The objective of this study was to measure gait abnormalities in elderly fallers with the Locometrix™ gait-analysis system. This accelerometric device provided the following gait variables: walking speed and stride frequency, length, symmetry, and regularity. The variables were analyzed over a 20-s period of stable walking on a flat track of 40 m. Participants were 20 elderly patients hospitalized for falls (mean age 80.8 ± 5.0 years) and 33 older adults living at home (mean age 77.2 ± 6.5 years). All gait variables were found to be significantly lower in the faller group (p < .05). The lower gait speed, stride length, and stride frequency were previously recognized as nonrelevant in predicting the risk of falling, whereas lower stride symmetry was related to an underlying pathology and lower stride regularity was correlated to the risk of falls. The Locometrix appears to be well suited to measure gait regularity in routine practice.

Key Words: falls, gait analysis, aging

Falls in elderly adults represent a major health-care problem. About 30% of people over 65 years old living in the community fall at least once a year (Cumming, 1998). In addition, fall rate increases with age: It increases from 47% in people age 70–74 years to 94% among 80- to 84-year-olds (Campbell, Borrie, & Spears, 1989). Furthermore, the rate of growth of the older adult population is still increasing; age-adjusted incidence of fall-related injury has increased since 1970 without any obvious explanation (Kannus, Niemi, Palvanen, & Parkani, 1997). It is well established that a large proportion of falls occur during walking (Campbell et al.; Nickens, 1985; Wild, Nayak, & Isaacs, 1981), and the relationship between risk of fall and gait impairment has been demonstrated (Rubenstein, Robbins, Schulman, et al., 1988; Tinetti, Speechley, & Ginter, 1988).

Previous studies have demonstrated associations between changes in gait and falling in older adults (Maki, 1997). For example, older patients who fall tend to walk more slowly, to have a shorter stride length, and to spend a greater proportion...
of the gait cycle in double-leg support. These findings are common for gait characteristics in elderly people in general, however (Grabiner, 1997). More recently, other authors (Hausdorff, Elderberg, Mitchell, Goldberger, & Wei, 1997; Maki; Wolfson, Whipple, Amerman, & Tobin, 1990) have suggested that stride-to-stride variability might be a stronger predictor of falling. Because this variability cannot be discerned by visual evaluation, it is necessary to use a gait-analysis system; hence, a major aim is to make gait analysis possible in clinical practice (Whittle, 1995).

A new gait-analysis method was recently proposed for routine use (Auvinet, Chaleil, & Barrey, 1999a). This method is based on craniocaudal and mediolateral acceleration recordings at a point close to the body’s center of gravity. It provides relevant information such as stride frequency, symmetry, and regularity. The objective of our study was to use this gait-analysis system to measure gait-pattern irregularities associated with falls in the elderly. We compared two populations: One group included healthy older adults, and the other consisted of elderly fallers.

Method

PARTICIPANTS

**Controls.** The control group consisted of 33 healthy elderly participants (15 women and 18 men) recruited from the relatives of Laval Hospital’s employees. They had no history of musculoskeletal, neurological, or gait disorders; were living at home; and had good mobility, as well as normal activity levels. They took no more than three types of medicine and no psychotropic drugs. All of them were examined to check for marked pelvic asymmetry (none of our participants had a difference of more than 1 cm in leg length) and to exclude clinically patent scoliosis.

**Patients.** The faller group included 18 women and 2 men who had been hospitalized for recent falls. All patients were living at home, and their history of falls had lasted for several years (mean 3.4 ± 2.9), with an average of 3.6 ± 4.0 falls during the preceding year. All patients suffered from more than one pathology and took an average of 7.3 ± 3.5 drugs per day, including psychotropic drugs. Informed oral consent was obtained from both controls and patients. Demographic and morphological characteristics of the two groups are given in Table 1.

GAIT-ANALYSIS SYSTEM

The gait-analysis system (Locometrix™; Figure 1) included accelerometric sensors, a recording device, and a computer program for processing the accelerometric signals. The sensor is composed of two accelerometers placed perpendicularly to one another and housed in a molded box (40 × 18 × 18 mm). Its total weight is 20 g. The sensor is incorporated in a semielastic belt, which is fastened around the participant’s waist so that the sensor is over the L3–L4 intervertebral space (Figure 2). One accelerometer is aligned with the mediolateral axis of the body, and the other, with the craniocaudal axis. A portable data logger records signals from the sensor with an acquisition frequency of 50 Hz. This device can record continuously for 10 min. It weighs 140 g and is housed in a box measuring 65 × 22 × 12 mm. Recorded signals are transferred to a laptop computer, then formatted in files
Figure 1. The Locometrix™ gait-analysis system includes a sensor composed of two accelerometers and a portable recorder.

Table 1 Demographic and Morphological Characteristics in the Control and Faller Groups

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Controls, n = 33</th>
<th>Fallers, n = 20</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>M</td>
<td>SD</td>
<td>M</td>
</tr>
<tr>
<td>Age (years)</td>
<td>77.2</td>
<td>6.5</td>
<td>80.7</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>163.1</td>
<td>7.5</td>
<td>156.5</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>63.0</td>
<td>16.3</td>
<td>59.8</td>
</tr>
</tbody>
</table>

Note. n.s. = difference not significant at p > .05.

analyzed by software developed in the Matlab 5 environment (Scientific Software, Sèvres, France).

A period of stable walking was manually selected on the recording. The duration (20.48 s) corresponded to 1,024 points of acceleration measurements and provided an optimal variable-calculation time. In healthy participants, it corresponds to 19–21 gait cycles, that is, about 28 m. The selected period was used to calculate the following gait variables: stride frequency, symmetry, and regularity.

In normal walking, movements on the left and right sides are identical from one cycle to the next. Walking at a constant speed can therefore be considered the
sum of a series of periodic stationary movements. Using a fast Fourier transform, the analysis program derived the fundamental frequency of periodic movements, which corresponded to the frequency of the steps, from the craniocaudal acceleration signal. By definition, a stride includes two steps; thus, the stride frequency represented half the value of the fundamental frequency.

Stride symmetry and regularity are derived from two coefficients of correlation, C1 and C2, obtained by calculating the autocorrelation function on the vertical-acceleration signal. Stride symmetry is calculated as \((C1/C2) \times 100\) and expresses the similarity of craniocaudal movements on the left and on the right independently from fluctuations in successive craniocaudal movements of each limb. Stride regularity is calculated as \((C1 + C2) \times 100\) (score on 200) and expresses the similarity of vertical movements over time. To obtain a normal distribution and a linear measurement scale for stride regularity and symmetry, a Fischer Z transformation, 
\[
z(x) = 0.5 \log\left(\frac{1 + x}{1 - x}\right),
\]
was applied to the coefficients C1 and C2 (Spiegel, 1993).

**GAIT TEST**

The recording was carried out while the participant walked at his or her chosen speed down a 40-m straight hospital corridor and back. The 40-m distance was long enough to record a stable walking period between the start and the end of the test.
An electronic synchronized stopwatch measured the walking speed. Patients and controls wore their usual walking shoes; they were advised against high heels and hard-soled shoes.

**Statistical Analysis**

ANOVA tests were performed for each variable in both groups. The corresponding ROC (receiver operating characteristic) curves were then calculated by plotting sensitivity versus $(1 – \text{specificity})$. The most discriminating parameter has the highest area under the ROC curve, that is, the better compromise between sensitivity (ability to detect fallers) and specificity (ability to detect nonfallers). Statistical tests were done using NCSS 97 (Logilab).

**Results**

The results of the ANOVA revealed that all variables were significantly different between the faller group and the control group: walking speed and stride length, frequency, regularity, and symmetry. Descriptive statistics and statistical differences between the two groups are given in Table 2.

Faller patients walked more slowly than participants in the control group did: $0.73 \pm 0.22 \text{ m/s}$ versus $1.24 \pm 0.19 \text{ m/s}$. This lower walking speed was mainly a result of shorter stride length ($0.86 \pm 0.26 \text{ m}$ vs. $1.28 \pm 0.17 \text{ m}$), and lower stride frequency ($0.86 \pm 0.07$ vs. $0.97 \pm 0.08$) also contributed to this reduction. There was significantly lower symmetry in fallers than in the control group ($173.4 \pm 45.5$ vs. $210.9 \pm 39.4$), as well as considerably lower regularity ($191.3 \pm 56.0$ vs. $291.9 \pm 51.9$).

Sensitivity and specificity were evaluated by using ROC-curve analysis. We calculated the ROC curves for each gait variable using data from the control and patient groups. The area under the curve was high for all five parameters, establishing

<table>
<thead>
<tr>
<th>Variable</th>
<th>Fallers</th>
<th>Controls</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>M</td>
<td>SD</td>
<td>M</td>
<td>SD</td>
<td>p</td>
</tr>
<tr>
<td>Speed (m/s)</td>
<td>0.73</td>
<td>0.22</td>
<td>1.24</td>
<td>0.19</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Stride frequency (Hz)</td>
<td>0.86</td>
<td>0.07</td>
<td>0.97</td>
<td>0.08</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>0.86</td>
<td>0.26</td>
<td>1.28</td>
<td>0.17</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Stride symmetry $z$</td>
<td>173.4</td>
<td>45.5</td>
<td>210.9</td>
<td>39.4</td>
<td>&lt;.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stride regularity $z$</td>
<td>191.3</td>
<td>56.0</td>
<td>291.9</td>
<td>51.9</td>
<td>&lt;.001</td>
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</table>
the fact that there was a good compromise between sensitivity and specificity (Table 3); the ROC curves are presented in Figure 3.

**Discussion**

**FALLER AND CONTROL GROUPS**

The characteristics of the two groups differed greatly. The control group consisted of 33 elderly participants who were healthy and active, with an approximately equal number of men and women, whereas the faller group, with a long history of falls, included only 2 men versus 18 women. This was in accordance with the fact that
women fall more often than men do (Tinetti & Speechley, 1989). It was therefore not possible to study the influence of sex using the different variables. The faller group was older than the control group, but the mean age of each group was over 75 years.

GAIT-ANALYSIS SYSTEM

Recent technical advances have resulted in the development of numerous methods of gait analysis in both laboratory and ambulatory conditions. For practical applications, in order to evaluate the risk of fall in ambulatory conditions, two major constraints must be taken into account: a long walking test and information on gait regularity. It has been demonstrated that at least seven gait cycles are needed to obtain temporal data, but these results lack precision. At least 22 gait cycles are needed to obtain highly precise temporal-distance data (Kaufman, Chambers, & Sutherland, 1996). Thus, mainly two gait-analysis systems could be discussed: foot switches and accelerometry devices.

Using foot switches is a method of measuring routine gait-temporal parameters. Sensors can be included in modified shoes (Marey, 1894) or insoles (Hausdorff et al., 1997; Patterson, Gorman, & Wetzel, 1984) or attached to the sole of the foot (Blanc, Balmer, Landis, & Vingerhoets, 1999). Foot switches included in modified insoles are well adapted to routine examination for outpatients (Doutrellet, Durlent, Lagersie, et al., 1997; Peruchon, Jullian, & Rabischong, 1994). Some of these devices include 248 plantar-pressure sensors distributed uniformly in two insoles, allowing data recording at a frequency of 40 Hz over 25 s. In this way, it is possible to record a free walk of 35 m in a healthy individual. Such a system does not allow precise measurements of vertical force but gives clinically interesting information on foot-to-floor-contact evolution during stance phase (Doutrellet et al.). Nevertheless, if more information about the sequential foot rollover is required, foot switches attached to the sole of the foot appear to be more precise than are foot switches included in insoles (Blanc et al.). Both conditions allow the measurement of all spatiotemporal parameters over prolonged sequences, hence permitting the study of stable walking over a large enough number of cycles and providing satisfactory information on the variability of parameters such as velocity, stride length, and double-stance variation over time.

The recent development of a new generation of accelerometers suited for investigating lower frequency mechanical events has sparked renewed interest in their use in the analysis of walking and running (Bouten, Westerterp, Verduin, & Janssen, 1994). Accelerometry could be used to evaluate energy expenditure associated with walking in humans (Schultz, Sparti, & Herren, 1997) or quantify balance during human walking (Moe-Nilssen, 1998a). It has been suggested that accelerations in the middle of the lower back, over L3 (which is close to the center of gravity during walking), should be recorded (Auvinet, Chaleil, & Barrey, 1999b; Moe-Nilssen, 1998b).

Accelerometry has a number of advantages as a tool for gait analysis: It can provide gait analysis over a long temporal distance, and sensor position is easily reproducible in a given individual because it relies on simple anatomic landmarks and functional parameters such as the person’s waist. The information obtained reflects global motion at a point close to the body’s center of gravity. Mathematical
processing of the recorded signals supplies simple quantified variables, namely stride frequency, symmetry, and regularity. Because this noninvasive test takes little time and does not require a special environment, it is well suited for everyday clinical practice.

GAIT-ANALYSIS TEST

Data were collected over a distance of 40 m for both fallers and healthy older adults (the initiation and termination of the walk were excluded from data analysis). All patients and controls walked at their normal pace, conforming to the great majority of studies (Bendall, Nayak, & Pearson, 1989; Bohannon, 1997; D’Angeli-Chevassut & Gaviria, 1994; Hageman & Blanke, 1986; Larish, Martin, & Mungiole, 1987; Maki, 1997; Nigg, Fisher, & Ronsky, 1994; Okuzumi, Tanaka, Haishi, et al., 1995). At the self-determined pace in healthy people, walking variability is minimized (Yamasaki, Sasaki, Tsuzki, & Torri, 1984). A long-distance test strengthens the analysis of the interstride variability. We did not find any statistical differences between the there-and-back walking tests among any variables. This means that our test conditions allowed a real free-walking test without any effects from the distance or focused walking.

WALKING SPEED

Loss of speed is the main age-related change in walking (Bohannon, 1997; D’Angeli-Chevassut & Gaviria, 1994; Hageman & Blanke, 1986; Larish et al., 1987; Maki, 1997; Nigg et al., 1994; Okuzumi et al., 1995). Conflicting results have been published, however. Dahlstedt (1977) reviewed 15 articles with respect to walking speed in adults and found speeds ranging from 1.3 to 1.5 m/s for women and 1.3 to 1.6 m/s for men. One study found that in active, community-dwelling older adults with a mean age of 80 years the average speed is approximately 0.9–1.2 m/s (Murray, Kory, & Clarkson, 1969), whereas a study of unaided gait in ambulatory nursing-home residents reported an average speed of about 0.5 m/s (Wolfson et al., 1990).

It has been reported that loss of speed starts at age 60 (Costes-Salmon, Lafont, Dupui, et al., 1999) or 70 years (Judge, Ounpuu, & Davis, 1996). The magnitude of age-related reduction in gait velocity reported by Öberg, Karsznia, and Öberg (1993) varies between 0.1% and 0.7% per year. This decline in walking speed can rise to a rate of 12–16% after the seventh decade (Judge et al.), possibly even to a rate of 20% (Costes-Salmon et al.). Such contrasts among different studies might be the result of differences among the examined groups and varying gait-analysis methods (Öberg et al.).

Gait speed and stride length have been shown to be reduced in some but not all studies of elderly fallers (Hausdorff, Ladin, & Wei, 1995), so average gait speed is unlikely to be a successful predictor of falling in a moderately mobile older population (Maki, 1997). The loss of speed is primarily caused by decreased stride length (Murray et al., 1969; Winter, Patla, Franks, & Walt, 1990) and, eventually, decreased stride frequency (Costes-Salmon et al., 1999; D’Angeli-Chevassut & Gaviria, 1994). An unchanged stride frequency has been reported in advanced age (Judge et al., 1996; Öberg et al., 1993), but cycle frequency has been shown to be
negatively correlated with limb length (Murray et al.) and to be greater in women than in men (D’Angeli-Chevassut & Gaviria; Gabell & Nayak, 1984). We can observe lower stride frequency in our healthy older patients in comparison with our previous results obtained for a healthy adult patient group ($N=139$, mean age $= 41.4 \pm 11.2$ years, stride frequency $= 1.02 \pm 0.07$). Furthermore, the lower walking speed in the faller group is mainly a result of decreased stride length and is secondary to the decreased stride frequency.

GAIT SYMMETRY

There is general agreement that walking is symmetric in both genders and in all age groups (D’Angeli-Chevassut & Gaviria, 1994), in both spatial and temporal parameters (Costes-Salmon et al., 1999). Some authors have concluded that asymmetry is a good indicator of gait abnormalities (Blanc et al., 1999; Dewar & Judge, 1980; Giakas & Baltzopoulos, 1997).

Asymmetry has been described during free walking in healthy participants. For example, Maupas, Paysant, Martinet, and André (1999) found asymmetry in the total movement of the knees in the sagittal plane in 51% of young, healthy participants. This conflicting result could depend on the gait-analysis method and the different variables recorded. Previously, with our apparatus, the symmetry index was reported as a relevant variable to distinguish patients with hip or knee osteoarthritis from a control group (Auvinet et al., 1999a). The value of the asymmetry was correlated to the Lequesne functional index (Lequesne, Mery, Samson, & Gérard, 1987), so we can hypothesize that the low value of the symmetry index in the faller population is related to an underlying pathology such as osteoarthritis or neurological disease.

STRIDE REGULARITY

Since 1979 (Imms & Edholm, 1979), it has been reported that gait variability could be related to falls in the elderly. Variability in stride length in hospitalized fallers has already been reported (Guimaraes & Isaacs, 1980). Gabell and Nayak (1984) reported that spatial (step length and stride width) and temporal (stride time and double-support time) variabilities in an elderly adult are abnormal and have pathological etiology.

Wolfson, cited by Hausdorff et al. (1997), showed that gait unsteadiness can be used to identify people at risk of falling (a measure of inconsistency and of stepping). On the other hand, Feltner, MacRae, and McNitt-Gray (1994), using videotape analysis in a retrospective and prospective study in 17 community-dwelling older women, found that none of the kinematic variables distinguished the retrospective fallers from nonfallers, nor were they significant predictors of prospective falls.

Hausdorff et al. (1997), in a retrospective study, measured the variability of the temporal parameters of gait in community-dwelling elderly fallers during an extended walk (6 min) with a foot-switch system and found significantly more stride-to-stride temporal variations (stride time, stance time, swing time, and percent stance time) in gait in elderly fallers than in a control group. The stride variability in spatiotemporal gait parameters measured by foot switches could be an
independent predictor of the likelihood of experiencing a future fall, independent of fear of falling (Maki, 1997).

More recently, Costes-Salmon et al. (1999), using the kymographic method of Bessou, Duput, Montoya, and Pages (1988), suggested that the variability of swing time could be a predictor of falls among the elderly. Furthermore, risk of falls tended to be explained by means of 3D gait analysis (Kemoun, Benaim, Blatt, Thevenon, & Guieu, 1999) with modification of peak-torque values of different joints in fallers.

We concluded from our results that there was a lower regularity index in the faller group than in the control group. In addition, in a previous study with 139 healthy adult participants (mean age 41.4 ± 11.2 years) the regularity index was the same (329.7 ± 37.4) as in the control group of that study (Auvinet, Chaleil, & Barrey, 1999a). Such results had been previously reported by Hausdorff et al. (1997).

The calculation of our regularity index is different from the previous method of calculation described to measure variability of the gait. Our regularity index expresses the sum of two coefficients of correlation based on the autocorrelation analysis of the craniocaudal acceleration signal. This method simultaneously took into account both the acceleration pattern of each stride and the temporal variation between strides.

Conclusion

This new ambulatory gait-analysis system is well adapted for hospital outpatients. It provides useful information about gait variables such as walking speed, stride length, stride frequency, symmetry index, and regularity index. All these variables were highly different between the faller group and the control group. Some of these variables, however (walking speed, stride length, stride frequency), have been recognized by other studies as nonrelevant in predicting the risk of falling. The symmetry index appears to be related to underlying pathologies. The regularity index, which has been associated with the risk of fall, seems to be the most relevant variable for predicting falls risk. The Locometrix™ provides an easy way to measure gait regularity. The next step will be a prospective study to evaluate the role of these different variables in predicting falls risk.

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References


