Journal of Gerontology: BIOLOGICAL SCIENCES
2000, Vol. 55A, No. 9, B425–B431

Standing Up From a Chair as a Dynamic Equilibrium Task: A Comparison Between Young and Elderly Subjects

France Mourey,[1] Alexandre Grishin,[2] Philippe d'Athis,[3] Thierry Pozzo,[4] and Paul Stapley[4]

[1]Groupe d'Etude et de Recherche sur le Handicap (GERSH), Faculté de Médecine, Dijon, France.
[2]Institute for Information Transmission Problems, Russian Academy of Sciences, Moscow.
[3]Service de Biostatistique et Informatique Médicale, CHRU, Dijon, France.
[4]Groupe d'Analyse du Mouvement (GAM), Université de Bourgogne, Dijon, France.

The purpose of this study was to analyze and compare the features of center of mass (CoM) control along anterior/posterior axis in young and elderly subjects during sit-to-stand (STS). From a sitting position, seven healthy young subjects and seven healthy elderly subjects were asked to stand up from a chair under different experimental conditions (visual conditions: normal and blindfolded; speed: normal and as fast as possible). Analysis of results was based upon the concept of a "dynamic equilibrium area" (DEA), which in turn identified the dynamic limits of balance. The results showed that both the maximal CoM velocity in the horizontal axis and the CoM velocity at the instant of seat-off were found to be lower in elderly compared with young subjects. Concerning the maximal CoM velocity, the difference was increased under blindfolded condition. The position of CoM in the phase plane (i.e., velocity according to displacement) at the instant of seat-off was found to be shifted backward in elderly subjects. From these results we can deduce that age-related modifications can be observed in the control of the horizontal CoM motion during STS in healthy elderly subjects.

THE transition of posture associated with sit-to-stand (STS) movements requires a large displacement of the body center of mass (CoM) toward a base of support (BoS) becoming smaller in size. To achieve this, it has been shown that both CoM displacement and velocity are controlled along anterior/posterior and vertical axes (1). Following seat-off, such control must be achieved within limits of the BoS provided by the feet, in order to maintain a final static upright posture.

Using pattern analyses of CoM velocity along both anterior/posterior and vertical axes, several authors (2,3) have described three phases during STS: (i) the acceleration phase, which is defined from the beginning of the movement until maximal CoM horizontal velocity; (ii) the transition phase, beginning at maximal CoM horizontal velocity and terminating at maximal vertical CoM velocity; and (iii) the deceleration phase, which is defined from peak vertical CoM velocity until the end of movement. In terms of equilibrium, the transition phase corresponds to reduction in BoS from three-point support to two-point support. Moreover, during this phase a coordinated control of CoM position during STS is necessary in two directions, CoM velocity decreases along the horizontal axis just before seat-off, but increases along the vertical axis at the instant of seat-off. The control of CoM horizontal motion has been well documented by Pai and colleagues in young subjects (1,4). Results from these studies showed that subjects deliberately limited peak horizontal momentum (product of the velocity of the CoM and the body mass) when faced with conditions of increasing speed. However, they were able to voluntarily increase CoM momentum above that required to rise from the chair when permitted to fall forward using the arms on a support bar to stop the fall, rather than maintaining upright stance. Such features of CoM motion were interpreted as deliberate strategies of dynamic balance during STS.

Age-related changes in balance control have previously been analyzed using a movable platform in order to test the automatic postural responses to support surface perturbations. The results showed both modifications in the temporal organization of motor response and in the amplitude (5). Links between posture and voluntary movement have also been studied in elderly subjects (6). These authors highlighted evidence for a slowing of postural responses related to arm movements. However, the effects of aging upon postural control during a task requiring large displacements of the CoM in both forward and upward directions, such as is common to STS, has until now been poorly documented in healthy elderly subjects.

Nevertheless, clinical studies have demonstrated a correlation between difficulties in performing STS movements and balance disorders in elderly persons (7). Kaya and colleagues (8) have demonstrated changes in dynamic parameters of STS motion and gait in elderly subjects with bilateral vestibular hypofunction. In these subjects the linear and angular momentum was decreased compared to healthy elderly subjects. This result showed modification in strategy aiming to decrease disequilibrium created by the locomotor activities. Moreover, Alexander and associates (9) have compared a group of young subjects to a group of elderly subjects including those able to rise without the use of armrests and those unable to rise without the use of armrests. Results of this study showed the importance of postural stability control in STS, particularly in the impaired group. In order to understand the mechanisms underlying an effect of age on strategies used during STS, postural or equilibrium control must be explored in greater depth.

A previous study, comparing young and elderly subjects during standing and sitting movements, described angular velocity adjustments of trunk and knee in the elderly group, particularly following seat-off (10). Such results could be interpreted in terms of an alteration of CoM control during the critical transition phase. The present study attempted to explore the nature of these alterations by examining the parameters of dynamic equilibrium under different experimental conditions of light and speed. Analysis of results was based upon the concept of a "dynamic equilibrium area" (DEA). CoM position and velocity were used to define this area, which in turn identified the dynamic limits of balance. A similar approach has been used for postural control during bipedal robot locomotion (11). This method allows the comparison of strategies used for the control of CoM motion between young and healthy elderly subjects. It was hypothesized that consistent differences between the two groups would be found in the position and velocity of the CoM at the instant of seat-off. This in turn would reflect changes with age in the equilibrium control system during STS. Implications of this method in clinical studies are also discussed.

Materials and Methods

Subjects

Data were obtained from 7 healthy young subjects (3 women and 4 men), aged between 20 and 25 years (mean age 22.8 ± 1.5 years) and 7 healthy older subjects (6 women and 1 man) aged between 71 and 82 years (mean age 75.1 ± 4.4 years) (Table 1). Young subjects were physiotherapy students. Older subjects were recruited from a senior citizens' club. Potential elderly subjects were screened by a geriatrician in order to exclude any neurological diseases, peripheral neuropathologies, or musculoskeletal conditions that could have limited their movements. All subjects volunteered and gave their informed consent prior to participation.

Procedures

Subjects were seated on an armless chair, the height of which was adjusted to correspond to 100% of each subject's knee height. A back support was also adjusted so that the trunk was aligned in a vertical position. The arms were folded across the chest. The feet were placed flat, 10 cm apart at the heels, with the shanks making a 20° forward flexion relative to the vertical. Subjects were instructed to stand up from the chair, to remain standing for 2 seconds, and to return to the seated position. Subjects executed four blocks of three trials. Two blocks consisted of changing visual conditions (normal and blindfolded) and two blocks were of different speed (normal and as fast as possible). Subjects were given a rest period of approximately 2 minutes between each block.

Table 1. Characteristics of Studied Groups

	Young	Elderly
Number	7	7
Mean age and *SD* (years)	22.8 ± 1.5	75.1 ± 4.4
Range (years)	20–25	71–82
Height (cm)	167.5 ± 8.4	164.5 ± 8.9
Weight (kg)	63.5 ± 7.5	61.6 ± 12.9
Shank length (cm)	43.8 ± 6.1	45.1 ± 3.6

Recording system.—Movements of specific anatomical sites of the body were measured using a 100 Hz optoelectronic movement ELITE analyzer (BTS, Milan, Italy) that computed the spatial coordinates of small reflective markers (0.5 cm in diameter) glued to the skin. Two cameras (sampling frequency 100 Hz) placed 3 m from the subject's sagittal axis (left side), one on top of the other, 1 m and 2 m from the ground, recorded movements of eight markers fixed on the left side of the body at the following sites:

- the head: external canthus of the eye; auditory meatus.
- the trunk: acromion; side of the trunk (at the level of the seventh rib).
- the lower limb: hip (trochanter); knee (interstitial joint space); ankle (external malleolus); foot (fifth metatarsophalangeal).

During the experiments, both the chair and the subject's feet were placed on the 6-component dynamometric force platform (AMTI, Watertown, MA) allowing the measurement of three components (horizontal, vertical, and mediolateral) of applied force, and horizontal and mediolateral coordinates of the center of foot pressure (CP).

Data Analysis

Timing.—Recorded data were filtered using a Butterworth fourth order low-pass filter (cutoff frequency 6 Hz). In order to determine the beginning and the end of the motion, an angle (θ_1) between the trunk (the link between the acromion and the trochanter marker) and the vertical was calculated. The onset of intentional movement was defined as the first 10-millisecond frame at which values of forward angular velocity of the trunk exceeded a threshold of 10% of peak angular velocity during the entire STS movement. In the same way, the end of the movement was defined as this value decreased below a threshold of 10% of the peak. The threshold of 10% of peak of vertical linear displacement of the trochanter marker was used to identify the instant of seat-off.

CoM positions.—Sagittal CoM positions were calculated using a five-segment, rigid mathematical model (Figure 1), consisting of the following appendicular and axial segments: head, thorax, lower trunk, thigh, and shank. The arms were included in the thorax segment, and the leg segments comprised both the thigh and the shank. Only six of the eight markers were used: auditory meatus, acromion, side of the trunk, hip, knee, and ankle. Only motions of the CoM in the sagittal plane were explored in the present study. The coordinate system was defined as illustrated in Figure 1: *Y*-axis = upward-directed, *X*-axis = backward-directed, with the origin being placed at the center of the ankle joint.

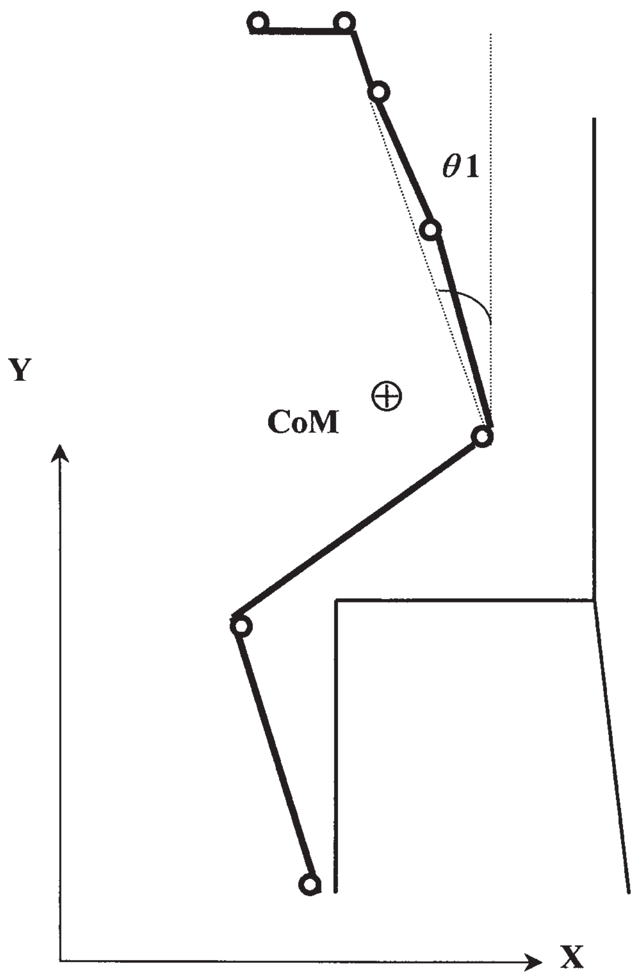

Figure 1. Stick diagram of the five-limb model including the head, thorax, lower trunk, thigh, and shank used in the calculation of the CoM position. θ_l correspond to the spatial angle between the trunk and the vertical.

The coordinates X_i, Y_i of the center of mass of a segment number i were calculated using the following formulae:

$$X_i = X_{1i} + l_i(X_{2i} - X_{1i}),\quad Y_i = Y_{1i} + l_i(Y_{2i} - Y_{1i}), \qquad (1)$$

where X_{1i}, Y_{1i}, X_{2i}, Y_{2i} are coordinates of the end markers of the i-th limb; l_i is the ratio between the distance of the proximal marker to the segments CoM, and its length. Coordinates X, Y of the common CoM were thus calculated using the formulae:

$$X = \sum_{i=1}^{5} m_i X_1 / \sum_{i=1}^{5} m_i \quad Y = \sum_{i=1}^{5} m_i Y_i / \sum_{i=1}^{5} m_i \qquad (2)$$

with m_i being the mass of the i-th segment. Anthropometric parameters m_i, l_i were assumed to be equal to average ones, and thus were taken from the literature (12). The coordinates of the position of the CoM were corrected using force platform data, and assuming that before the movement the position of the CP corresponded to that of the CoM. Thus, a constant value was added to X in order that calculated CoM position equalled the CP during quiet sitting. The velocity of the CoM was calculated by numeric derivation of position (displacement) values.

Dynamic equilibrium area (DEA).—For the CoM motion analysis, a model of simple inverted pendulum has been used (Figure 2).

The equation of motion of this model for a small angle α between the leg and the vertical is:

$$\tau^2 \ddot{\alpha} = \alpha - M/(mgl) \qquad (3)$$

where time constant τ equals $\sqrt{l/g}$ where l is the height of the CoM, g is the acceleration due to gravity, and points denote time derivation. After multiplying both parts of the equation by l and replacing $l\alpha$ with X (abscissa of the CoM) and with P (abscissa of a center of pressure of ground reaction force [CP]), the following formula can be obtained:

$$\tau^2 \ddot{X} = X - P \qquad (4)$$

The CP must lie between bounds: $X_t \le P \le X_h$, where X_t, X_h are abscissas of the toe and the heel, respectively. There are two lines in the phase plane X,X separating phase trajectories of Equation (4) when the value of P is constant and equals to one of the two limiting values. These two lines bound the region:

$$X_t < X + \tau\dot{X} < X_h \qquad (5)$$

where X_t and X_h are coordinates of the toe and heel, respectively. This condition describes a zone inside of the phase plane bounded by two parallel lines, which cross the X axis at the points $x = X_t$ (toe) and $x = X_h$ (heel) (Figure 3). For zero velocity, Equation 5 yields the well known condition of static equilibrium, that the projection of the CoM lies within the supporting area. Thus, Equation 5 can be regarded as a generalization of static equilibrium conditions, where CoM

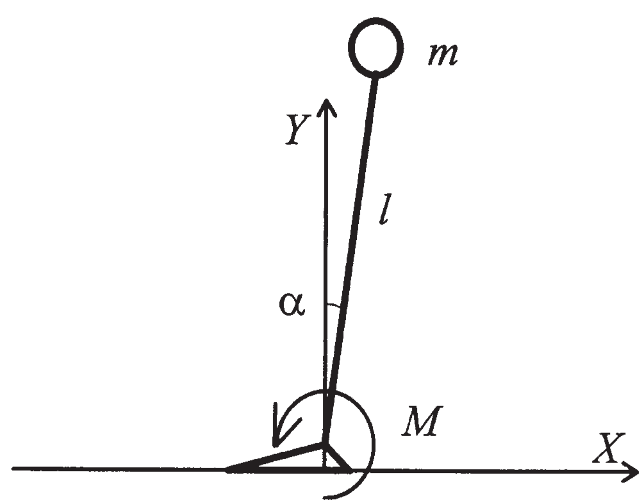

Figure 2. The pendulum consists of weightless "leg" with a point mass m comprising the whole mass of the body. The leg stands on a foot. A torque M can be applied in the ankle joint, but this torque is limited by foot length.

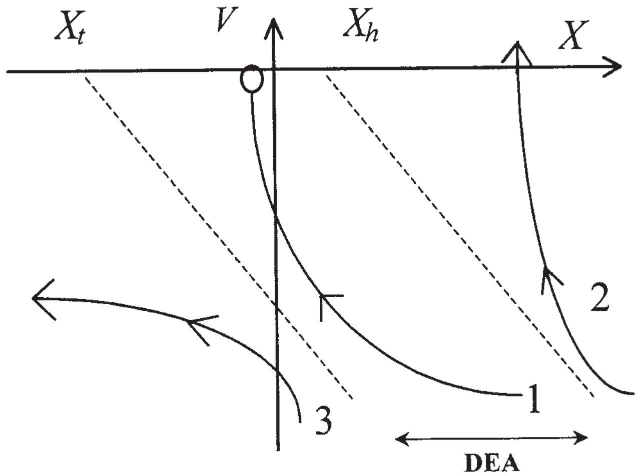

Figure 3. The zone described by Equation 5, dynamic equilibrium area. If the starting position is outside the region between the dashed lines, it is impossible to cross the boundary of the region and to bring the CoM into area of support with zero velocity. Thus, the system will fall backward (curve 2) or forward (curve 3). So, only motion of CoM represented by points from the region, Equation 5, can be executed within an equilibrium state.

velocity exceeds zero. We named this zone, defined by Equation 5, as the dynamic equilibrium area (DEA).

Normalization.—In order to compare subjects with different anthropometric parameters, a normalization was made using the following formulae:

$$\tilde{X} = X/L_f,\ \tilde{V} = V/(L_f/\tau)$$

where $\tilde{X}$,$\tilde{V}$ are normalized CoM position and velocity, and $L_f = X_h - X_t$ is foot length.

Statistical Analysis

The different periods corresponding to the different combinations of vision and speed conditions were not randomized over periods owing to the complexity of the measurement: all the periods were of four types: normal speed/normal vision; as fast as possible/normal vision; normal speed/blindfolded; and as fast as possible/blindfolded. Data were then studied with an analysis of variance considering as a block the four successive combinations of vision and speed of each subject and taking into account:

- the age effect (variation between young and elderly);
- the period effect (variation between the 4 different periods);
- the Age × Period interaction.

When a period effect was found to be significant, vision and speed effects were studied separately by linear contrasts between the corresponding means for normal speed versus as fast as possible, and normal vision versus blindfolded. If the Age × Period interaction was significant, these means were compared separately in the young subjects and in the elderly subjects.

A result was considered significant at the $p < .05$ level.

Results

General Movement Characteristics in Young Subjects

A description of the general characteristics of STS has been achieved on the basis of both dynamic and kinematic analyses. Figure 4 shows, from top to bottom, angular velocity relative to the vertical of trunk, the vertical displacement of hip marker, the CoM and CP displacement along the anterior/posterior axis, and CoM velocity in the forward direction for one typical young subject. Preceding seat-off, the trunk demonstrated forward flexion. Following seat-off, the direction of trunk movement was inverted (moving in extension). There was a backward displacement of the CP preceding seat-off, indicating that CP and CoM moved in opposite directions. The maximal value of CoM velocity was attained before seat-off. The delay between time to maximal CoM velocity and seat-off was small (0.03 ± 0.02 s).

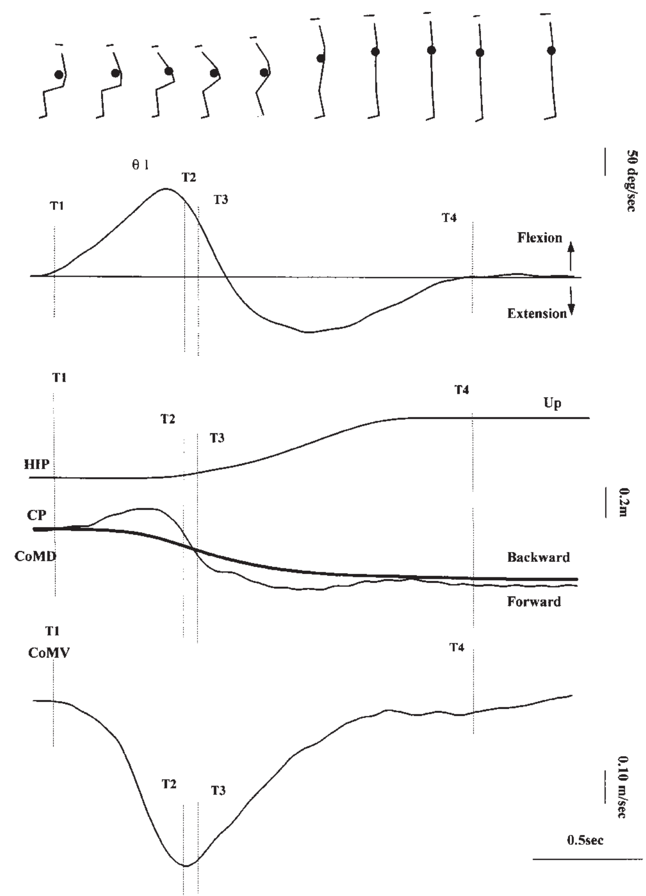

Figure 4. Kinetic and kinematic analysis of a young subject. Top: angular velocities of trunk (θ_l). Middle: displacement of center of mass (CoMD in bold) and displacement of center of pressure of ground force (CP), both along the anterior/posterior axis, vertical displacement of hip marker (HIP). Bottom: horizontal projection of center of mass velocity (CoMV). The following *instants* were identified: beginning (T1); CoM maximal horizontal velocity (T2); seat-off (T3); end of motion (T4). The complete motion is represented by the sequence of stick figures, and displacement of CoM is shown as filled black dots.

Comparison Under Different Experimental Conditions

Movement times.—For movement times there were significant age and period effects without a significant interaction: the means were higher in the elderly subjects and, within each group, lower at high speed but not significantly different with vision (Table 2).

CoM velocity.—The normalized *X*-axis projection of CoM velocity ($\tilde{V}$) was analyzed at the peak of this velocity ($\tilde{V}$ max) and at the instant of seat-off ($\tilde{V}$ seat-off). The negative value of CoM velocity corresponds to the orientation of X axis only.

For $\tilde{V}$ max there was a significant Period by Age interaction ($p < .05$) that may be interpreted in different ways (Table 3):

- The absolute means were significantly lower in elderly subjects, the difference being greater in blindfolded condition (mean elderly–mean young = 0.168 for normal speed, 0.175 for fast speed) than under normal vision (mean elderly–mean young = 0.089 for normal speed, 0.133 for fast speed).
- The absolute means were significantly lower in blindfolded condition in elderly subjects only.
- The absolute means were significantly higher in fast condition than in normal condition for both the elderly subjects and the younger subjects.

For $\tilde{V}$ seat-off there were significant age and period effects without significant interaction: the absolute values were significantly lower in elderly subjects, and only the speed effect within each group was significant (Table 4).

CoM position in the phase plane at the seat-off.—Figure 5 shows the position of the CoM into DEA at the instant of seat-off under normal speed and normal vision, for all subjects. CoM positioning near the toe boundary of the DEA increased the risk to fall forward, whereas a positioning near the heel boundary increased the risk of a backward fall. Position and velocity values have been normalized. All points lie inside the DEA except for a few trials of elderly subjects. The position of CoM outside the DEA may be explained by the fact that these subjects continued to touch the chair after the detected instant of seat-off.

The points corresponding to the elderly subjects have been shifted toward a lower velocity, as previously mentioned, and toward the heel boundary of the DEA. The distance to this boundary (Δ_{DEA}) was estimated using the formula:

$$\Delta_{DEA} = \left|\tilde{V}seatoff\right| - (\tilde{X}seatoff - \tilde{X}_h)$$

where $\tilde{X}seatoff$ and $\tilde{V}seatoff$ are normalized CoM position and velocity at the seat-off instant. This formula gives a length of a line segment parallel to the *X* axis drawn along the phase plane from the representative point up to the heel boundary of the DEA (the distance from this point to the heel boundary of DEA is equal to $\Delta_{DEA}/\sqrt{2}$).

This length (Δ_{DEA}) was found to be significantly smaller in elderly subjects than in young ones ($p < .05$) (Table 5). In other words, the position of CoM in the phase plane was located nearer to the limits of a fall backward in elderly than in young subjects. Neither effect of vision nor speed effect was found significant.

Table 2. Statistical Analysis for Movement Times

Period	Vision†	Speed‡	Age*: Young Mean (*SD*)	Age*: Elderly Mean (*SD*)
1	Normal	Normal	1.366 (0.244)	1.821 (0.309)
2	Normal	Fast	0.977 (0.142)	1.396 (0.358)
3	Blindfolded	Normal	1.353 (0.167)	1.826 (0.453)
4	Blindfolded	Fast	1.040 (0.190)	1.326 (0.315)

Note: Nonsignificant interaction between period and age.
*Significant age effect ($p < .05$).
†Nonsignificant vision effect.
‡Significant speed effect ($p < .05$).

Table 3. Statistical Analysis for the Peak of Center of Mass Velocity: $\tilde{V}$ max

Period	Vision†	Speed‡	Age*: Young Mean (*SD*)	Age*: Elderly Mean (*SD*)
1	Normal	Normal	–0.632 (0.104)	–0.543 (0.103)
2	Normal	Fast	–0.737 (0.104)	–0.604 (0.103)
3	Blindfolded	Normal	–0.671 (0.085)	–0.503 (0.069)
4	Blindfolded	Fast	–0.752 (0.094)	–0.577 (0.093)

Note: Significant interaction between age and period.
*Significant age effect ($p < .05$).
†Vision effect
– not significant in the younger subject.
– significant in the elderly subject ($p < .05$).
‡Speed effect
– significant in the younger subject ($p < .05$).
– significant in the elderly subject ($p < .05$).

Discussion

During STS movements, the area of support decreases. At the moment of sitting, the CoM is located behind the position of the feet. In order to achieve an upright posture, however, it must be brought inside the foot area. This modification requires the control of both position and velocity of CoM along the horizontal axis (13).

In agreement with a previous study (3), the analysis of kinematic data indicated that the CoM reached maximal horizontal velocity before seat-off. The comparison between the two groups showed that the maximal CoM velocity in the horizontal axis and the CoM velocity at the instant of seat-off were lower in elderly compared to young subjects. Thus,

Table 4. Statistical Analysis for $\tilde{V}$ Seat-off

Period	Vision†	Speed‡	Age*: Young Mean (*SD*)	Age*: Elderly Mean (*SD*)
1	Normal	Normal	–0.602 (0.111)	–0.495 (0.096)
2	Normal	Fast	–0.716 (0.078)	–0.556 (0.084)
3	Blindfolded	Normal	–0.634 (0.775)	–0.478 (0.088)
4	Blindfolded	Fast	–0.729 (0.096)	–0.535 (0.081)

Note: Nonsignificant interaction between period and age.
*Significant age effect ($p < .05$).
†Nonsignificant vision effect.
‡Significant speed effect ($p < .05$).

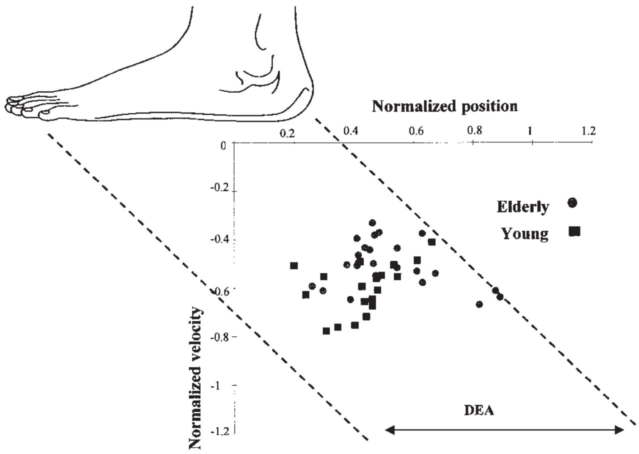

Figure 5. The normalized velocity of CoM was plotted relevant to displacement of CoM at the instant of seat-off in young (■) and elderly (●) subjects for normal condition. Dashed lines indicate the boundary of the dynamic equilibrium area (DEA).

elderly subjects began vertical motion (at the instant of seat-off) with a lower horizontal velocity compared to young subjects. This result can appear as expected if we consider that elderly subjects achieved STS more slowly than young subjects. The elderly subjects were able to increase motion speed, but under all experimental conditions the movement time remained greater than in young subjects. If we consider the movement time, no vision effect was found, and we can think that STS is a very automatized task mainly dependent upon somatosensory information (10).

However, the analysis of vision revealed a greater difference between elderly and young subjects under blindfolded condition for the maximal CoM velocity along the horizontal axis. This finding shows that the visual feedback is necessary to elderly subjects when the CoM velocity reaches the peak value. In this case, the movement control needs visual information and cannot be based on feedforward process only. This inability to adapt to vision privation suggests a difficulty in integration within the central nervous system (14,15). Moreover, this finding shows that the maximal CoM velocity allows a pertinent approach of movement control in the aging process.

The analysis of horizontal CoM velocity as a function of displacement allowed the definition of DEA, which determined the mechanical limits of motion. Assuming that CoM motion following seat-off depended upon ankle torque only, movement termination at a state of equilibrium was found to be impossible outside the DEA. A similar zone has been constructed by Pai and Patton (13), using computer analysis of the human whole body model. Here it has been constructed analytically.

Our results showed a difference in the control of the CoM within the DEA between healthy young and elderly subjects. The position of CoM in the phase plane at the instant of seat-off was found to be shifted backward in elderly subjects. This decreased a risk of falling forward but simultaneously increased the risk of a backward fall. This modification of equilibrium conditions in elderly subjects can be interpreted as a protective attitude against the risk of forward falling. A change in the perceived stability limits with aging could explain such a protective attitude, which can be interpreted as a compensatory process with regards to a decrease in postural stability (16). It is also interesting to note that clinical observations often show a trend toward backward disequilibrium in elderly patients who present a history of falls.

According to the method, the whole body weight cannot influence the results, but an effect of mass distribution should be found. We think that the difference between mass distribution in the young and elderly subjects in our population would have a minor impact. However, in future studies the influence of this parameter could be analyzed in other groups. Moreover, further research will be required to fully elucidate the relationship between the decrease of muscular strength with age and the strategy modification. An obvious relationship between muscle weakness and functional loss in elderly subjects presenting functional limitations has been reported (17). On the contrary, this relationship has not been found in healthy elderly subjects.

Finally, we suggest that the present analysis might be used in clinical practice in order to identify the strategies adopted and to assess results of rehabilitation in pathological elderly subjects.

Table 5. Statistical Analysis for the Distance From Center of Mass Positioning to the Hill Boundary of the Dynamic Equilibrium Area: Δ_{DEA}

			Age*	
Period	Vision	Speed	Young Mean (*SD*)	Elderly Mean (*SD*)
1	Normal	Normal	0.381 (0.211)	0.170 (0.175)
2	Normal	Fast	0.379 (0.186)	0.147 (0.164)
3	Blindfolded	Normal	0.354 (0.185)	0.132 (0.185)
4	Blindfolded	Fast	0.392 (0.208)	0.144 (0.150)

Note: Nonsignificant interaction between period and age.
*Significant age effect ($p < .05$).
†Nonsignificant vision effect.
‡Nonsignificant speed effect ($p < .05$).

Acknowledgments

This research was supported by a grant from the Institut National de la Santé et de la Recherche Médicale (INSERM/CRE n°92107). The authors gratefully acknowledge the contribution of the Centre National d'Etudes Spatiales for providing a part of the equipment needed for this research.

Address correspondence to F. Mourey, Service de Médecine Gériatrique, Centre Hospitalier Régional, Université de Dijon, 2, rue Jules Violle, 21034 Dijon Cédex, France. E-mail: france.mourey@planetb.fr

References

1. Pai YC, Lee WA. Effect of a terminal constraint on control of balance during sit-to-stand. *J Mot Behav*. 1994;26:247–256.
2. Riley PO, Schenkman ML, Mann RW, et al. Mechanics of a constrained chair rise. *J Biomech*. 1991;24:77–85.
3. Roebroek ME, Doorenbosh CAM, Harlaar J, et al. Biomechanics and muscular activity during sit-to-stand transfer. *Clin Biomech*. 1994;9:235–244.
4. Pai YC, Rogers MW. Control of body mass transfer as a function of speed of ascent in sit-to-stand. *Med Sci Sports Exer*. 1990;22:378–384.

5. Woollacott MH, Shumway-Cook A, Nashner LM. Aging and posture control: changes in sensory organization and muscular coordination. *Int J Aging Hum Dev*. 1986;23:97–114.
6. Inglin B, Woollacott M. Age-related changes in anticipatory postural adjustments associated with arm movements. *J Gerontol Med Sci*. 1988;43:M105–113.
7. Tinetti ME. Performance assessment of mobility problems in elderly patients. *J Am Geriatr Soc*. 1986;34:119–126.
8. Kaya BK, Krebs DE, Riley PO. Dynamic stability in elders: momentum control in locomotor ADL. *J Gerontol Med Sci*. 1998;53A:M126–M134.
9. Alexander NB, Schultz A, Warwick DN. Rising from a chair: effects of age and functional ability and performance biomechanics. *J Gerontol Med Sci*. 1991;46:M91–M93.
10. Mourey F, Pozzo T, Rouhier-Marcer I, et al. A kinematic comparison between elderly and young subjects during standing up from and sitting down to a chair. *Age Ageing*. 1998;27:137–146.
11. Grishin AA, Formal'sky AM, Lensky A, et al. Dynamic walking of a vehicle with two telescopic legs controlled by two drives. *Int J Robot Res*. 1994;13:137–147.
12. Winter DA. *Biomechanics and Motor Control of Human Movement.* New York: Wiley; 1990.
13. Pai YC, Patton J. Centre of mass velocity-position predictions for balance control. *J Biomech*. 1997;30:347–354.
14. Teasdale N, Stelmach GE, Breunig A, et al. Age differences in visual sensory integration. *Exp Brain*. 1991;85:691–696.
15. Alexander NB. Postural control in older adults. *J Am Geriatr Soc*. 1994;42:93–108.
16. Horak FB, Shupert CL, Mirka A. Components of postural dyscontrol in the elderly: a review. *Neurobiol Aging*. 1989;10:727–738.
17. Schenkman M, Hughes MA, Samsa G, et al. The relative importance of strength and balance in chair rise by functionally impaired older individuals. *J Am Geriatr Soc*. 1996;44:1441–1446.

Received August 11, 1998
Accepted February 17, 2000
Decision Editor: Jay Roberts, PhD