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

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Submitted 3 December 2002; accepted in final form 25 July 2003

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 optimal speed of 1.3 m/s (4, 23, 25, 27). Several studies have reported higher Cw in elderly subjects across the different walking speeds (P < 0.05). Stride time variability at preferred walking speed was significantly greater in G80 (2.31 ± 0.68%) and G65 (1.93 ± 0.39%) compared with G25 (1.40 ± 0.30%; P < 0.05). There was no significant correlation between gait instability and energy cost of walking at preferred walking speed. These findings demonstrated greater energy expenditure in healthy elderly subjects while walking and increased gait instability. However, no relationship was noted between these two variables. The increase in energy cost is probably multifactorial, and our results suggest that gait instability is probably not the main contributing factor in this population. We thus concluded that other mechanisms, such as the energy expenditure associated with walking movements and related to mechanical work, or neuromuscular factors, are more likely involved in the higher cost of walking in elderly people.

In this study, 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 77–86 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 difficulty 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 the healthy elderly subjects was measured with the Berg test. 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/progesterin 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 measurement (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 heart rate were measured continuously during this time.

**Preferred walking speed.** After treadmill acclimation across experimental walking speeds (0.67, 0.89, 1.11, 1.33, 1.56 m/s) subject’s preferred 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.

**C. and the three-compartment model.** After 4 min of rest without measurement, expired gases were collected and analyzed for 4 min in the standing position (20). The subjects were then asked to complete six 6-min level walking trials on the treadmill at five equally spaced speeds (0.67, 0.89, 1.11, 1.33, 1.56 m/s) and preferred walking speed, in randomized order. They were allowed to establish their own preferred stride rate combination for each condition and were given 5 min of rest between walking trials. All subjects walked on the treadmill (Gymrol 1800 Control, Tech Machine) without using handrail support but were secured by a cross-belt fixed to the handrails that did not impede arm swing. During the walking trials, oxygen uptake (\(\dot{V}O_2\)), \(CO_2\) output, and ventilatory equivalent for carbon dioxide were measured breath by breath (MGA-110 mass spectrometer, Marquette Electronics), and the calibration was checked before each trial with standard calibration gases. A 3-liter syringe was used to calibrate the volume turbine by using flow rates similar to subject ventilation. Metabolic data were averaged over 15-s intervals, and heart rate was monitored during the 6 min of walking (Cardioline Max-1, Marquette Electronics). \(\dot{V}O_2\) values from the last 2 min were averaged and normalized to body mass (ml \(O_2\) · kg \(^{-1}\) · min \(^{-1}\)). Data were subsequently normalized with respect to walking speed to produce the desired descriptor of \(\dot{C}_w\), i.e., aerobic demand per unit of distance walked (ml \(O_2\) · kg \(^{-1}\) · m \(^{-1}\)).

The data collected during the different walking trials were used to calculate the characteristics of the individual three-compartment models. Linear regression analyses of \(\dot{V}O_2\) vs. squared walking speed were performed with the data of each subject to obtain the slopes and intercepts from these equations (the \(a\) and \(b\) coefficients for the equation \(y = ax^2 + b\), where \(y\) is \(\dot{V}O_2\) in ml \(O_2\) · kg \(^{-1}\) · min \(^{-1}\) and \(x\) is the speed in m/s) (20). The model defines the three compartments as follows: **compartment 1** is the basal metabolic rate, **compartment 2** is the metabolic cost associated with maintaining balance, and **compartment 3** is the metabolic cost associated with walking movements. **Compartment 2** (i.e., index of energy expenditure to maintain balance while walking) is calculated by subtracting basal metabolic rate from \(b\), the ordinate intercept that theoretically defines the metabolic demand associated with the zero walking speed (20). This compartment also includes the metabolic cost associated with respiration and heart contraction during walking, which is relatively small. Workman and Armstrong (33) demonstrated that **compartment 2** remains almost constant across walking speeds. **Compartment 3** is described by the constant \(a\), the slope of the relation of metabolic cost of walking and speed according to the equation \(y = ax^2 + b\).

**Gait instability.** Walking rhythm was measured on the treadmill at preferred walking speed with pressure-sensitive insoles (Force Sensing Resistor model FSR 174) placed in the subjects’ shoes. These inserts measured the pressure applied to the ground during ambulation and were connected to a computer that digitized the data at 100 Hz. The stride time was determined as the time elapsed between the initial ground contact of one foot and its subsequent contact. This was determined for each stride of the 6-min walk, and a time series of stride times was generated for each subject. Both the magnitude and the dynamics of the fluctuations were then evaluated. Because gait dynamics are independent of speed (16), we chose to study the preferred walking speed, which has been shown to provide the lowest walking variability (34).

**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 variance properties. The local value of these equations (the \(a\) and \(b\) coefficients) changed with time. This time series was then divided into blocks of five strides each, and in each segment the (local) average and the (local) SD were computed. The nonstationary index, defined as the SD of the local averages, was then calculated to estimate the dispersion of these normalized, local means. The inconsistency of the variance, the SD of the local SDs, was calculated to evaluate how the local SD changed with time (14). A block of five strides was chosen because it was both small enough to be unaffected by fatigue, changes in intention, or external changes and big enough to contain some stride-to-stride variability.

A third measure of the fluctuation dynamics, detrended fluctuation analysis (DFA), was applied to 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 (\(\alpha\)). For a process in which the value of one stride is completely uncorrelated with any previous value (white noise), \(\alpha\) equals 0.5. We calculated \(\alpha\) over the region \(10 \leq n \leq 20\) (\(n\) 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 \(\dot{V}O_2\), 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 \(\dot{V}O_2\) and energy cost. Linear regression analyses of \(\dot{V}O_2\) 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 \(y = ax^2 + b\), where \(y\) is the \(\dot{V}O_2\) and \(x\) is the speed) and **compartment 2** (\(\dot{V}O_2\) corresponding to intercepts minus basal metabolic rate \(\dot{V}O_2\)) were then compared among groups with one-factor ANOVA. Concerning each ANOVA, when global difference was identified, Tukey's
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>7 F, 3 M</td>
<td>7 F, 3 M</td>
<td>7 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 ± 8.8</td>
</tr>
<tr>
<td>Percent body fat, %</td>
<td>26.4 ± 6.9</td>
<td>24.2 ± 6.6</td>
<td>23 ± 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>10.12 ± 5.4‡</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. †Female; ‡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 65-yr-old group (G65) and G25. §Significant difference between G80 and G25. ¶Significant difference between 80-yr-old group (G80) and young controls (G25) (P < 0.05).
Hoffman model and stride time variability at preferred walking speed ($r = 0.24, P = 0.23$) (Fig. 5).

### 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 correspondingly 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 ($r = -0.49, r = -0.52, r = -0.50, r = -0.54$, and $r = -0.59$, 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).

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### Table 2. Three-compartment model

<table>
<thead>
<tr>
<th></th>
<th>G80</th>
<th>G65</th>
<th>G25</th>
</tr>
</thead>
<tbody>
<tr>
<td>$a$ Coefficient</td>
<td>$4.10 \pm 1.06^a$</td>
<td>$4.06 \pm 0.85^f$</td>
<td>$3.31 \pm 0.58$</td>
</tr>
<tr>
<td>$b$ Coefficient</td>
<td>$10.37 \pm 1.82^a$</td>
<td>$9.1 \pm 1.22$</td>
<td>$8.51 \pm 0.93$</td>
</tr>
<tr>
<td>Compartment 2, ml $\dot{O}_2$ kg$^{-1}$ min$^{-1}$</td>
<td>$7.58 \pm 1.45^a$</td>
<td>$6.5 \pm 1.05$</td>
<td>$5.9 \pm 1.05$</td>
</tr>
</tbody>
</table>

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 kg$^{-1}$ min$^{-1}$ and $x$ is the speed in m/s for G80 [$y = 4.10 \pm (1.06)x^2 + 10.37 \pm (1.82)$], G65 [$y = 4.06 \pm (0.85)x^2 + 9.1 \pm (1.22)$] and G25 [$y = 3.31 \pm (0.58)x^2 + 8.51 \pm (0.93)$]. *Significant difference between G80 and G25 ($P < 0.05$).

### Table 3. Gait instability indexes

<table>
<thead>
<tr>
<th></th>
<th>G80</th>
<th>G65</th>
<th>G25</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride time variability coefficient of variation, %</td>
<td>$2.31 \pm 0.68^a$</td>
<td>$1.93 \pm 0.39^f$</td>
<td>$1.40 \pm 0.30$</td>
</tr>
<tr>
<td>Nonstationary index</td>
<td>$0.79 \pm 0.13$</td>
<td>$0.69 \pm 0.18$</td>
<td>$0.66 \pm 0.10$</td>
</tr>
<tr>
<td>Inconsistency of variance</td>
<td>$0.28 \pm 0.05$</td>
<td>$0.31 \pm 0.08$</td>
<td>$0.30 \pm 0.04$</td>
</tr>
<tr>
<td>Fractal scaling index ($\alpha$)</td>
<td>$0.68 \pm 0.17$</td>
<td>$0.78 \pm 0.22$</td>
<td>$0.78 \pm 0.17$</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 ($P < 0.05$).

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**Fig. 3.** Three-compartment model for G80. **Compartment 1**: basal metabolic rate. **Compartment 2**: metabolic cost associated with walking movements. **Compartment 3**: metabolic cost associated with walking movements. Speed$^2$, squared walking speed.

**Fig. 4.** Correlation between the $C_w$ and stride time variability at preferred walking speed ($r = 0.19, P = 0.35$) for G80 ($n = 8$), G65 ($n = 10$), and G25 ($n = 8$). CV, coefficient of variation.
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 older fallers (11, 17). Moreover, this high variability in older adults has been associated with decreased...
higher $C_w$ in the girls compared with matched young

Yet despite these interesting results, we did

characterized by greater gait instability than G25, had

the gait instability of the elderly is associated with

ficient of variation was positively cor-

strive to recruit a greater number of motor units per

able 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 G80 subjects, characterized by greater gait instability than G25, 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 G80, a group of subjects also characterized by greater gait instability than G25, 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-
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 extensors 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.

We thank Jean-Paul Micalef and Didier Delignières for technical assistance, Nicola Maffuletti for helpful suggestions and criticism, the staff of the Service d’Exploration Fonctionnelle Respiratoire, the Université du Tiers Temps de Montpellier for recruiting efforts, and the subjects for their participation.

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