Energy cost of walking and gait instability in healthy 65- and 80-yr-olds

Davide Malatesta,1 David Simar,2 Yves Dauvilliers,1,3 Robin Candau,2 Fabio Borrani,2 Christian Préfaut,1 and Corinne Caillaud2

1Unite Propre de Recherche de l’Enseignement Supérieur (UPRES) 701, Physiologie des Interactions, Hôpital Arnaud de Villeneuve, 34295 Montpellier; 2UPRES 2991, Sport Performance et Santé, Faculté des Sciences du Sport, 34090 Montpellier; and 3Service de Neurologie B, Hôpital Gui de Chauliac, 34293 Montpellier, France

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IN NORMAL SUBJECTS, the energy cost of level walking (Cw), i.e., the energy expenditure per distance unit, depends on the speed. The speed-Cw relationship follows a U-shaped curve, showing a minimum for an optimal speed of ~1.3 m/s (4, 23, 25, 27). Several studies have reported higher Cw in elderly subjects at several walking speeds. The speed-Cw relationship is thus shifted upward in the elderly compared with young subjects (21, 23, 24). The mechanisms involved in this adjustment remain unclear, but locomotor impairment such as gait instability may be involved, as was suggested by Allor et al. (1) for a population of adolescent girls. Although the relationship between gait instability and Cw has never been investigated in healthy elderly subjects, Hausdorff and colleagues (13, 14) showed greater gait instability in these subjects than in young subjects.

These authors further demonstrated that gait instability can be assessed by stride-to-stride changes in the gait cycle duration, i.e., the stride time (13, 14). Stride time reflects the walking rhythm and represents the final output of the locomotor system (14). The stride-to-stride changes in stride time can be described with respect to the fluctuation magnitude (e.g., the variance, the size of the fluctuation) and the fluctuation dynamics (how one stride changes from the next, independently of the variance) (13, 14). Both of these aspects can be used to identify gait instability, but the clinical relevance may be different. In healthy elderly subjects, the fluctuation magnitude is similar to that observed in healthy young adults, although there are age-related changes in the fluctuation dynamics (13). In this study, we hypothesized that the greater gait instability observed in older adults, even in the absence of overt disease, contributes to the higher Cw.

To further investigate the effect of gait instability on Cw, the energy expenditure specifically related to balance maintenance can be distinguished from that related to walking movements. Hoffman et al. (20) used a three-component model derived from Workman and Armstrong (33) to accomplish this in subjects with bilateral above-the-knee amputation presenting both instability in walking and higher Cw. This model differentiates the metabolic cost of walking into three compartments: basal metabolic rate (compartment 1), metabolic cost associated with balance preservation (compartment 2), and metabolic cost of walking movements (compartment 3). Hoffman et al. (20) showed that the higher Cw in amputees was accounted for in great part by the additional energy needed for the walking movement itself, but they also demonstrated...
that more energy is expended to maintain balance during walking. This model thus offers an interesting tool for investigation of the relationship between gait instability and $C_w$ in the elderly during walking. We hypothesized that greater gait instability is associated with greater energy expenditure to maintain balance.

The first purpose of this study was to determine whether greater gait instability is involved in the higher energy cost of walking in healthy elderly subjects. We therefore measured $C_w$ on a treadmill at different walking speeds, assessed gait instability by the fluctuation magnitude and fluctuation dynamics of stride time, and studied the relationships among these variables in healthy subjects aged 25, 65, and 80 yr. Our second objective was to determine whether greater gait instability is associated with greater energy expenditure to maintain balance while walking, using compartment 2 of Hoffman’s model.

METHODS

Subjects

Three groups of subjects participated in the study: G80 ($n = 10$; 81.6 ± 3.3 yr, range 77–88 yr), G65 ($n = 10$; 65.3 ± 2.6, range 60–69 yr), and G25 ($n = 10$; 24.6 ± 2.6, range 20–29 yr). The older subjects (G80 and G65) were recruited from associations that offer different activities for the elderly, including regular country walks, gymnastics, and cultural activities. The subjects were healthy, were living independently, and went on country walks once a week, at least 9 mo per year. Potential subjects had been screened before testing to eliminate those with medical problems that might affect gait, balance, or energy cost of walking. A neurologist, who was aware of the study objectives, extensively screened each older subject with a complete medical history, a physical examination, and the Mini Mental State Examination (8). A score below 26 on this test was an exclusion criterion, because diminished function may result in difficulties following instructions. All subjects were medically healthy and free of clinically significant orthopedic, neurological, cardiovascular, or respiratory problems. In addition, the postural balance of the healthy elderly subjects was measured with the Berg Balance Test, a validated balance scale for the elderly, and all subjects obtained a score above 47 (53.2 ± 3.1 and 56 ± 0.0 for G80 and G65, respectively), indicating good balance (3). None of the participants was taking medication with known major effects on balance or gait, such as benzodiazepines, neuroleptics, and antidepressants. However, seven subjects were taking drugs for hypertension (2 from G65 and 5 from G80) and two for hypercholesterolemia (1 in each elderly group), four were being treated with aspirin (1 G65 and 3 G80), and three women followed estrogen/progestin replacement therapy (G65). The young sedentary subjects (G25) were healthy and had not previously engaged in any form of regular physical exercise. Groups were matched according to gender, height, leg length, body mass, lean body mass, percent body fat, and basal metabolic rate. The protocol and the consent form were approved by the local ethics committee, and all subjects provided informed, written consent.

Experimental Design

Each subject completed two test sessions. In the first session, a physician took the medical history and performed a physical examination, and the subject was then introduced to the experimental procedures. Muscle function was assessed, and each subject was familiarized with treadmill walking. After 30 min of treadmill accommodation (23) across experimental walking speeds and a brief rest period, the preferred walking speed of the subject was determined according to the procedure proposed by Martin et al. (23). For the second session 1 wk later, each subject returned early in the morning for measurement of basal metabolic rate and estimation of body composition. Each subject completed the resting measurement in standing position and then performed six 6-min treadmill walking trials at six different speeds, separated by 5-min resting periods. The data from these trials were used to determine individual $C_w$–speed relationships and to build Hoffman’s three-compartment model (20).

Assessments

Estimation of body composition. Each subject rested in a supine position for 30 min and was then asked to empty his or her bladder. Four-electrode multifrequency bioelectrical impedance analysis was then performed on the nondominant side following the methodology for total body measurements (22). Impedance was measured at four frequencies: 1, 5, 50, and 100 Hz, by use of a Human-Im Scan (Dieto-System, Milan, Italy) and adhesive electrodes (Ag/AgCl, Paris, France). We then calculated the fat-free body mass by using the bioelectrical impedance analysis equation of Deurenberg et al. (7).

Physical activity level. The physical activity of the older subjects (G80, G65) was estimated from a physical activity questionnaire for the elderly (30). Different scores were used to quantify household activities, sports activities, and other physically active leisure time activities, altogether resulting in a total activity score. The questionnaire provides a method for classifying elderly subjects into categories of high, medium, and low physical activity, with cutoff points of 9.4 and 16.5 as proposed by Voorrips et al. (30). The scores have been shown to be positively associated with repeated 24-h activity recalls, pedometer measurements, and test-retest reliability (30).

Leg length. With subjects in standing position, the examiner measured the distance between the great trochanter and the ground for the right leg.

Muscle function. The force of a maximum voluntary isometric contraction (MVC) for right and left knee extensor muscles was measured by using the one-leg-extension machine (Banc de Koch, Genin Medical). A single tester encouraged each subject to exert maximal effort, and MVC force was measured in newtons with a dynamometer (Salter model 235, Testut). The subjects were in a seated position with a 90° angle at the hip and knee joints. To minimize the influence of upper limbs, all subjects crossed their arms over the chest. At least three attempts were allowed, and the highest for each leg was retained as right or left MVC (2).

Basal metabolic rate. The subjects were asked to abstain from tea and coffee and to avoid any physical activity for 1 day. The following morning, they came to the laboratory after an overnight fast and before any food intake and rested quietly in the supine position for 30 min, breathing with a mask. Expired gas, ventilation, and heat rate were measured continuously during this time.

Preferred walking speed. After treadmill accommodation across experimental walking speeds (0.67, 0.89, 1.11, 1.33, 1.56 m/s) and a brief rest period, each subject’s preferred walking speed was determined according to the methodology proposed by Martin et al. (23). Briefly, each subject began treadmill walking at the lowest experimental speed (0.67 m/s), which was then slowly increased until the subject subjectively identified his or her preferred walking speed.
This speed was maintained for 1 min and was then slightly modified. The subject was again asked to evaluate the speed and adjustments were made according to subject directive. This procedure was repeated starting with the highest experimental speed (1.56 m/s) and gradually reducing to the preferred speed. The final preferred walking speed was considered to be the mean of the two speeds selected by the subject during both the increasing and decreasing speed trials.

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Fluctuation magnitude. Before any calculation, the first 15 s of the recorded data were excluded to minimize any start-up effects, and a median filter was then applied to remove the data points that were 3 SDs above or below the median value. The average stride time was determined for each time series, and stride time variability was defined as the stride time coefficient of variation (CV), where the variability is normalized to each subject’s mean stride time (CV = 100 × SD/mean) (14).

Fluctuation dynamics. To quantify the temporal “structure” of each time series (independently of overall variance), we calculated three parameters of the stride time fluctuation dynamics. The first two parameters, calculated for each time series, were the “nonstationary index” and the “inconsistency of the variance” (12, 14, 17). To minimize the effects of any differences in mean or variance, each time series was first normalized with respect to its mean and SD, yielding a new time series with mean = 0 and SD = 1, but with different time scales. The slope and intercepts from these equations (the a and b coefficients for the equation y = ax² + b, where y is the VO2 corresponding to x) were then calculated for each time series as described by Hausdorff et al. (13, 15, 16). DFA is a modified random walk analysis that can be used to quantify the long-range, fractal properties of a relatively long time series (15). Briefly, the root-mean-square fluctuations of the integrated and detrended time series are calculated at different time scales, and the slope of the relation between the fluctuation and the time scale determines a fractal scaling index (α). For a process in which the value of one stride is completely uncorrelated with any previous value (white noise), α equals 0.5. We calculated α over the region 10 ≤ n ≤ 20 (n is the number of strides in the window of observation). This region was chosen because it has been shown to provide a statistically robust estimate of stride time correlation properties that are most independent of finite size effects (length of data) (12, 13).

Statistical Analysis

A one-factor analysis of variance (ANOVA) was used to determine differences in the descriptive characteristics (i.e., height, leg length, body mass, lean body mass, percent body fat, basal metabolic rate, standing VO2, preferred walking speed, physical activity questionnaire score, and MVC) among the groups. A two-factor ANOVA was used to determine the effects of age and speed of walking on mean VO2 and energy cost. Linear regression analyses of VO2 vs. the squared speed were performed on the data of each subject. The slopes and intercepts from these equations (the a and b coefficients for the equation y = ax² + b, where y is the VO2 and x is the speed) and compartment 2 (VO2 corresponding to VO2 minus basal metabolic rate VO2) were then compared among groups with one-factor ANOVA. Concerning each ANOVA, when global difference was identified, Tukey’s
post hoc analyses were performed. When the assumption or the equality of variance was violated, an ANOVA (Kruskal-Wallis) for nonparametric values was used.

A Kruskal-Wallis test was used to identify differences in gait analysis parameters [stride time coefficient of variation, nonstationary index, inconsistency of the variance, and the fractal scaling index (a)] among the three groups. If this test showed significant group differences, Mann-Whitney U-tests were performed to compare two groups at a time. These nonparametric tests make no assumption about the underlying distribution of the data being compared. Correlations between gait instability measures and age, physical activity score in the elderly subjects, which were performed to compare two groups at a time.

RESULTS

Subject Characteristics

There were no significant differences in height, leg length, body mass, lean body mass, percent body fat, and basal metabolic rate among groups (Table 1). A significant effect of age (P = 0.02) was observed for the physical activity score in the elderly subjects, which was lower in G80 (12.63 ± 4.54) than in G65 (18.93 ± 6.3). According to the standard previously established (30), G80 and G65 were classified in the medium and high physical activity categories, respectively.

Preferred walking speed was lower in G80 (1.14 ± 0.12 m/s) than in G65 (1.35 ± 0.08 m/s) and G25 (1.31 ± 0.06 m/s) (P < 0.001 and P = 0.001, respectively).

Energy Expenditure of Walking (V\(_\text{O}_2\) and C\(_w\))

Preexercise standing V\(_\text{O}_2\) values were similar among groups (P = 0.49). This result was consistent with the lack of statistical difference in lean body mass. V\(_\text{O}_2\) was significantly higher in G80 than in G25 for all walking speeds, whereas G65 differed from G25 for only two speeds (1.33 and 1.56 m/s; P = 0.03 and P = 0.02, respectively) (Fig. 1). A significant intragroup speed effect was observed for V\(_\text{O}_2\) (P < 0.01). All subjects exhibited respiratory exchange ratio <1 for all walking speeds, indicating a negligible participation of anaerobic metabolism in the energy cost.

A significant effect of age was observed for C\(_w\) (P < 0.01) (Fig. 2). This parameter was significantly higher in G80 than in G25 for all walking speeds, and mean C\(_w\) was 22% higher in G80. As for V\(_\text{O}_2\), C\(_w\) was higher in G65 compared with G25 for two walking speeds (1.33 and 1.56 m/s; P = 0.03 and P = 0.02, respectively). A significant intragroup speed effect was observed (P < 0.01), with the exception of the fifth speed compared with the third and fourth speeds.

C\(_w\) at preferred walking speed was higher in G80 (0.229 ± 0.03 ml O\(_2\)·kg\(^{-1}\)·m\(^{-1}\)) and G65 (0.205 ± 0.02 ml O\(_2\)·kg\(^{-1}\)·m\(^{-1}\)) than in G25 (0.179 ± 0.02 ml O\(_2\)·kg\(^{-1}\)·m\(^{-1}\)) (P < 0.001 and P = 0.002, respectively).

Three-compartment model. The individual linear regressions of V\(_\text{O}_2\) against squared speed showed coefficients of determination (r\(^2\)) ranging from 0.92 to 0.99 for all subjects. The P value for these coefficients of determination was <0.05 in all cases. Table 2 summarizes the equation y = ax^2 + b, where y is the V\(_\text{O}_2\) (in ml·kg\(^{-1}\)·min\(^{-1}\)) and x is the walking speed (in m/s), with mean ± SD values for coefficient a and b, for the three groups. The a coefficient, the actual metabolic cost related to walking movements (compartment 3), was significantly higher in G80 and G65 than in G25 (P = 0.04 and P = 0.03, respectively). The b coefficient, the metabolic cost related to basal metabolic rate (compartment 1) and to maintaining balance while walking.

Table 1. Subject characteristics

<table>
<thead>
<tr>
<th>Variable</th>
<th>G80</th>
<th>G65</th>
<th>G25</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gender</td>
<td>F 3 M</td>
<td>F 3 M</td>
<td>F 3 M</td>
</tr>
<tr>
<td>Age, yr</td>
<td>81.6 ± 3.3†</td>
<td>65.3 ± 2.5‡</td>
<td>24.6 ± 2.6</td>
</tr>
<tr>
<td>Height, m</td>
<td>1.61 ± 0.1</td>
<td>1.62 ± 0.1</td>
<td>1.64 ± 0.07</td>
</tr>
<tr>
<td>Leg length, m</td>
<td>0.89 ± 0.05</td>
<td>0.84 ± 0.06</td>
<td>0.85 ± 0.04</td>
</tr>
<tr>
<td>Body mass, kg</td>
<td>60 ± 12.9</td>
<td>60.7 ± 14.9</td>
<td>58.9 ± 8.6</td>
</tr>
<tr>
<td>Lean body mass, kg</td>
<td>44 ± 10</td>
<td>46.2 ± 12.5</td>
<td>45.4 ± 5.8</td>
</tr>
<tr>
<td>Percent body fat, %</td>
<td>26.4 ± 6.9</td>
<td>24.2 ± 6.6</td>
<td>23.7 ± 7.1</td>
</tr>
<tr>
<td>Basal metabolic rate, ml·kg(^{-1})·min(^{-1})</td>
<td>2.8 ± 0.6</td>
<td>2.6 ± 0.3</td>
<td>2.6 ± 0.3</td>
</tr>
<tr>
<td>Standing oxygen uptake, ml·kg(^{-1})·min(^{-1})</td>
<td>6.4 ± 1.32</td>
<td>5.85 ± 0.79</td>
<td>6.05 ± 0.53</td>
</tr>
<tr>
<td>Physical activity score</td>
<td>12.63 ± 4.54</td>
<td>18.93 ± 6.3</td>
<td>22.17 ± 9.2‡</td>
</tr>
<tr>
<td>MVC (right), N</td>
<td>221.7 ± 99.2</td>
<td>265.4 ± 68.3</td>
<td>253.2 ± 102.4</td>
</tr>
<tr>
<td>MVC (left), N</td>
<td>186.4 ± 75.4</td>
<td>244.3 ± 60.9</td>
<td>326.7 ± 79.4</td>
</tr>
</tbody>
</table>

Values are means ± SD. F, female; M, male; MVC, maximum voluntary isometric contraction. †Significant difference between 80-yr-old group (G80) and young controls (G25) (P < 0.05). ‡Significant difference between G80 and 65-yr-old group (G65) (P < 0.05). §Significant difference between G65 and G25 (P < 0.05).
Table 2. Three-compartment model

<table>
<thead>
<tr>
<th>Compartment 1 (standing)</th>
<th>G80</th>
<th>G65</th>
<th>G25</th>
</tr>
</thead>
<tbody>
<tr>
<td>a Coefficient (ml (\text{O}_2)/kg (\cdot)min (^{-1}))</td>
<td>4.10 ± 0.16*</td>
<td>4.06 ± 0.85†</td>
<td>3.31 ± 0.58</td>
</tr>
</tbody>
</table>
| b Coefficient (kg \(\cdot\)
\(\text{m}^2\)/kg \(\cdot\)min \(^{-1}\)) | 10.37 ± 1.82* | 9.1 ± 1.22 | 8.51 ± 0.93 |

Values are means ± SD. The constant \(a\) defines compartment 3, and the constant \(b\) is the sum of compartments 1 and 2; coefficients of the model equation \(y = ax^2 + b\), where \(y\) is oxygen uptake in ml \(\cdot\)kg \(\cdot\)min \(^{-1}\) and \(x\) is the speed in m/s for G80 \([y = 4.10 (±0.16)x^2 + 10.37 (±1.82)]\), G65 \([y = 4.06 (±0.85)x^2 + 9.1 (±1.22)]\), and G25 \([y = 3.31 (±0.58)x^2 + 8.51 (±0.93)]\). *Significant difference between G80 and G25 (P < 0.05).†Significant difference in the constant \(b\) between G80 and G65 (P = 0.007). Compartment 2 was significantly higher (28.8%; P < 0.01) for G80 than for G25 (P = 0.007), whereas there was no significant difference between G65 and G25 (P = 0.17; Table 2). A graphical display of this model for the G80 subjects is shown in Fig. 3. The coefficient \(a\) was directly correlated with the energy cost of walking at the five imposed speeds (\(r = 0.54, r = 0.64, r = 0.68, r = 0.83, r = 0.91\), respectively; P < 0.05) and at preferred walking speed (\(r = 0.65; P < 0.05\)).

Table 3. Gait instability indexes

<table>
<thead>
<tr>
<th>Gait instability index</th>
<th>G80</th>
<th>G65</th>
<th>G25</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride time variability (% change)</td>
<td>2.31 ± 0.68*</td>
<td>1.93 ± 0.39†</td>
<td>1.40 ± 0.30</td>
</tr>
<tr>
<td>Coefficient of variation (% change)</td>
<td>0.79 ± 0.13</td>
<td>0.69 ± 0.18</td>
<td>0.66 ± 0.10</td>
</tr>
<tr>
<td>Inconsistency of variance</td>
<td>0.28 ± 0.05</td>
<td>0.31 ± 0.08</td>
<td>0.30 ± 0.04</td>
</tr>
<tr>
<td>Fractal scaling index ((\alpha))</td>
<td>0.68 ± 0.17</td>
<td>0.78 ± 0.22</td>
<td>0.78 ± 0.17</td>
</tr>
</tbody>
</table>

Values are means ± SD. *Significant difference between G80 (\(n = 8\)) and G25 (\(n = 8\)) (P < 0.05).†Significant difference between G65 (\(n = 10\)) and G25 (P < 0.05).

Alternate Three-Compartment Model

An alternate three-compartment model was defined by substituting the basal metabolic rate in compartment 1 with standing \(\dot{V}_{O_2}\), as follows: 1) compartment 1 became the metabolic cost of standing; 2) the corresponding new compartment 2 became the metabolic cost associated with maintaining balance during walking based on the difference between the \(\dot{V}_{O_2}\) equivalent to coefficient \(b\) and the \(\dot{V}_{O_2}\) of standing; and 3) compartment 3 remained the metabolic cost associated with walking movements (Fig. 6). The correlation between the new compartment 2 and the stride time variability almost reached statistical significance (\(r = 0.35, P = 0.07\)).

Muscle Function

Right and left MVC were significantly lower in G80 than in G25 (P = 0.009 and P < 0.001, respectively). No significant difference in MVC was observed between G65 and G25 (Table 1). The average MVC of the right and left limbs of the three groups was inversely correlated with the energy cost of walking at the five imposed speeds (\(r = −0.49, r = −0.52, r = −0.50, r = −0.54, r = −0.59\), for 0.67, 0.89, 1.11, 1.33, 1.56 m/s, respectively; P < 0.01) and at preferred walking speed (\(r = −0.49; P = 0.009\) (Fig. 7)).
DISCUSSION

The main finding of the present study was that healthy octogenarians exhibited higher $C_w$ and greater stride time variability. However, no significant correlation was found between the indexes of gait instability and $C_w$. The increase in energy cost is probably multifactorial, and our results suggest that gait instability is probably not the main contributing factor in our subjects. The $C_w$ was respectively 22 and 12% higher in G80 and G65 compared with G25. Previous reports have shown that the $C_w$ for a comparable speed is generally higher for healthy elderly subjects, particularly those above 65 yr, compared with young subjects (23, 24, 31). Although our results corroborate these findings, we found a greater difference in $C_w$ between young and older people than described in the literature (23, 31). This is likely related to the fact that our three groups were matched in terms of fat and lean body mass, in contrast to the groups of previous studies. The carefully matched groups are a unique aspect of the present study and were considered to be a necessary condition for accurately delineating the effect of age on walking mechanics and economy, particularly because fitness may also explain the additional energy expenditure for treadmill walking in older men (25). To facilitate group matching, we recruited healthy and active elderly subjects, as shown by the results from the physical activity questionnaire, and all our elderly subjects (G80, G65) were regular recreational walkers. These healthy older subjects may not be representative of the general elderly population, but this selectivity was necessary to study the effect of normal aging without serious pathology on the relationship between $C_w$ and gait instability.

The stride time coefficient of variation was significantly higher in the healthy elderly subjects than in the young subjects, whereas the parameters associated with the fluctuation dynamics of stride time (nonstationary index, inconsistency of variance, and fractal scaling index) did not significantly differ among the three groups. These findings contrast with previous results (9, 11) that showed that the stride time coefficient of variation was very similar in healthy elderly subjects compared with young and middle-aged subjects and that the fractal scaling index ($\alpha$) was significantly lower for the elderly subjects ($\alpha = 0.68 \pm 0.14$) than for the young subjects ($\alpha = 0.87 \pm 0.15$) (13). These differences may be explained by the modality of walking because the subjects in the previous studies walked on level ground, whereas our subjects walked on a treadmill. Treadmill locomotion at imposed speed may have reduced the possibility of free regulation of stride, which is possible in level ground walking, and therefore caused a loss of gait automaticity in our young subjects (10). Previous work has shown that with external pacing (e.g., via a metronome), the fractal scaling index breaks down and the fluctuations in gait become more random (16). Indeed, the G25 subjects in the present study had a lower $\alpha$ coefficient than the young subjects in Hausdorff’s experiments. However, it should be noted that the $\alpha$ coefficient calculated in G80 was both lower than in G25 and closer to 0.5, suggesting that stride-interval fluctuations were more random in the elderly subjects, as previously shown by Hausdorff et al. (13). On the other hand, the significant difference in the stride time coefficient of variation between the elderly (G80 and G65) and the young subjects clearly attests that the elderly subjects of this study were characterized by greater gait instability. This is important because previous reports have suggested that increased stride time variability is a sensitive measure for quantifying the gait instability of elderly fallers (11, 17). Moreover, this high variability in older adults has been associated with decreased
physiological capacity (e.g., reduced exercise capacity, dynamic balance, and functional reach) (14), also typical of “normal” aging. Our results showed that the stride time coefficient of variation was positively correlated with age and inversely correlated with physical activity level in the elderly subjects, and this latter finding, in agreement with others (14), indicates that the gait instability of the elderly is associated with functional status and the ability to perform the activities of daily living.

Several authors have separately investigated the effect of age on $C_w$ (21, 23, 24) and on stride time variability (9, 11, 13). To our knowledge, this study is the first to investigate the relationship between gait instability and $C_w$ in the healthy elderly. Recently, Allor et al. (1) suggested a relationship between gait instability and $C_w$ in adolescent girls after observing higher $C_w$ in the girls compared with matched young women during treadmill walking. Although mean stride length and frequency did not differ between groups, the authors suggested that the gait dynamics (although this was not investigated) would explain the observed difference in exercise economy. Hausdorff et al. (18) found that stride time variability decreased as a function of age in children. We thus hypothesized that gait instability may play a role in the higher $C_w$ in healthy elderly subjects. Greater gait instability may increase muscular contractions to stabilize walking, thus contributing to the extra energy expenditure of walking in this population. Compatible with this hypothesis, our results showed that the $G_{80}$ subjects, characterized by greater gait instability than $G_{25}$, had higher $C_w$. Yet despite these interesting results, we did not find a significant correlation between gait instability and the energy cost of walking at preferred walking speed, indicating that gait instability was not the main determinant of the higher energy cost of walking in the elderly. This might be explained by the relationship between energy expenditure and shortening velocity (the Fenn effect), which is such that, for a null shortening velocity (isometric condition), the energy expenditure is very low (19). The metabolic cost of the isometric muscular contractions involved in balance maintenance during walking may thus be negligible with respect to the metabolic cost of the concentric muscular actions associated with walking movements.

Our second objective was to assess the metabolic cost of these isometric muscle actions to maintain balance during walking using compartment 2 of Hoffmann’s model. Our results showed that compartment 2 was higher for $G_{80}$, a group of subjects also characterized by greater gait instability than $G_{25}$, but there was no significant correlation between these two variables. The lack of significant correlation might be related to the small sample size of this study, however, so we cannot exclude the possibility of a relationship between gait instability and compartment 2.

Although Hoffman et al. (20) used changes in compartment 2 to assert that amputees were more unstable than healthy subjects, no attempt was made to assess instability independently. Workman and Arm-strong (33), in the development of their theoretical model, suggested that maintaining posture while walking will cost more, metabolically speaking, than maintaining posture while standing still. The single support phase during gait makes balance maintenance more difficult and costly than the static balance with double support during standing. To better assess this difference in energy expenditure between static and dynamic balance, we propose substituting the metabolic basal rate in compartment 1 of the three-compartment model with the standing oxygen uptake. This amounts to a new compartmentalization of the model and to a new compartment 2, i.e., the metabolic cost associated with maintaining balance during walking, corresponding to the difference between the VO$_2$ equivalent to coefficient $b$ and the VO$_2$ of standing. In this study, the correlation between the new compartment 2 and the stride time variability almost reached statistical significance (see RESULTS). This energy expenditure seems to be better correlated with gait instability than compartment 2 of Hoffmann’s model and more related to the muscle contractions to maintain balance while walking.

The results also showed that compartment 3 was significantly higher for our elderly subjects ($G_{80}$ and $G_{65}$) than for the young subjects and correlated with $C_w$ at the experimental walking speeds. This suggests that the additional metabolic cost of walking movements in healthy elderly subjects is probably accounted for by an impaired exchange of potential and kinetic energy, leading to increased mechanical work. During walking, potential and kinetic energy change in opposite phases and the recovery of mechanical energy through pendular motion attains a maximum (~65%) at intermediate speeds (1.11–1.33 m/s). There is good agreement among the speeds at which maximum recovery, minimum external work per unit distance, and minimum energy expenditure all occur (6). In elderly subjects, the increased mechanical work of walking may be associated with excessive activation of the antagonist muscles during agonist action, i.e., coactivation, as in postural control (32). This coactivation of the lower limb muscles, which causes joint stiffening and limits the degrees of freedom needed for walking, partially explained the higher $C_w$ observed in children with cerebral palsy (29). This hypothesis may be considered as an alternative explanation for the greater energy cost observed in the elderly.

The present study also showed that the maximal isometric strength of the knee extensor muscles was inversely correlated with the energy cost of walking for both the young and old healthy subjects. This result agrees with the findings of Martin et al. (23), who suggested that the higher $C_w$ for a given speed in the elderly compared with young subjects was likely attributable to the higher cost of generating muscle force. These authors further suggested that the decreased muscle mass, strength, and force per cross-sectional area that occur with aging (26) forced the elderly subjects to recruit a greater number of motor units per muscle for a given task. Because walking at a given
speed requires a higher relative intensity for the elderly compared with young subjects (31), it is logical that a greater proportion of fast-twitch fibers, which have been shown to be less economical than slow-twitch fibers, (28) will be recruited. In addition, this greater proportion of fast-twitch fibers was likely recruited and forced to shorten at less than optimal velocity, i.e., the velocity at which maximum power is developed and efficiency is optimized (5). That this occurred was supported by the fact that the maximal isometric strength of the knee extensor muscles and the energy cost of walking were more correlated at faster speeds than at slower speeds. At faster walking speeds, the force developed by the knee extensor muscles increased and a greater proportion of fast-twitch fibers was thus recruited.

In summary, this study shows that both Cw and stride time variability are greater in octogenarians than in young adults. However, no relationship was found between these two variables, suggesting that gait instability is not the main explanation of the higher energy cost of walking in the elderly. Differences in compartment 3, which represents the energy expenditure associated with walking movements, and in the MVC of the knee extensors instead appear to be responsible for the higher energy cost of walking in the healthy elderly. Future research should examine the biomechanical and neuromuscular factors accounting for the differences in walking economy between healthy elderly and young subjects.

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