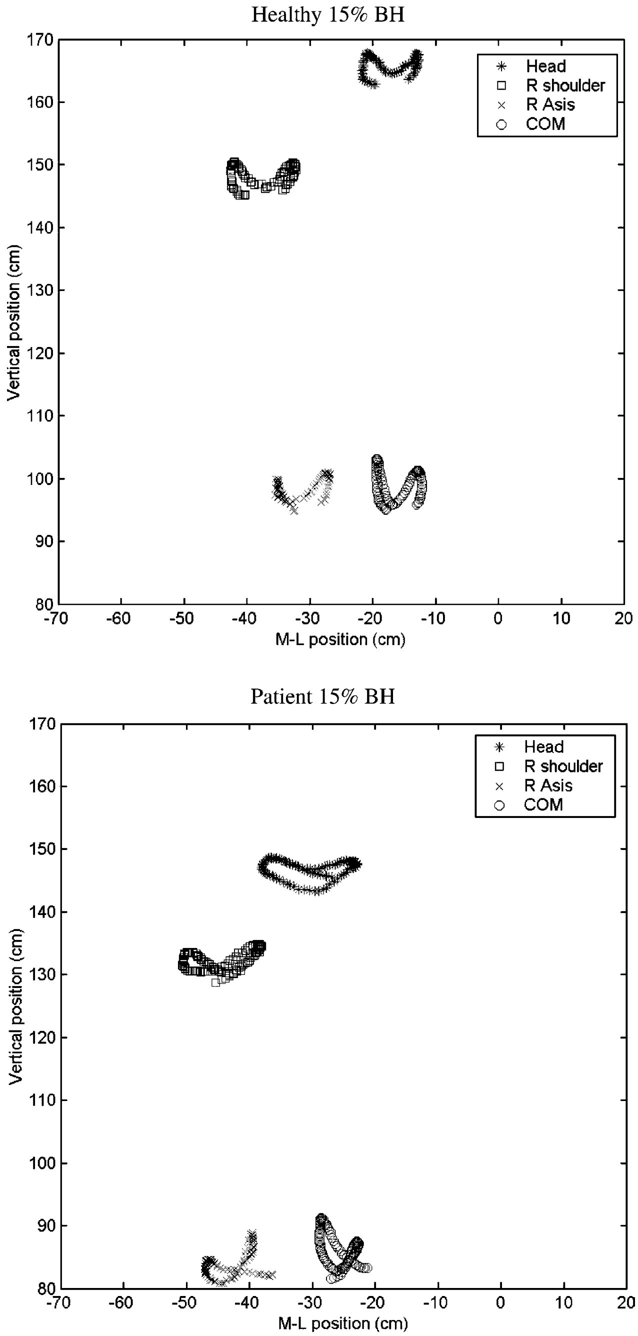

Fig. 1. Representative frontal plane trajectories of markers on the head, trunk, and pelvis, and whole body CoM during the crossing stride over a 15% BH obstacle. Trajectories from a healthy elderly subject (above) and an elderly patient with balance disorder (below). BH: body height.

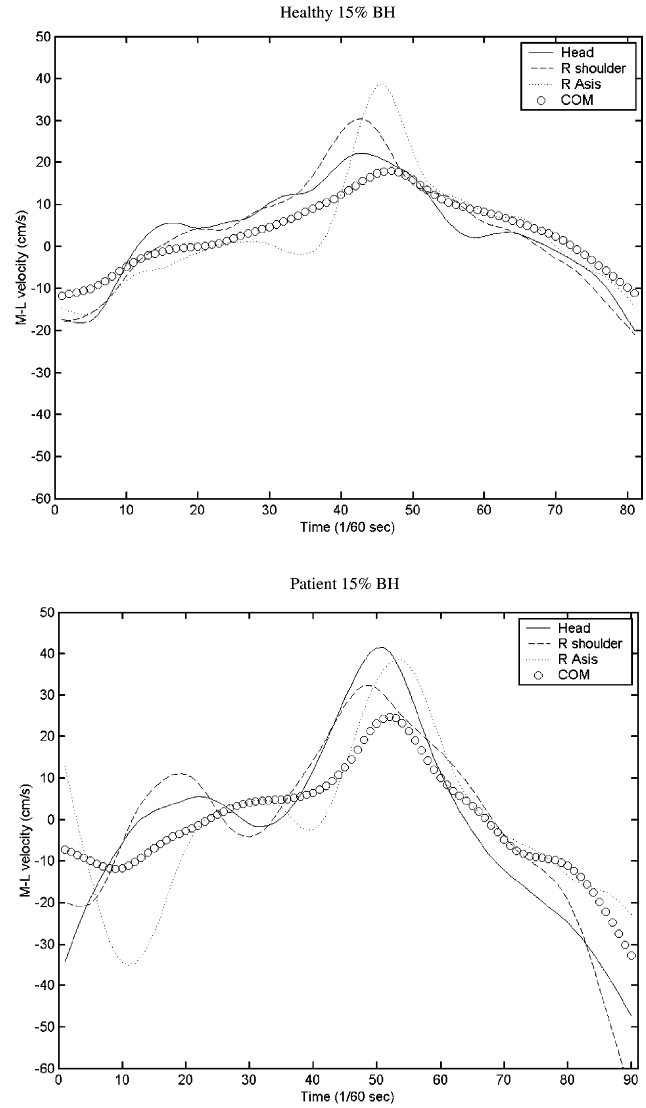

Fig. 2. Medio-lateral velocity patterns of the head, trunk, pelvis and CoM during the crossing stride over a 15% BH obstacle, representing a healthy elderly subject (above) and an elderly patient with balance disorder (below). The whole body CoM velocity was noted to maintain a tighter range than those of the segmental markers.

from previous segment motion studies with the current findings.

Increased M-L CoM motion in the elderly patients during obstacle crossing demonstrated the difficulty of maintaining dynamic stability in the frontal plane. Both the healthy elderly and elderly patient groups maintained similar magnitudes of M-L CoM motion during unobstructed walking. On the other hand, while negotiating obstacles, the elderly patients demonstrated significantly greater and faster M-L motion of the CoM than the healthy elderly adults. Results of the post hoc analysis revealed significant increases for both M-L CoM displacement and peak velocity between the level walking condition and each individual obstacle height ($P < 0.01$ with Bonferroni adjustment). The level of challenge to the dynamic balance control system appears to increase as the obstacle height increases, allowing otherwise subtle differences to be detected between the CoM motion of healthy elderly adults and elderly patients with balance disorders. However, similar

Table 2
Ranges of medio-lateral displacements (cm) of the head, trunk, pelvis, and CoM during the crossing stride; group mean (SD)

Obstacle height	None		2.5%		5%		10%		15%		Effect
Segment	Control	Patients	Control	Patients	Control	Patients	Control	Patients	Control	Patients	
Head	6.6 (2.2)	5.2 (1.3)	7.6 (3.3)	8.1 (3.1)	7.1 (1.5)	7.1 (3.2)	7.8 (3.2)	8.4 (2.7)	8.4 (3.8)	10.3 (4.4)	a
Trunk	5.2 (1.0)	5.4 (1.7)	6.1 (1.7)	8.3 (3.5)	6.5 (1.6)	7.6 (3.6)	7.1 (2.5)	8.6 (3.1)	8.5 (3.3)	10.0 (4.5)	
Pelvis	4.6 (0.9)	5.9 (2.4)	5.5 (1.3)	8.3 (2.5)	5.6 (1.9)	6.9 (2.7)	5.7 (1.1)	8.3 (2.4)	6.6 (1.7)	9.0 (3.4)	a
CoM	3.6 (0.8)	4.4 (1.9)	4.3 (1.1)	7.1 (3.0)	4.4 (1.3)	5.3 (1.9)	4.5 (1.5)	6.0 (2.2)	5.2 (1.5)	6.8 (3.2)	b,a

[a] Significant obstacle height effect ($P < 0.05$).
[b] Significant group effect ($P < 0.05$).

Table 3
Peak medio-lateral velocities (cm/s) of the head, trunk, pelvis, and CoM during the crossing stride; group mean (SD)

Obstacle height	None		2.5%		5%		10%		15%		Effect
Segment	Control	Patients	Control	Patients	Control	Patients	Control	Patients	Control	Patients	
Head	25.3 (10.0)	18.2 (4.5)	25.3 (8.0)	26.1 (6.3)	26.5 (6.6)	27.4 (9.4)	28.8 (6.4)	28.8 (9.8)	31.7 (8.9)	33.1 (12.5)	a
Trunk	20.9 (4.7)	18.7 (4.8)	25.0 (4.3)	27.1 (9.7)	25.6 (5.1)	26.5 (9.1)	28.3 (4.3)	29.3 (10.7)	30.6 (5.0)	34.3 (13.9)	a
Pelvis	19.3 (4.7)	23.3 (9.0)	23.7 (5.5)	33.2 (10.9)	24.4 (6.6)	29.0 (9.6)	25.4 (5.1)	28.8 (7.9)	25.5 (4.7)	30.1 (9.4)	
CoM	12.9 (3.3)	13.6 (6.0)	14.6 (3.0)	20.6 (8.9)	13.9 (3.5)	17.7 (7.1)	14.8 (3.0)	17.9 (7.2)	15.7 (2.4)	21.0 (9.7)	b,a

[a] Significant obstacle height effect ($P < 0.05$).
[b] Significant group effect ($P < 0.05$).

differences were not detected in the motion of head, trunk or pelvis.

Large variation in individual segment motion could be related to diverse movement strategies adopted by individuals in response to perturbed balance during locomotion. It is likely that individual segment motion may be compensatory in response to motion of other segments, or in response to a sense of instability itself. Adjustments of segmental motion are governed to maintain a smooth and controllable whole body CoM motion, which represents the resultant product of balance control on the entire system. Therefore, M-L motion of the CoM could better distinguish dynamic instability in patients with balance disorders, as compared to motion of the individual segments of the head, trunk, and pelvis.

It might be speculated that body height could have an influence on the M-L segmental motion given that our healthy elderly adults were taller than the elderly patients. If such an association exists, it is then reasonable to expect that a taller person would exhibit greater M-L upper body motion. However, our results indicated that, in general, the M-L segmental motion of elderly patients was greater than that of healthy elderly adults. Beyond this, further analysis of our data was performed to include individual body height as a covariate. The results indicated that there were no significant associations between any of the M-L segmental/CoM motion values and body height (P values ranged from 0.417 to 0.897). Nevertheless, further investigations are needed to address any possible stature-related differences in the CoM motion.

A limitation of this study is the apparent gender difference between the two groups tested. The majority of patients with imbalance in this study were female, compared to a primarily male control group. Significant gender differences in both the crossing speed and step length during obstacle crossing were reported previously (Chen et al., 1991). However, no meaningful gender difference was found for the step width, which is more related to the COM motion in the frontal plane. Further investigation is needed to address any gender-related differences in COM motion. One other limitation of this study is the sample size of the patient group. As this group consisted of only six subjects, sample variation within the segment motion parameters may have contributed to insignificant trends found in the group comparisons. A larger sample size may enhance the detection of sideways instability with the pelvic motion. However, small sample size did not inhibit the findings of significant group difference in M-L CoM displacement or peak M-L velocity.

The findings of this study demonstrated that examining M-L motion of the head, trunk or pelvis does not provide sensitive detection of dynamic instability. However, relative increases in M-L CoM motion due to the increased challenge of crossing obstacles can successfully identify persons who are at greater risk of imbalance. Information about an individual's ability to control their CoM trajectory during obstacle crossing will allow us to better identify individuals at risk for imbalance and falls.

Acknowledgements

This study was supported in part by the Mayo Foundation, the National Institutes of Health (AG 05770) and the Oregon Medical Research Foundation.

References

Alexander, N.B., Shepard, N., Gu, M.J., Schultz, A., 1992. Postural control in young and elderly adults when stance is perturbed: kinematics. Journals of Gerontology: Medical Science 47 (3), M79–M87.

Ashley, M.J., Gryfe, C.I., Annies, A., 1977. A longitudinal study of falls in an elderly population. II. Some circumstances of falling. Age and Ageing 6, 211–220.

Begg, R.K., Sparrow, W.A., 2000. Gait characteristics of young and older individuals negotiating a raised surface: implications for the prevention of falls. Journals of Gerontology: Medical Sciences 55 (3), 147–154.

Berthoz, A., Pozzo, T., 1988. Intermittent head stabilization during postural and locomotory tasks in humans. In: Amblard, B., Berthoz, A., Clarac, F. (Eds.), Development, adaptation and modulation of posture and gait. Elsevier, Amsterdam, pp. 189–198.

Campbell, A.J., Reinken, J., Allan, B.C., Martinez, G.S., 1981. Falls in old age: a study of frequency and related clinical factors. Age and Ageing 10, 264–270.

Campbell, A.J., Borrie, M.J., Spears, G.F., Jackson, S.L., Brown, J.S., Fitzgerald, J.L., 1990. Circumstances and consequences of falls experienced by a community population 70 years and over during a prospective study. Age and Ageing 19, 136–141.

Chen, H.-C., Ashton-Miller, J.A., Alexander, N.B., Schultz, A.B., 1991. Stepping over obstacles:Gait patterns of healthy young and old adults. Journals of Gerontology 46, M196–M203.

Chou, L.S., Draganich, L.F., 1997. Stepping over an obstacle increases the motions and moments of the joints of the trailing limb in young adults. Journal of Biomechanics 30, 331–337.

Chou, L.-S., Kaufman, K.R., Brey, R.H., Draganich, L.F., 2001. Motion of the whole body's center of mass when stepping over obstacles of different heights. Gait and Posture 13, 17–26.

Chou, L.-S., Kaufman K.R., Hahn M.E., Brey R.H. Medio-lateral motion of the center of the center of mass during obstacle crossing distinguishes elderly individuals with imbalance. Gait and Posture, in press.

Crowe, A., Schiereck, P., de Boer, R.W., Keessen, W., 1995. Characterization of human gait by means of body center of mass oscillations derived from ground reaction forces. IEEE Transactions on Biomedical Engineering 42 (3), 293–303.

Folstein, M.F., Folstein, S.E., McHugh, P.R., 1975. "Mini-mental state": a practical method for grading the cognitive state of patients for clinicians. Journal of Psychiatric Research 12, 189–198.

Greenspan, S.L., Myers, E.R., Kiel, D.P., Parker, R.A., Hayes, W.C., Resnick, N.M., 1998. Fall direction, bone mineral density, and function: risk factors for hip fracture in frail nursing home elderly. American Journal of Medicine 104, 539–545.

Hayes, W.C., Myers, E.R., Robinovitch, S.N., Van Den Kroonenberg, A., Courtney, A.C., McMahon, T.A., 1996. Etiology and prevention of age-related hip fractures. Bone 18, 77S–86S.

Jian, Y., Winter, D.A., Ishac, M.G., Gilchrist, L., 1993. Trajectory of the body COG and COP during initiation and termination of gait. Gait and Posture 1, 9–22.

Kaya, B.K., Krebs, D.E., Riley, P.O., 1998. Dynamic stability in elders: momentum control in locomotor ADL. Journals of Gerontology 53A, M126–M134.

Krebs, D.E., Jette, A.M., Assmann, S.F., 1998. Moderate exercise improves gait stability in disabled elders. Archives of Physical Medicine and Rehabilitation 79, 1489–1495.

MacKinnon, C.D., Winter, D.A., 1993. Control of whole body balance in the frontal plane during human walking. Journal of Biomechanics 26, 633–644.

Maki, B.E., McIlroy, W.E., 1996. Postural control in the older adult. Clinical Geriatric Medicine 12 (4), 635–658.

McFadyen, B.J., Prince, F., 2002. Avoidance and accommodation of surface height changes by healthy, community-dwelling, young, and elderly men. Journals of Gerontology: Biological Sciences 57 (4), 166–174.

Means, K.M., 1996. The obstacle course:a tool for the assessment of functional balance and mobility in the elderly. Journal of Rehabilitation Research and Development 33, 413–428.

Meglan D.A., 1991. Enhanced analysis of human locomotion, Ph.D. dissertation, The Ohio State University, OH, USA.

Mouchnino, L., Aurenty, R., Massion, J., Pedotti, A., 1990. Coordinated control of posture and equilibrium during leg movement. In: Brandt, T. (Ed.), Xth International Symposium on Disorder of Posture and Gait. Verlag fur Medizin, Munich, pp. 68–72.

Murray, M.P., Seireg, A., Scholz, R.C., 1967. Centre of gravity, centre of pressure and supportive forces during human activities. Journal of Applied Physiology 23, 831–838.

Overstall, P.W., Exton-Smith, A.N., Imms, F.J., Johnson, A.L., 1977. Falls in the elderly related to postural imbalance. British Medical Journal 1, 261–264.

Patla, A.E., Rietdyk, S., 1993. Visual control of limb trajectory over obstacles during locomotion: effect of obstacle height and width. Gait and Posture 1, 45–60.

Polcyn, A.F., Lipsitz, L.A., Kerrigan, D.C., Collins, J.J., 1998. Age-related changes in the initiation of gait: degeneration of central mechanism for momentum generation. Archives of Physical Medicine and Rehabilitation 79, 1582–1589.

Pozzo, T., Berthoz, A., Lefort, L., 1990. Head stabilization during various locomotory tasks in humans I. Normal subjects. Experimental Brain Research 82, 97–106.

Pozzo, T., Levik, Y., Berthoz, A., 1995. Head and trunk movements in the frontal plane during complex dynamic equilibrium tasks in humans. Experimental Brain Research 106, 327–338.

Prieto, T.E., Myklebust, J.B., Myklebust, B.M., 1993. Characterization and modeling the postural steadiness in the elderly: A review. IEEE Transaction of Rehabilitation Engineering 1, 26–34.

Prince, F., Winter, D.A., Stergiou, P., Walt, S.E., 1994. Anticipatory control of upper body balance during human locomotion. Gait and Posture 2, 19–25.

Prudham, D., Evans, J.G., 1981. Factors associated with falls in the elderly: a community study. Age and Ageing 10, 141–146.

Saini, M., Kerrigan, D.C., Thirunarayan, M.A., Duff-Raffaele, M., 1998. The vertical displacement of the center of mass during walking: a comparison of four measurement methods. Journal of Biomechanical Engineering 120 (1), 133–139.

Sattin, R.W., 1992. Falls among older persons: a public health perspective. Annual Review of Public Health 13, 489–508.

Tinetti, M.E., Speechley, M., 1989. Prevention of falls among the elderly. New England Journal of Medicine 320, 1055–1059.

Tinetti, M.E., Speechley, M., Ginter, S.F., 1988. Risk factors for falls among elderly persons living in community. New England Journal of Medicine 319, 1071–1077.

Tucker, C.A., Ramirez, J., Krebs, D.E., Riley, P.O., 1998. Center of gravity dynamic stability in normal and vestibulopathic gait. Gait and Posture 8, 117–123.

Winter, D.A., 1990. Biomechanics and Motor Control of Human Movement. John Wiley & Sons, New York. pp. 56–57.

Winter, D.A., 1995. Human balance and posture control during standing and walking. Gait and Posture 3, 124–193.

Winter, D.A., Patla, A.E., Prince, F., Ishac, M., Gielo-Perczak, K., 1998. Stiffness control of balance in quiet standing. Journal of Neurophysiology 80, 1211–1221.

Woltring, H.J., 1986. A FORTRAN package for generalized, cross-validatory spline smoothing and differentiation. Advances in Engineering Software 8, 104–113.

Wu, G., 1998. Age-related differences in body segmental movement during perturbed stance in humans. Clinical Biomechanics 13, 300–307.