Biomechanical Analysis of Older Adults Stepping Up: A Method of Evaluating Balance

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The aim of this study was to analyze differences in biomechanical parameters between elderly and control participants when stepping up, to evaluate control of balance. Eleven control and 14 elderly participants performed a step from an initial static posture onto a 7-cm-high force plate. For the spontaneous-velocity condition, elderly participants performed a slower progression velocity than control participants. Elderly participants spent proportionally more time in stance phase, with a corresponding decrease in swing phase, than the control participants, irrespective of movement velocity. In contrast, at spontaneous velocity the parameters related to ground-reaction force (GRF) showed that anteroposterior and mediolateral forces at toe-off of the support limb and the slope of vertical force during weight transfer were significantly smaller for the elderly than for control participants. These GRF parameters depended on the stepping-up velocity. The elderly develop a spatiotemporal strategy and reduced movement velocity to control support balance.

Keywords: postural balance, elderly, ground-reaction forces, force plate

Falls in the elderly are one of the most important public health problems for health care services in the 21st century. As part of a fall-prevention strategy, it would be of interest to evaluate balance regularly to detect risk of falls. Clinical tests such as the timed get-up-and-go (Mathias, Nayak, & Isaacs, 1986), the Berg Balance Scale (Shumway-Cook, Baldwin, Polissar, & Gruber, 1997), and the Tinetti Balance Scale (Tinetti, Speechley, & Ginter, 1988) are used regularly to evaluate movement and thus dynamic balance.

All of these tests are descriptive, however, and thus cannot explain the motor process related to any underlying degradation of balance and the concomitant increase in fall risk. Although the incidence of falls is well known to increase with age, the motor mechanisms related to this phenomenon remain poorly understood. Therefore, it is necessary to develop a biomechanical analysis to
enhance understanding of the mechanisms of dynamic balance control, as well as the degradation in balance with age.

To test dynamic balance it is necessary to disturb a person’s static balance and measure his or her compensatory response to control balance. Such a perturbation can be performed by individuals themselves by initiating voluntary movement from an initial static posture. Indeed, initiating a movement from a static posture requires the application of ground-reaction forces (GRFs), called propulsive forces, which constitute a source of perturbation for postural balance that must be controlled by the nervous system. To successfully perform a movement, the nervous system must control the destabilizing effect generated by GRF generation; that is to say that control of dynamic balance must be involved (Bouisset, 1991).

Control of dynamic balance can be affected by environmental constraints. For instance, when stepping up from a quiet stance position, one generates greater vertical GRF during the double-stance phase, in which body weight (BW) was transferred from one leg to another, than during level walking (Begg & Sparrow, 2000; Gelat & Brenière, 2000; Zachazewski, Riley, & Krebs, 1993). Such a movement therefore constitutes a perturbation to balance. The population group that is the most vulnerable to accidents while climbing steps has been identified as the elderly, who have a corresponding increase in the incidence of falls when weight is transferred onto the step (Archea, 1985; Hamel, Okita, Bus, & Cavanagh, 2005). Thus, a possible movement to study for a dynamic home-based test would be stepping up onto a higher level.

A number of studies of whole-body dynamics have reported changes in the pattern of stepping up from a stationary posture for elderly participants (Buckley, Heasley, Scally, & Elliott, 2005; Heasley, Buckley, Scally, Twigg, & Elliott, 2004). These studies, however, analyzed only the parts of the movement from the initial quiet stance through to the swing phase after the preparatory phase (anticipatory postural adjustment; Buckley et al.; Heasley et al.). Given the difficulties encountered by elderly participants during the weight-transfer phase (Begg & Sparrow, 2000), it would be of interest to incorporate an analysis of this phase.

Therefore, the main purpose of the current study was to investigate differences in biomechanical parameters between elderly and control participants when they performed a stepping-up movement from quiet stance to extract an indicator of dynamic postural balance and an indication of the capacity to generate GRFs. This preliminary study aimed to find parameters that could be used to assist with fall prediction in a clinical setting. Given that stepping up requires the generation of GRFs that perturb postural balance, the hypothesis of this study was that elderly participants would have decreased GRFs relative to control participants, to minimize postural perturbation.

**Methods**

**Participants**

Eleven control (CT) participants (7 women and 4 men) of mean age 36 ± 10 years and mean height 1.69 ± 0.06 m and 14 elderly (EL) participants (8 women and 6 men) of mean age 74 ± 1.5 years and mean height 1.66 ± 0.07 m participated in the study. The experiments were conducted in accordance with the Declaration
of Helsinki, and all participants gave their informed consent and completed an
in-house questionnaire to ensure that they had no musculoskeletal or neurological
conditions that precluded their participation in the study. No participants reported
a previous history of falls.

Procedure

The single-step protocol of Dietrich and Brenière (1994) was used in the current
investigation. This protocol required participants to perform a single step onto a
new level starting from a stable position, ending with both feet on the step (Buckley
et al., 2005; Dietrich & Brenière; Heasley et al., 2004). This protocol imposed two
major constraints on the participants: First, it was necessary to generate antero-
posterior, mediolateral, and vertical GRFs to propel the body forward and upward.
Second, braking forces needed to be generated to maintain as stable as possible a
position after the stepping-up movement. Because the generation of GRF consti-
tutes a perturbation of balance, the protocol is well adapted for testing capacity to
control postural balance.

Two force plates were used in the study (see Materials section). Participants
started from a stable upright position on the first force plate before stepping 7 cm
onto the second force plate (the second force plate was used as the step) on verbal
instruction, thereafter remaining as stable as possible until being asked to descend
(Figure 1). The height of the force plate of 7 cm was chosen because it was equiva-
 lent to the height of an average roadside curb (Buckley et al., 2005; Heasley et al.,
2004), an obstacle routinely negotiated in daily life.

The first limb to come in contact with the first force plate during the stepping-
up movement will henceforth be referred to as the leading limb (stepping limb),
and the limb that remained on the ground will be referred to as the supporting limb
(stance limb).

For both CT and EL participants, the movement was performed at a spontaneous
velocity, with participants free to choose the distance between the starting position
and the edge of the force plate. To control for the effects of velocity, CT participants

![Figure 1](image)

**Figure 1** — Experimental protocol. FC1 = first foot contact on the step; FO2 = second foot
off the ground; FC2 = second foot contact on the step; WT = weight-transfer phase; SW =
swing phase; dTOTAL = total stepping-up movement.
were also tested at a self-determined slow velocity. Participants were required to use the same foot position for all subsequent trials. They were barefoot and kept their arms at their sides during the movement. To ensure that visual information did not influence balance, participants were required to look down at a cross placed centrally 4 cm from the back of the second force plate.

Materials

Two AMTI OR6-5 force plates measuring 50.8 by 46.4 cm were used to collect GRF and moment data (Advanced Mechanical Technology Inc., Watertown, MA). Data acquisition was performed using Protags software (version 4.0, J-Y. Hogrel, Institute of Myology, Paris, France) developed using Labview (Version 7.0, National Instruments, Austin, TX). Data were sampled at 100 Hz using a PCI-MIO16E1 DAQ card (National Instruments, Austin, TX) with an eighth-order Butterworth low-pass filter and a cut-off frequency of 10 Hz. The data collected consisted of anteroposterior, mediolateral, and vertical forces (\( R_x, R_y, \) and \( R_z \), respectively) and anteroposterior, mediolateral, and vertical moments (\( M_x, M_y, \) and \( M_z \), respectively), which were subsequently processed before parameter calculation.

Extraction of all parameters was performed using Matlab (Version 6.0, The MathWorks Inc., Natick, MA). The location of all temporal events identified was performed automatically, as follows.

The moment at which the first foot contacted the second force plate (FC1) was detected using the \( R_z \) signal of the second force plate. The mean and the standard deviation of \( R_z \) without any force on the force plate were calculated to determine baseline noise, with a threshold corresponding to 6 times the standard deviation taken to indicate FC1. The instant when the second foot left the ground (FO2) was calculated using the first peak of the \( R_z \) signal measured with the second force plate. This point, at which time all of the participant’s BW was directed through the second force plate, was validated using the first force plate, whereby FO2 was considered to occur when no signal was measured.

The instant that the second foot touched the force plate (FC2) was identified by using a previously validated technique (Brenière, Do, & Sanchez, 1981; Dietrich & Brenière, 1994; Gelat & Brenière, 2000; Ledebt, Bril, & Brenière, 1998) with the mediolateral center of pressure. FC2 was considered to occur when the mediolateral center of pressure from the second force plate underwent a lateral displacement after FO2.

The times at which FC1, FO2, and FC2 occurred were subsequently detected automatically by programs written in Matlab. The results of the algorithms were validated using manual (visual) detection by an expert for 8 participants chosen at random. Comparison of the two values revealed errors to be 0.03 ± 0.03, 0.04 ± 0.02, and 0.04 ± 0.03 s for FC1, FO2, and FC2, respectively.

Concerning FC1 detection, the threshold value of 6 times the standard deviation was chosen empirically based on a comparison with FC1 values chosen by an expert. This validation indicated that 6 times the standard deviation gave the closest result to that of an expert for the validation trail using data from 8 participants. Indeed, the mean error for FC1, which was 0.03 ± 0.03 s, was considered acceptable for voluntary movement detection. Obviously such a technique is only suited to data in the current study, given their empirical nature.
Parameter Calculation

Given that one of the applications of the study was to develop a simple test for use on a single force plate, parameters extracted from the second force plate were essentially used for comparison between groups. The first force plate was used to validate the temporal events calculated using the second force plate and to provide velocity and stride-length parameters. The maximum anteroposterior center-of-mass velocity and the estimated step length were calculated using the two force plates, as follows: These two parameters were used to determine whether any subsequent differences observed between EL and CT participants were related to velocity or stride length. The maximum anteroposterior center-of-mass velocity was calculated by summing the anteroposterior GRFs of both force plates. The signal thus obtained was divided by the body mass of the participant to give the anteroposterior acceleration of the center of mass. An integration of the anteroposterior acceleration of the center of mass was then used to calculate the anteroposterior velocity of the center of mass, the peak value of which was taken as the velocity of stepping up onto the force plate.

The anteroposterior position of the center of pressure was measured before movement initiation while the participant was in a static position on the first force plate. It was also measured when the participant was completely stabilized on the second force plate, after FC2. The difference between the two positions provides an estimation of a participant’s step length.

Parameters calculated from the second force plate included the durations of the different phases of the step (Figure 2): total movement duration (dTOTAL), delimited between FC1 and FC2; weight-transfer-phase duration (WT), delimited between FC1 and the second toe-off from the ground (FO2); swing-phase duration (SW), delimited between FO2 and FC2; weight-transfer-phase duration, expressed as a percentage of dTOTAL (%WT); and swing-phase duration, expressed as a percentage of dTOTAL (%SW).

The slope of the vertical GRFs (loading rate) and impulse at toe-off were calculated from GRF and phase durations as follows: During the weight-transfer phase, the slope of the vertical GRFs (LR) normalized by BW was measured between FC1 and FO2. This variable represented the rate of loading (Stacoff, Diezi, Luder, Stussi, & Kramers-de Quervain, 2005) and has been thought to describe the “intensity” with which force is developed at impact (Nigg & Morlock, 1987). It was calculated as $LR = \frac{(Rz_{\text{FO2}} - Rz_{\text{FC1}})}{(\text{FO2} - \text{FC1})}$, where $Rz_{\text{FC1}}$ and $Rz_{\text{FO2}}$ were the vertical GRFs normalized by BW measured at FC1 and FO2, respectively (Figure 2).

The anteroposterior, mediolateral, and vertical impulse variables ($I_x_{\text{FO2}}, I_y_{\text{FO2}}, I_z_{\text{FO2}}$) extracted from the GRFs were measured at FO2. They were calculated as simple integrations of the GRF normalized by the weight of the participant.

The anteroposterior, mediolateral, and vertical forces normalized by participant weight were measured at FO2 ($Rx_{\text{FO2}}, Ry_{\text{FO2}}, Rz_{\text{FO2}}$).

Statistical Analyses

Statistical analyses were performed with the Statistical Package for Social Sciences (SPSS Inc., Chicago, IL). Data were checked for normality using the Kolmogorov–Smirnov test. All data were retained for subsequent statistical analysis.
Figure 2 — Biomechanical data obtained from a force plate for a typical adult participant during stepping up. $R_x$, $R_y$, and $R_z$ = ground-reaction forces normalized to participant body weight for anteroposterior, mediolateral, and vertical directions, respectively; $Y_p$ = center-of-pressure displacement for mediolateral direction; $LR$ = slope of vertical force; $FC_1$ = first foot contact on the step; $FO_2$ = second foot off the ground; $FC_2$ = second foot contact on the step; $WT$ = weight-transfer phase; $SW$ = swing phase; $dT_{TOTAL}$ = total movement; $f$ = forward; $b$ = backward; $l$ = left; $r$ = right; $u$ = up; $d$ = down.
Multivariate analysis of variance (MANOVA) was used to compare results between CT and EL participants at both spontaneous and comparable movement velocity, with participant group as the independent variable. The dependent variables were the following biomechanical parameters: maximum anteroposterior center-of-mass velocity; estimated step length; dTOTAL; WT; SW; %WT; %SW; LR; anteroposterior, mediolateral, and vertical impulse variables measured at FO2 ($I_{xFO2}$, $I_{yFO2}$, $I_{zFO2}$); and anteroposterior, mediolateral, and vertical force measured at FO2 ($Rx_{FO2}$, $Ry_{FO2}$, $Rz_{FO2}$). A $p$ value less than .05 was considered an indication of statistical significance.

Results

GRF Traces

The GRFs recorded under the leading limb, which was the first to contact the force plate, can be seen in Figure 2. At FC1, the anteroposterior-GRF trace progresses toward negative values, demonstrating that a braking GRF was generated. Examination of the mediolateral GRF showed that at FC1 GRFs were directed to the leading limb and at FC2 to the contralateral limb. After FC1, the vertical-GRF trace moves toward positive values, reaching a peak at FO2, thus demonstrating upward GRF generation. After FO2, the trace continues progressively toward the baseline, with all the oscillations gradually decreasing after FC2.

Maximum Anteroposterior Velocity of the Center of Mass

For both CT and EL participants, the trace of the anteroposterior velocity of the center of mass (Figure 3) progresses toward positive values, thus demonstrating the forward movement of the participant. The trace reached a maximum value after FC1, whereupon the trace progressed toward the baseline. The reduction of the velocity comes from the braking forces generated by the supporting limb at FC1. When participants took their second foot off the first force plate (FO2), velocity continued to decrease until FC2, whereupon velocity values tended toward zero, thus demonstrating that participants had a stable posture on the second force plate.

For the spontaneous-velocity condition, EL had a slower center-of-mass velocity than CT (0.30 ± 0.07 and 0.46 ± 0.05 m/s, respectively; $p < .001$). When CT participants adopted a self-determined slow velocity, no significant difference was detected between EL and CT groups (0.30 ± 0.07 and 0.30 ± 0.05 m/s, respectively; $p = .392$).

Spatiotemporal Variables

For the spontaneous-velocity condition, no significant differences were detected for dTOTAL between the CT and EL participants ($p = .061$). The duration of the WT, however, was longer for EL than for CT ($p < .001$), whereas there were no significant differences in SW between participant groups ($p = .122$; Figure 4).

In contrast, at comparable velocity, dTOTAL was shorter for EL than for CT ($p < .001$). Consequently, the durations of the WT and the SW were shorter for EL than for CT ($p < .001$ for both comparisons).
Figure 3 — Anteroposterior center-of-mass velocity for (a) a typical control participant and (b) a typical elderly participant during stepping up at spontaneous velocity. FC1 = first foot contact on the step; FO2 = second foot off the ground; FC2 = second foot contact on the step; Vm = maximum anteroposterior center-of-mass velocity.

When phase durations were expressed as percentages of dTOTAL, significant differences were found between CT and EL participants for both SW and WT durations (Figure 5). Indeed, for both spontaneous- and comparable-velocity conditions, %WT was longer for EL than for CT ($p < .001$ for both comparisons). Thus, %SW was shorter for EL than for CT ($p < .001$ for both comparisons).
In keeping with these results, the estimated step length was shorter for EL than for CT for spontaneous- and comparable-velocity conditions ($p = .005$ and $p = .003$, respectively; Figure 6).

At spontaneous velocity, anteroposterior ($R_{x_{FO2}}$) and mediolateral ($R_{y_{FO2}}$) force measured at FO2 were lower for EL than for CT ($p = .001$ and $p = .002$, respectively;
Figure 6 — Estimated step length for control and elderly participants. CT (SP) = control participants, spontaneous velocity; CT (SL) = control participants, slow velocity; EL (SP) = elderly participants, spontaneous velocity. *Significantly different from control participants at spontaneous-velocity condition ($p < .05$). ‡Significantly different from control participants at slow-velocity condition ($p < .05$).

Figure 7 — Ground-reaction force normalized by body weight for control and elderly participants. CT (SP) = control participants, spontaneous velocity; CT (SL) = control participants, slow velocity; EL (SP) = elderly participants, spontaneous velocity; $R_{x_{FO2}}$, $R_{y_{FO2}}$ = ground-reaction forces at FO2 for anteroposterior and mediolateral directions, respectively. *Significantly different from control participants at spontaneous-velocity condition ($p < .05$). ‡Significantly different from control participants at slow-velocity condition ($p < .05$).

Figure 7). In contrast, for comparable velocity, no change was measured for anteroposterior force ($R_{x_{FO2}}$) between the two groups ($p = .06$), but mediolateral force ($R_{y_{FO2}}$) was higher for EL than for CT ($p = .002$; Figure 7). No differences were observed in vertical force measured at FO2 ($R_{z_{FO2}}$) for the spontaneous- (0.995 ± 0.003 BW and 0.993 ± 0.004 BW for EL and CT, respectively; $p = .198$) and comparable-velocity conditions (0.995 ± 0.003 BW and 0.980 ± 0.040 BW for EL and CT, respectively; $p = .320$).
Figure 8 — Estimation of loading rate when stepping up. $R_z$ = vertical force normalized to body weight for control and elderly participants; LR = slope of vertical force in control (LR CT) and elderly (LR EL) participants; FC1 = first foot contact, FO2 = second foot off the ground; CT SP = control participants, spontaneous velocity; CT SL = control participants, slow velocity; EL SP = elderly participants, spontaneous velocity.

Figure 9 — Impulse at FO2 for control and elderly participants. CT (SP) = control participants, spontaneous velocity; CT (SL) = control participants, slow velocity; EL (SP) = elderly participants, spontaneous velocity; $I_{x_{FO2}}, I_{y_{FO2}}, I_{z_{FO2}}$ = impulse normalized by body weight for anteroposterior, mediolateral, and vertical directions, respectively. *Significantly different from control participants ($p < .05$).
At spontaneous velocity, LR was lower for EL than for CT (0.013 ± 0.003 and 0.020 ± 0.006 BW/s, respectively; \( p < .001 \); Figure 8). In contrast, at comparable velocity, LR was higher for EL than for CT (0.013 ± 0.003 and 0.009 ± 0.001 BW/s, respectively; \( p < .001 \); Figure 8), as would be expected given that the ST duration was longer in EL than in CT.

Concerning the impulse results, there were no significant differences in anteroposterior (\( I_x^{FO2} \)) and mediolateral (\( I_y^{FO2} \)) impulse at FO2 between EL and CT participants (\( p = .669 \) and \( p = .573 \), respectively; Figure 9) at spontaneous velocity. In contrast, vertical impulse (\( I_z^{FO2} \)) was significantly higher for EL than for CT (\( p < .001 \); Figure 9).

When impulses were measured at comparable velocity, however, anteroposterior (\( I_x^{FO2} \)), mediolateral (\( I_y^{FO2} \)), and vertical impulses (\( I_z^{FO2} \)) at FO2 were lower for EL and CT participants (\( p < .001, p = .02, \) and \( p < .001 \), respectively; Figure 6).

**Discussion**

Stepping up onto a higher level requires the generation of greater GRFs during the WT phase than those generated for level walking (Gelat & Brenière, 2000). Given that the generation of propulsive forces constitutes a perturbation of postural balance (Bouisset, 1991), the WT phase during stepping up would require greater control of postural balance to successfully perform the movement. In the elderly, functional changes in the sensory-motor (Brocklehurst, Robertson, & James-Groom, 1982; Kenshalo, 1986; Skinner, Barrack, & Cook, 1984) and neurological systems (Stelmach, Zelaznik, & Lowe, 1990; Teasdale, Stelmach, Breunig, & Meeuwse, 1991) alter the capacity to generate GRF and maintain postural balance. The principal aim of the study was to assess the biomechanical changes measured in elderly participants to control balance and successfully perform the stepping-up movement.

In agreement with the literature on locomotion (Ferrandez, Pailhous, & Durup, 1990; Menz, Lord, & Fitzpatrick, 2003; Murray, Kory, & Clarkson, 1969), the results of the current study showed that the spontaneous velocity generated by EL participants was slower than for CT participants. The relative durations %WT and %SW and estimated step length were, respectively, longer, shorter, and shorter for EL than for CT. When these spatiotemporal parameters were analyzed at comparable velocity, the differences between EL and CT participants remained the same.

A number of studies have analyzed locomotor patterns in elderly relative to young-adult participants (Ferrandez et al., 1990; Menz et al., 2003; Murray et al., 1969). In all cases, a relationship between spatiotemporal parameters and velocity was demonstrated. More precisely, step length and relative stance-phase duration were respectively reduced and lengthened when gait velocity diminished (Ferrandez et al.). The authors concluded that the gait of elderly participants was normal but was adapted to their velocity (Ferrandez et al.).

In contrast, the current study showed that %ST, %SW, and step-length results did not depend on stepping-up velocity. During the WT phase, participants were required to generate anteroposterior and vertical propulsive forces to break balance and initiate forward movement and to generate vertical propulsive forces to perform weight transfer. Generating propulsive forces imposed a constraint on balance that EL needed to deal with (Bouisset, 1991). Furthermore, during this
phase, EL needed to prepare for FO2, which imposed a constraint on balance. This can be confirmed by the greater $R_y^{\text{FO2}}$ reached at comparable velocity for EL, thus demonstrating a greater mediolateral disequilibrium in EL. Indeed, FO2 created disturbing propulsive forces that were associated with the reduction of the postural support base. Furthermore, after toe-off, the center of mass was located behind the limb that was on the ground, that is, behind the center of pressure (Lee & Chou, 2006). This dissociation between center of mass and center of pressure created a backward disequilibrium.

The changes in %ST, %SW, and step length between EL and CT were linked to an alteration of mechanisms developed to control postural balance. Therefore, to compensate for this degradation, EL reorganized the motor pattern of stepping up; that is, they developed a new spatiotemporal strategy relative to CT to favor balance. EL lengthened %ST duration to counterbalance the balance perturbations related to the WT phase, while reducing %SW duration to minimize the time spent with a single lower limb support, when control of postural balance is precarious. With a reduced step length the position of the center of mass will be closer to the leading limb, thus minimizing the backward disequilibrium at FO2.

Anteroposterior and vertical accelerations measured under the leading limb demonstrated that braking and upward GRFs were created by both CT and EL participants to comply with the protocol, which required a step up from an initial quasi-static posture, followed by a period of stable posture.

In the spontaneous-velocity condition, EL participants had a lower $R_x^{\text{FO2}}$ and LR than did CT participants. In contrast, when parameters were measured at comparable velocity, comparable $R_x^{\text{FO2}}$ and higher LR values were measured in EL.

In respect to spontaneous velocity, the lower $R_x^{\text{FO2}}$ and LR for EL participants indicated that for a given instance or a given time period, the generation of braking forces and upward GRFs was lower in EL participants than in CT participants. In contrast, for a comparable velocity, generation of braking forces was comparable between the two groups, and upward GRF was higher in EL than in CT participants. Therefore, these parameters depended on stepping-up velocity, with the lower $R_x^{\text{FO2}}$ and LR a result of the slower velocity spontaneously adopted by EL.

A number of studies have claimed that the lower velocity spontaneously adopted by elderly participants constitutes an adaptation frequently adopted by the elderly (Ferrandez et al., 1990; Lajoie, Teasdale, Bard, & Fleury, 1996; Menz et al., 2003; Murray et al., 1969). Such a strategy might provide a greater number of adaptive solutions related to the capacity to generate forces, control dynamic balance (Ferrandez et al.), or increase reaction time (Lajoie et al.).

In the literature, a number of studies have identified a decrease in muscle mass (Frontera et al., 2000) and a corresponding decrease in lower limb muscle force (Frontera et al.; Vandervoort, 2002) and muscle power (Herman et al., 2005; Whipple, Wolfson, & Amerman, 1987) with increasing age. Given that the modulation of GRF intensity is linked to a variation in muscle force, any alteration in muscle force would have repercussions on the capacity of the individual to create GRFs (Bouisset, 2002). It is probable that the muscle capacity of the knee and ankle extensors that are used to support and propel the body against gravity and to generate GRFs to climb from one step to the next would be altered (McFadyen & Winter, 1988; Nadeau, McFadyen, & Malouin, 2003). Consequently, EL participants would develop a lower velocity because their muscle capacity would not
allow sufficient braking-forces $R_{x_{FO2}}$ to be developed to successfully step up. It is probable that if the stepping-up velocity were increased, a second step would be performed by elderly participants to stop the movement and thus maintain a stable posture after FC2.

The successful performance of a movement from a static posture was linked not only to muscle capacity to generate propulsive force but also to the capacity to control the disturbing effect produced by propulsive-force generation (Bouisset, 1991). In previous studies, lower GRFs have been interpreted in terms of an alteration in muscle capacity. Such results can also be discussed, however, in terms of the control of postural balance.

In a study in which slower gait velocity was reported in the elderly, Murray et al. (1969) concluded that the elderly have a guarded walking style to remain stable. Similar conclusions have been proposed by Menz et al. (2003), who claimed that the lower velocity represents the characteristically cautious gait pattern of the elderly. Furthermore, in the clinical domain, gait velocity is widely used to evaluate balance (Cromwell & Newton, 2004; Rogers, Cromwell, & Newton, 2005). In people with balance problems, gait was significantly slower than for healthy participants (Protas et al., 2005), and improvement in gait velocity was associated with a decreased risk of falling (Montero-Odasso et al., 2005).

At FO2 the reduced postural base imposes an additional difficulty in controlling postural balance. A lower $R_{x_{FO2}}$, $R_{y_{FO2}}$, and LR minimized the disruptive effects imposed on postural balance at FO2. Therefore, by using a slower velocity, EL could reduce $R_{x_{FO2}}$, $R_{y_{FO2}}$, and LR, thus reducing the degree of perturbation to balance. In such a manner, it was possible to perform the movement successfully, despite the altered mechanism of postural balance.

At comparable velocity, the higher $R_{y_{FO2}}$ and LR were not in contradiction with the proposition that EL adopt a spontaneously lower velocity to reduce the intensity of the perturbation to balance. In addition, the increase in mediolateral acceleration ($R_{y_{FO2}}$) at comparable velocity for EL compared with CT indicates a more precarious state of balance than CT at FO2. It would be expected that should EL perform the movement at a higher velocity than normally developed, intensity of the perturbation to balance in the mediolateral axis would be increased and they would be at greater risk of falling. With respect to LR, the change was simply a result of the shorter WT directly resulting from the shorter dTOTAL in EL relative to CT.

Because of muscle alteration or an altered mechanism of balance control, EL participants were less able than CT participants to generate braking and upward GRFs. Such an alteration meant that EL had difficulty stopping the forward displacement of the body when stepping up and were forced lengthen the duration of braking- and upward-GRF generation to step up and stay as stable as possible on the force plate.

In consequence, the impulses $I_{x_{FO2}}$ and $I_{y_{FO2}}$ were comparable between the two groups, whereas $I_{z_{FO2}}$ was higher for EL than for CT. Previous studies have shown a correlation between vertical impulse and running performance, with more economical runners having a lower impulse (Heise & Martin, 2001). It was suggested that the increased time taken to perform the task, thus providing a greater impulse, is indicative of a greater requirement for muscle force (Smith, Dyson,
The finding of the current study that EL participants had an increased $I_{FO2}$ would suggest that they developed greater total muscle force, thus requiring more energy to produce the same vertical performance as CT participants. In keeping with these results, impulse for the anteroposterior ($I_{FO2}$) directions, although comparable between groups, was related to an inferior performance in EL participants. This finding was most likely a result of the longer time needed to reach a lower $Rx_{FO2}$ and $Ry_{FO2}$ for elderly participants. Thus, the impulses measured for vertical and anteroposterior axes demonstrated a less economical stepping-up movement for EL than for CT participants.

**Conclusion**

The independence of the relative durations of the stepping-up phase and step length from velocity showed that elderly participants developed a new spatiotemporal strategy to successfully perform the stepping-up movement. At spontaneous velocity, EL participants performed the stepping-up movement more slowly than did CT participants, with the associated GRF parameters interpreted as changes favoring balance. These parameters could be used as an indicator of both muscle capacities and postural balance in an elderly population. From a technological standpoint, the automatic calculation of these parameters from a force-plate-based device could provide a complementary test for clinical use.

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