Research Report

Gait Characteristics of Elderly People With a History of Falls: A Dynamic Approach

Background and Purpose. This study investigated changes in the kinematics of elderly people who experienced at least one fall 6 months prior to data collection. The authors hypothesized that, in order to decrease variability of walking, people with a history of falls would show different kinematic adaptations of their walking patterns compared with elderly people with no history of falls. **Subjects and Methods.** Twenty-one elderly people who had fallen within the previous 6 months ("fallers"; mean age=72.1 years, SD=4.9) and 27 elderly people with no history of falls ("nonfallers"; mean age=73.8 years, SD=6.4) walked at their preferred stride frequency (STF) as treadmill speed was gradually increased (from 0.18 m/s to 1.52 m/s) and then decreased in steps of 0.2 m/s. Gait parameter measurements were recorded, and statistical analysis was applied using walking speed and STF as independent variables. **Results.** Fifty-seven percent of the fallers were unable to walk at the fastest speed, whereas all nonfallers walked comfortably at all walking speeds. Although the fallers showed significantly greater STF, smaller stride lengths, smaller center-of-mass lateral sway, and smaller ankle plantar flexion and hip extension during push-off, they showed increased variability of kinematic measures in their coordination of walking compared with the nonfallers. **Discussion and Conclusion.** Although the fallers' adaptations were expected to reduce variability in the coordination of walking, they showed less stable gait patterns (ie, greater variability) compared with the nonfallers. Increased variability of walking patterns may be an important gait risk factor in elderly people with a history of falls. [Barak Y, Wagenaar RC, Holt KG. Gait characteristics of elderly people with a history of falls: a dynamic approach. *Phys Ther.* 2006;86:1501–1510.]

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Key Words: *Elderly, Falls, Flexibility, Kinematics, Variability.*

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Falls are a major public health concern, particularly in the elderly population.[1] Approximately 25% to 35% of people aged 65 years or older experience falls each year, and the epidemiology of falls shows that more than 50% of the falls occur during some form of locomotion.[2] Although significant changes in gait parameters may render one third of the community-dwelling elderly population at risk for falls,[3] most gait studies of elderly people focus on gait risks for falls in clinical populations[4] and often are limited to the evaluation of stride parameters.[5] For example, in individuals older than 70 years, average gait speed declines 12% to 16% per decade,[6,7] stride frequency (STF) increases,[6] stride length (STL) decreases at a given walking speed,[6,8,9] and double-support duration increases from 18% (in young people) to 26% (in elderly people).[5,6,10]

Possible explanations for changes in walking speed and stride parameters with aging include reduction of energy cost,[11–13] compensation for muscle weakness,[5,14–17] balance impairments,[18–20] and coping with increased variability in walking.[3,21–23] For example, the decreased STL in elderly people coincides with weakness in hip extensors and ankle plantar flexors, reduced push-off phase, increased swing phase, and a reduced ability to propel the body forward during gait.[6,24–27] Muscle weakness and impaired balance also appear to be associated with increased STF and double support duration.[24,28] Some of these adaptations in the walking patterns of elderly people may be related to falls prevention and fall history. Decreasing variability of walking patterns may help to prevent falls by achieving a more stable gait pattern, thus improving control of the whole-body position and momentum, and by reducing the mediolateral momentum of the center of mass.[3,29] Walking more slowly with a higher STF and shorter STL also may help to stabilize the gait pattern and allow greater adjustment and flexibility to changes in walking conditions (eg, increasing or decreasing walking speed, especially in individuals who experienced falls in their past).[3,22] One feature of gait that has been used to identify people who are at risk for falling is gait unsteadiness, which is a measure of inconsistency. That is, an unsteady gait pattern will be characterized with greater kinematics variability. Walking conditions such as walking speed could (gradually or abruptly) change. In this regard, *flexibility* is defined as the ability to adopt new movement patterns following changes in task requirements.[22,30]

The main goals of this study were to investigate the adaptability of elderly people who are at risk for falling to changes in walking conditions (ie, various walking speeds) and to assess their kinematic variability during changes in walking conditions. The basic premise was that the adopted gait pattern should be one that minimizes their unstableness (ie, decreases variability) evidenced in their history of falls. Although STL, STF, and speed adaptations may improve gait steadiness, the mechanisms are poorly understood. The literature reports that gait patterns typified by shorter STL or higher STF impose a shorter stance phase and reduced durations of center of mass traveling outside the base of support.[5,6,15,17] By lifting the leg faster into swing (ie, increased angular velocity of hip and knee flexion) as opposed to normal push-off, it might be argued that the forces necessary for forward progression are minimized. Therefore, these kinds of spatiotemporal adaptations will most likely diminish propulsive and mediolateral forces acting on the whole body during walking. If this is true, the changes in the gait patterns should support this premise. Thus, we predicted that, compared with elderly people with no history of falls, elderly people who are at risk for falls will show the following

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Dr Barak developed the data analysis strategy and conducted the data analysis, and he was the primary author of the manuscript. Dr Wagenaar was the initiator and supervisor of the study's design and settings and was fully involved in writing the manuscript. Dr Holt served as the biomechanics advisor for this work. The authors thank Chia-Ling Ho for her contribution to the experiments and the data analysis.

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Table 1.
Subject Characteristics[a]

Characteristic	Nonfallers (n=27)	Fallers (n=21)	Overall (N=48)	*P*
Age (y)				
$\overline{X}$	72.1	73.8	72.9	NS[b]
SD	4.9	6.4	5.6	
Height (m)				
$\overline{X}$	1.65	1.64	1.64	NS
SD	0.10	0.08	0.09	
Weight (kg)				
$\overline{X}$	69.2	74.7	71.95	NS
SD	13.5	13.7	13.6	

[a] Fallers=elderly people who experienced at least one fall in previous 6 months, nonfallers=elderly people with no history of falls.
[b] NS=no significant differences between groups at the .05 alpha level.

characteristics: (1) decreased ankle plantar flexion and hip extension during push-off, (2) increased hip flexion during the swing phase, and (3) reduced mediolateral sway. Despite these changes, we hypothesized that elderly people who are at risk for falls will still show inherent unsteadiness in their movement patterns compared with elderly people with no history of falls.

Method

Subjects

Data were collected from 21 elderly people (13 women, 8 men) who experienced at least one fall 6 months prior to data collection ("fallers") and from 27 elderly people (13 women, 14 men) with no history of falls ("nonfallers"). All subjects were screened before the start of the study by filling out a medical history questionnaire. The questionnaire addressed whether the subjects had a history of cardiopulmonary, musculoskeletal, somatosensory, or neurological disorders or severe visual and vestibular loss. If so, they were excluded from the study. All subjects gave informed consent to participate in the study. Subject characteristics are detailed in Table 1.

Design

Subjects walked comfortably on a treadmill, were free to swing their arms without any restraints, and were not allowed to hold on to the handrails of the treadmill. The instructions given to the subjects were: "At any given speed, please walk at your preferred stride frequency and do not hold the treadmill handrails unless you feel unsafe or have a fear of falling." The speed of the belt was gradually increased from 0.18 m/s to 1.52 m/s, in increments of 0.225 m/s, and then decreased in similar steps. Each speed condition was maintained for 1 minute. Data were collected for 30 seconds starting 15 seconds after the subjects' adaptation to a new speed condition. Any walking condition during which subjects held the treadmill handrails or verbally complained that they felt unsafe was considered an "incompatible walking condition," and the data for that walking condition were not included in the data analysis. Once a trial was stopped, the subject was not allowed to walk at faster speeds.

Materials

Three-dimensional (3D) kinematic data were collected through an Optotrak 3020 system* while subjects walked on a treadmill (Ergo esi-90†). Small (0.64-cm [0.25-in]) infrared light-emitting diodes (IREDs) were attached to the following body landmarks of the subjects: head of fifth metatarsal and lateral maleolus (foot), lateral maleolus and femoral condyle (leg), femoral condyles and femoral greater trochanter (thigh), and greater trochanter and acromial process (trunk, including pelvis and thorax). The IRED locations were sampled simultaneously by cameras at a frequency of 100 Hz. Raw data were converted to 3D coordinates using a direct linear transformation algorithm processed by custom-written MATLAB software (version 6.5).‡ The 3D reliability values were below 2 mm. Body height and weight were measured for individual body center-of-mass calculations.[24]

Assessment Parameters

Heel-strike was determined using heel markers and an algorithm developed by Hreljac and Marshall.[31] Identification of heel-strike enabled calculation of stride length and STF. Body center-of-mass locations were computed using Dempster's equations[24] for the foot, leg, thigh, and trunk (see above) and subjects' anthropometric measurements. This procedure enabled us to detect the linear center-of-mass trajectories in the frontal plane. The maximal (peak-to-peak) mediolateral amplitude in the frontal plane at each stride was set as the mediolateral sway. Maximal ankle plantar flexion and maximal hip extension ranges of motion (ROMs) during stance phase and maximal hip flexion ROM during swing phase were calculated.

Data Analysis

In order to reveal a hysteresis effect of stepwise increasing and subsequently decreasing speed, an analysis of variance (ANOVA) for repeated measures with one within-group factor (2 levels: increasing versus decreasing speed) was applied for each group. Because no significant hysteresis effects were found for either group, data at increasing and decreasing speed conditions were combined. A between-group ANOVA for repeated measures with one within-group factor for speed (7 levels) was carried out. In cases of a significant finding, a *post hoc* analysis was carried out. Due to unequal variance, a Hochberg GT2 *post hoc* test was applied. Because at each

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† Woodway USA, W229 N591 Foster Ct, Waukesha, WI 53186.
‡ The MathWorks Inc, 3 Apple Hill Dr, Natick, MA 01760-2098.

Table 2.
Effects of Systematically Increasing and Decreasing Walking Speed on Gait Parameters[a]

	Group		Speed		Group×Speed	
Variable	***F* Value**	***P***	***F* Value**	***P***	***F* Value**	***P***
STF	18.15	.00*	6.84	.04*	1.05	.39
STL	6.32	.03*	4.16	.00*	0.12	.99
LAS	7.30	.01*	29.14	.00*	2.14	.052
APF	10.96	.00*	18.79	.00*	0.57	.76
HIE	20.90	.00*	2.22	.04*	0.55	.77
HIF	13.17	.00*	7.78	.00*	0.21	.97
STF SD	0.91	.89	1.77	.27	0.87	.82
STL SD	1.96	.13	1.09	.41	1.28	.27
LAS SD	1.19	.14	1.49	.18	0.61	.72
APF SD	4.32	.02*	25.19	.00*	1.06	.39
HIE SD	16.72	.00*	3.02	.03*	1.04	.40
HIF SD	15.42	.00*	2.89	.01*	0.39	.89

[a] STF=stride frequency, STL=stride length, LAS=lateral sway, APF=ankle plantar flexion, HIE=hip extension, HIF=hip flexion, SD=standard deviation. Asterisk indicates $P<.05$.

Table 3.
Effects of the "Self-Preferred" Stride Frequency on Gait Parameters[a]

	Group		Frequency		Group×Frequency	
Variable	***F* Value**	***P***	***F* Value**	***P***	***F* Value**	***P***
STL	7.87	.01*	4.87	.04*	0.25	1.00
LAS	2.93	.03*	3.76	.02*	1.18	.26
APF	9.40	.001*	1.47	.01*	0.86	.64
HIE	3.97	.04*	1.38	.03*	0.69	.83
HIF	9.36	.001*	2.00	.00*	0.93	.54
STL SD	2.85	.09	1.31	.10	0.81	.69
LAS SD	0.07	.78	1.24	.10	1.14	.10
APF SD	0.19	.66	1.14	.22	1.27	.08
HIE SD	6.50	.01*	2.20	.00*	0.81	.70
HIF SD	24.82	.00*	1.89	.00*	1.05	.09

[a] STL=stride length, LAS=lateral sway, APF=ankle plantar flexion, HIE=hip extension, HIF=hip flexion, SD=standard deviation. Asterisk indicates $P<.05$.

walking speed subjects could freely choose their preferred STF, a similar and separate analysis was carried out with STF as the within-group factor (9 levels). The STF data were clustered into 9 groups in the observed frequency range of 0.4 to 1.2 Hz (0.1-Hz increments). Mean values were computed from the group data. The standard deviation of each group was averaged from each subject's sum of strides at each speed conditions. Due to the uneven distribution of men between the fallers (38%) and the nonfallers (52%), a 2-way between-group ANOVA for repeated measures was carried out to evaluate the effects of sex (2 levels) and speed (7) and the interaction effect between sex and speed. The level of statistical significance was set at $P<.05$.

Results

All 21 fallers were able to walk at walking speeds between 0.18 and 1.07 m/s, whereas 4 and 12 fallers (19% and 57%, respectively) reported they reached the "incompatible walking condition" at the highest walking speeds (1.3 and 1.52 m/s, respectively). Data collected from these subjects during these trials (till the malfunction point) were excluded from the data analysis; however, the data for their successful trials at slower speeds were included for analysis. All 27 nonfallers were able to successfully complete all speed conditions; thus, all of their data were all included for data analysis. The statistical analysis using sex as a within-group factor revealed no statistically significant effects for any of the dependent variables, allowing us to examine the between group differences while including both sexes, regardless of their uneven distribution. The main effects of group and speed, using walking speed as the within-group factor, and the interaction effects between group and speed are reported in Table 2. The main effects of group and frequency, using STF as the within-group factors, and the interaction effects between group and frequency are reported in Table 3. No significant interaction effects between group and speed and no interaction effect between group and frequency were revealed for any of the dependent variables (Tabs. 2 and 3).

Stride Frequency

A significant main effect of group was found for STF. *Post hoc* analysis indicated that, in all speed conditions, the fallers had significantly higher STFs compared with the nonfallers (Fig. 1). A significant main effect of speed was found for STF. Both groups increased STF with increasing walking speed (Fig. 1). No significant main effects of group or speed were observed for standard deviation in STF.

Stride Length per Walking Speed

A significant main effect of group was found for STL. The nonfallers had longer strides at all speed conditions compared with the fallers; this finding was significant only in the 1.3-m/s walking speed condition (Fig. 2). A significant main effect of speed was found for STL. Both groups showed an almost linear increase in STL from 0.85 m/s onward (Fig. 2). *Post hoc* analysis for speed conditions revealed for both groups that, at 1.52 m/s, STL was significantly larger compared with speed conditions of 0.85 m/s and lower ($P<.05$). No significant

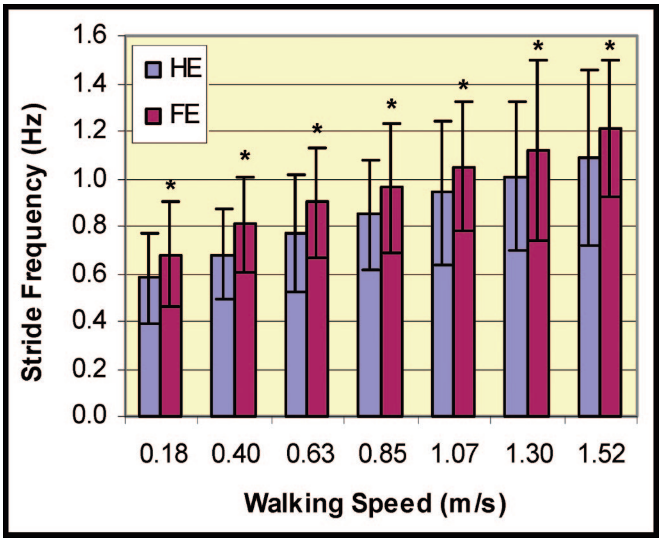

Figure 1.
Stride frequency at each walking speed for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Asterisk indicates $P<.05$ (*post hoc* analysis).

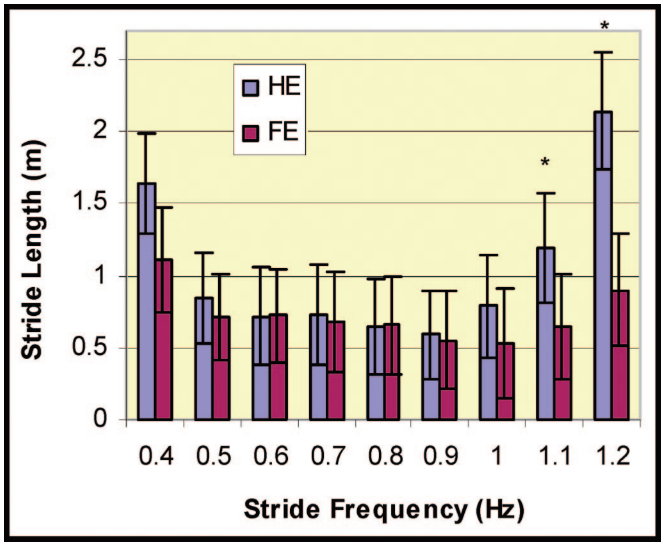

Figure 3.
Stride length at each stride frequency for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Asterisk indicates $P<.05$ (*post hoc* analysis).

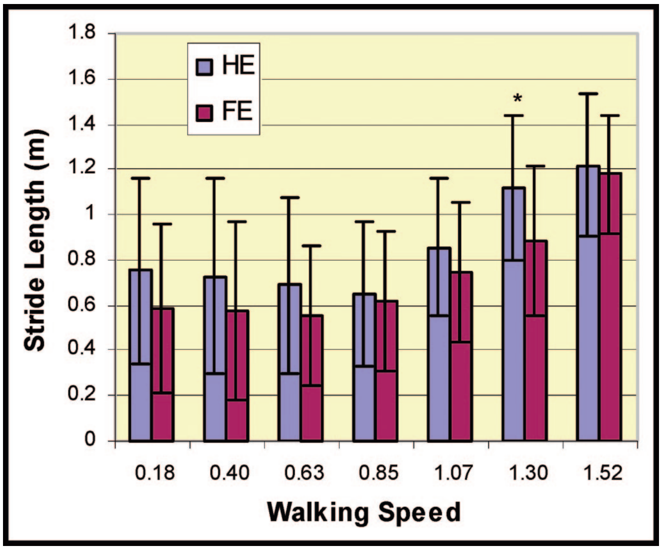

Figure 2.
Stride length at each walking speed for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Asterisk indicates $P<.05$ (*post hoc* analysis).

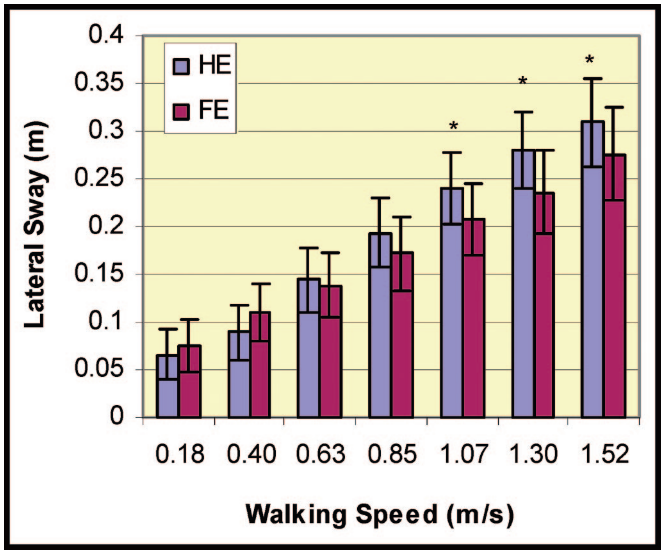

Figure 4.
Lateral sway at each walking speed for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Asterisk indicates $P<.05$ (*post hoc* analysis).

main effects of group or speed were found for standard deviation in STL.

Stride Length per Stride Frequency

A significant main effect of group was found for STL. Stride length was found to be larger in the nonfallers in most frequency conditions, reaching the level of significance at the highest STF conditions (1.1 and 1.2 Hz, $P<.05$) (Fig. 3). A significant main effect of frequency was found for STL. Both groups tended to increase STL at the relatively low (<0.5 Hz) and higher (1 Hz) STFs. Further analysis indicated that the effect of STF on STL was greater in the fallers compared with the nonfallers (Fig. 3). No significant main effects of group or speed were found for standard deviation in STL.

Lateral Body Sway per Walking Speed

A significant main effect of group was found for lateral sway. At most speed conditions, the nonfallers tended to have a wider mediolateral sway. The fallers had a significantly smaller lateral sway from 1.07 m/s onward compared with the nonfallers (Fig. 4). A significant main effect of speed was found for lateral sway. Both groups showed a lateral sway increase with increasing walking

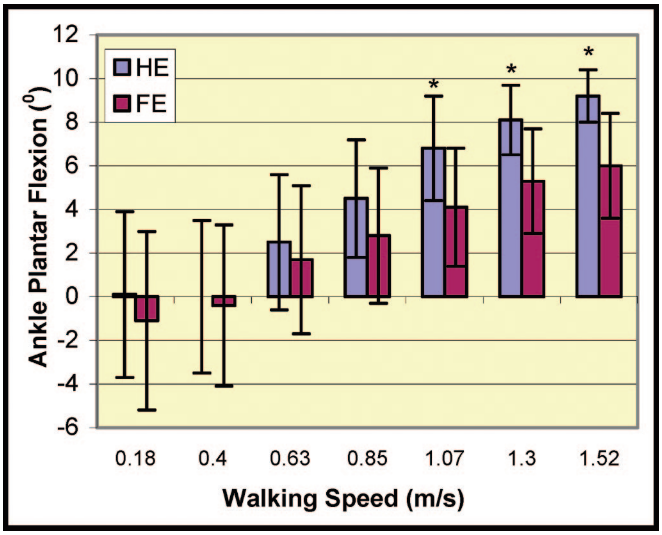

Figure 5.
Maximal ankle plantar flexion at each walking speed for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Positive value indicates ankle dorsiflexion; negative value indicates ankle plantar extension. Asterisk indicates $P<.05$ (*post hoc* analysis).

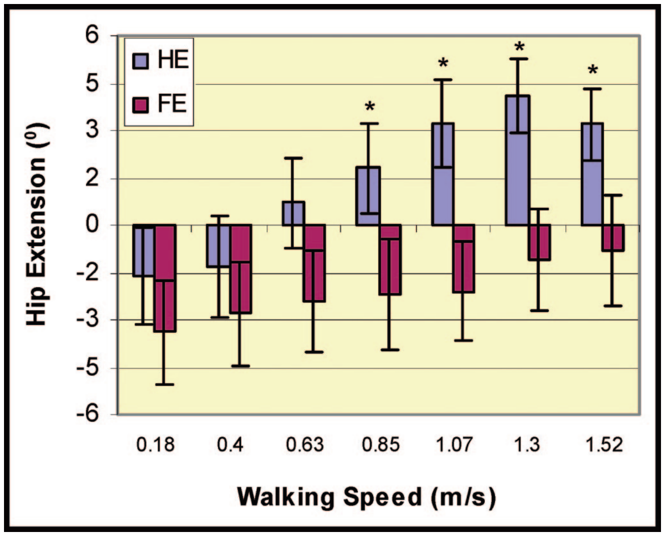

Figure 6.
Maximal hip extension angle at each walking speed for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Negative value indicates hip flexion; positive value indicates hip extension. Asterisk indicates $P<.05$ (*post hoc* analysis).

speed. The interaction effect between group and speed approached the level of significance ($P=.052$) (Fig. 4). No significant main effects of group or speed were found for standard deviation in lateral sway.

Lateral Sway per Stride Frequency

A significant main effect of group was found for lateral sway. The fallers showed a reduced lateral sway compared with the HE nonfallers at most frequency conditions (0.6–1.1 Hz), reaching the level of significance at the 1-Hz STF condition ($P=.039$). A significant main effect of frequency was found for lateral sway. Both groups showed a decrement in lateral sway with increasing STF. No significant main effects of group or speed were found for standard deviation in lateral sway.

Ankle Plantar Flexion per Walking Speed

A significant main effect of group was found for ankle plantar-flexion angle. The nonfallers showed a significantly larger ankle plantar-flexion angle from 1.07 m/s onward (Fig. 5). A significant main effect of speed also was found for ankle plantar-flexion angle. Both groups showed ankle plantar-flexion angle increases with increasing walking speed (Fig. 2). A significant main effect of group was found for standard deviation in ankle plantar-flexion angle. *Post hoc* analysis demonstrated that the fallers had a significantly larger standard deviation in ankle plantar flexion from 1.07 m/s onward ($P<.05$) compared with the nonfallers. A significant main effect of speed was found for standard deviation in ankle plantar-flexion angle, showing for both groups a decreased variability in ankle plantar-flexion angle with increasing walking speed.

Ankle Plantar Flexion per Stride Frequency

A significant main effect of group was found for ankle plantar-flexion angle. The fallers tended to develop a reduced ankle plantar-flexion angle when compared with the nonfallers regardless of STF. A significant main effect of frequency was found for ankle plantar-flexion angle. Both groups showed an ankle plantar-flexion angle reduction with increasing STF. No significant main effects of group or frequency were found for standard deviation in ankle plantar-flexion angle.

Hip Extension per Walking Speed

A significant main effect of group was found for hip extension angle. *Post hoc* analysis revealed that the nonfallers had a significantly larger hip extension from 0.85 m/s onward (Fig. 6). A significant main effect of speed was found for hip extension angle. Both groups showed a significantly increased hip extension ROM with increasing walking speed (Fig. 6). However, further analysis revealed that increasing walking speed had a greater effect on the nonfallers compared with the fallers, mainly at the higher speeds. A significant main effect of group was found for standard deviation in hip extension. The fallers showed a larger standard deviation in hip extension mainly at the higher speed conditions compared with the nonfallers (Fig. 7). A significant main effect of speed was found for standard deviation in hip extension. Further analysis revealed that there was a

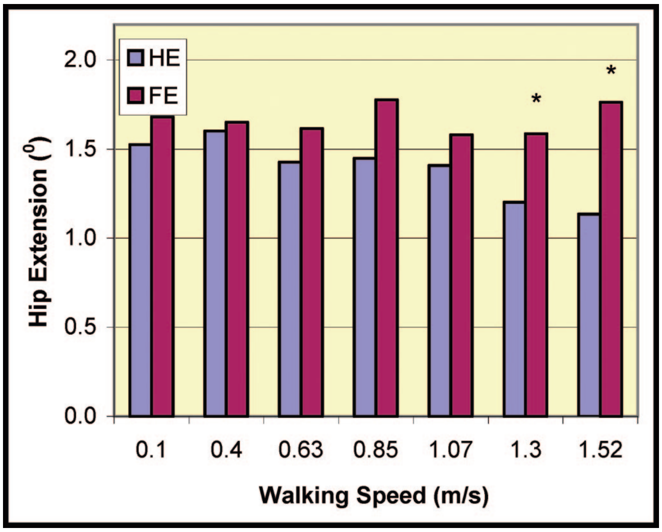

Figure 7.
Hip extension standard deviation (SD) at each walking speed for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Asterisk indicates $P<.05$ (*post hoc* analysis).

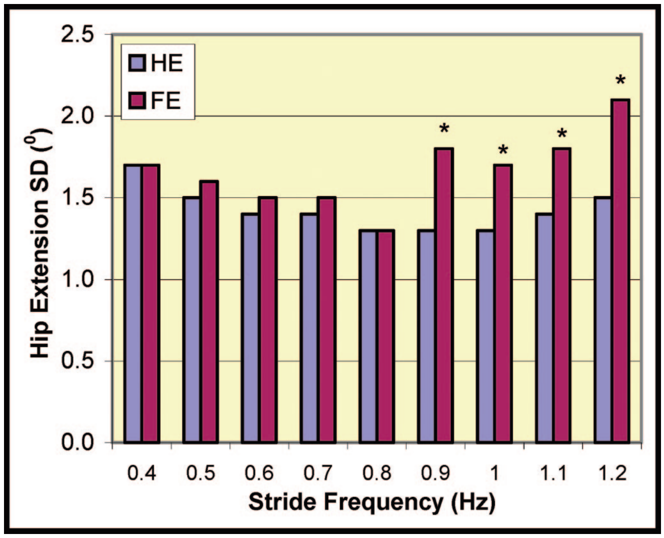

Figure 8.
Hip extension standard deviation (SD) at each stride frequency for elderly subjects who experienced at least one fall in the previous 6 months (FE) and elderly subjects with no history of falls (HE). Asterisk indicates $P<.05$ (*post hoc* analysis).

significant decrease in standard deviation in hip extension with increasing walking speed only in the nonfallers.

Hip Extension per Stride Frequency

A significant main effect of group was found for hip extension angle. *Post hoc* analysis revealed that, compared with the fallers, the nonfallers developed a significantly larger hip extension angle during the frequency range of 0.5 to 1 Hz ($P<.05$). A significant main effect of frequency was found for hip extension angle, showing that both groups showed a decrease their hip extension angle as STF increased. A significant main effect of group was found for standard deviation in hip extension. The fallers had significantly higher hip extension variability than the nonfallers at a frequency range of 0.9 to 1.2 Hz (Fig. 8). A significant main effect of frequency was found for standard deviation in hip extension. Both groups showed larger hip extension variability at the relatively low (<0.6 Hz) and higher (>1 Hz) frequency ranges (Fig. 8).

Hip Flexion per Walking Speed

A significant main effect of group was found for hip flexion. The hip flexion angle was larger at the higher walking speeds (0.85 m/s onward, $P<.05$) in the fallers compared with the nonfallers. A significant main effect of speed was found for hip flexion. Both groups showed hip flexion angle increases with increasing walking speed. A significant main effect of group was found for standard deviation in hip flexion. The fallers showed a larger variability in hip flexion in most speed conditions compared with the nonfallers, reaching the level of significance at the highest walking speed (1.52 m/s). A significant main effect of speed was found for standard deviation in hip flexion. Both groups showed a significant decrease in variability (standard deviation) in hip flexion with increasing speed. Further analysis revealed a greater effect of increasing and decreasing walking speed on hip flexion variability in the nonfallers than in the fallers.

Hip Flexion per Stride Frequency

A significant main effect of group was found for hip flexion angle. The fallers showed a greater hip flexion angle compared with the nonfallers while walking at the higher STFs (1 Hz onward, $P<.05$). A significant main effect of frequency was found for hip flexion angle. Although both groups showed increments in hip flexion angle with increasing STF, further analysis revealed that the effect of manipulating STF was greater in the fallers than in the nonfallers from 0.8 Hz onward. A significant main effect of group was found for standard deviation in hip flexion. Both groups showed a decreased standard deviation in hip flexion with increasing STF. The fallers showed a larger variability in hip flexion angle regardless of STF, reaching the level of significance in the higher frequency conditions (0.9–1.2 Hz). A significant main effect of frequency was found for standard deviation in hip flexion. A decrement in standard deviation in hip flexion angle with increasing STF was observed in both groups.

Discussion

Our main contribution to the existing data in the field of falls in elderly people derives from the different way data were collected and analyzed in our study, allowing us to trace differences in coordination patterns and in gait variability between fallers and nonfallers that, in turn, may affect the ability to prevent a fall. Data were collected using a large number of walking speeds (7), during which a relatively long (30-second) period of walking took place. In addition to the frequent method of analysis in which walking speed is used as the independent variable, the subjects' "self-preferred" STF served as independent factor. This method revealed findings that confirmed our hypotheses; that is, compared with the nonfallers, the fallers showed adaptations in STL, in STF and kinematic parameters, and in mediolateral body sway, adaptations that mechanically had the potential to decrease gait variability. Compared with the nonfallers, the fallers showed a higher STF at all walking speeds; decreased STL, ankle plantar flexion, hip extension, and lateral body sway; and increased hip flexion at the higher walking speeds (from 0.85 m/s onward). Nevertheless, the variability of the kinematic variables—that is, the standard deviation in ankle plantar flexion, hip extension, and hip flexion—was greater in the fallers, particularly at walking speeds of 1.3 and 1.5 m/s.

The limitations of the findings are that walking patterns were evaluated on a treadmill that differs from overground walking to the extent that the legs are driven by the treadmill itself and, thus, could impose changes in STL and STF and a slightly more stable coordination pattern. However, it has been demonstrated that the biomechanics of overground walking and treadmill walking are very similar[32]; thus, there are no dramatic reduced linear accelerations acting on the joint and body segments and hardly any reduced translations in space during treadmill walking. If the treadmill moves with sufficiently constant speed (which it does, as explained in the "Method" section), then no differences in kinetic and kinematic variables between overground and treadmill walking are expected. There also is possibly a perceptual difference; that is, when subjects are sampled during treadmill gait all feasible experiments will reflect the more stationary environment perceived from the treadmill. As mentioned in the "Design" section, subjects were instructed not to use the handrails during the experiment, and data for speed conditions during which handrails were used were excluded from the data analysis. Furthermore, the use of a treadmill in this study allowed us to substantially reduce the required volume for the movement recording, collecting data from more than 40 strides (per person per speed), and to easily and accurately control the walking speed.

Despite these limitations, our results confirm the findings of other gait studies that evaluated overground walking in elderly people. For example, an increased STF and a decreased STL have been observed in elderly people with limitations in muscle function[6,14,16] and balance impairments.[3,19,20] In elderly people who were healthy, a reduced angular motion in ankle plantar flexion and hip extension during push-off,[5,6] increased hip flexion during the swing phase,[5,6,17,24,25,27,33] and increased variability[34] were observed when compared with the walking patterns of young adults. In addition, an increased STF and a decreased STL have been observed during treadmill gait in patients with Parkinson disease[30] and in patients with stroke,[33] and our results coincided with those of earlier studies that demonstrated increased gait unsteadiness and stride-to-stride kinematic variations in elderly fallers compared with nonfallers.[35,36] Nevertheless, many potentially important aspects of gait variability of elderly fallers are not yet well understood. To quantitatively assess gait unsteadiness, it is helpful to measure gait over more than just a few strides. However, no previous studies have measured variability of stride-to-stride gait kinematics of elderly fallers during an extended period of walking as was done in the current study.

The spatiotemporal adaptations shown by the elderly fallers could be their mechanical solution while trying to diminish their increased gait variability, performing a direct attempt to minimize the forces acting on their musculoskeletal system during the stance phase. For example, the propulsion phase in gait is known to generate the largest mediolateral momentum that is resisted by the gluteus medius muscle to maintain the center of mass within the base of support.[37] Gluteus medius muscle weakness in the elderly fallers may have urged them to adopt a pattern that minimizes the mediolateral forces. A larger STL is associated with greater propulsive forces and thus a reduced STL (and a higher STF to maintain speed) will minimize these forces. Similarly, a higher STF shortens the whole gait cycle and thus the period of single-limb support, thereby decreasing the time and displacement amplitudes that the center of mass is outside the base of support.

Adaptations in the kinematic parameters that possibly minimize the propulsive forces were revealed in the elderly fallers. Decreased plantar flexion has been associated with low muscle power output.[6,24,25] Thus, the low muscle power output, in contrast to the normal interpretation, may be a result of a strategy to minimize the unsteadiness, rather than a cause of the changed gait pattern. The increased hip flexion at the higher walking speeds in the elderly fallers also may be a strategy that minimizes propulsive forces by achieving STL through leg lift instead of the more normal (and perturbed) push-off. Other studies[5,6] have shown similar results, suggesting that elderly adults use excessive hip flexion during the swing phase to increase mechanical energy transfer from the lower body to the upper body.

The present study also revealed that the fallers had greater variability in their walking patterns compared with the nonfallers from 0.9 Hz onward. The nonfallers showed a more or less U-shape curve for standard deviation values, with minimum standard deviation values occurring around a STF of 1 Hz, which is consistent with data obtained in other studies.[12,23,38,39] Figure 8 shows the corresponding U-shape curve for the variability of hip extension, which was not observed in the fallers. It can be hypothesized that the increased variability at the higher walking speeds in the fallers relates to a lack of flexibility to adopt new movement patterns between different body segments. Although assessing upper- and lower-extremity interactions during walking was not part of the current study, it could be argued that the changeover between 2 arm swings per stride (2:1) to 1 arm swing per stride (1:1) with increased walking speed (at approximately 0.75 m/s) demands intrinsic adaptability.[30,39–42] A reduced adaptability in the fallers may have caused them difficulties in changing a 2:1 frequency locking to a 1:1 frequency locking, thus increasing their kinematic variability. This theory was justified in earlier reports where the coordination of transverse pelvic and thoracic rotation was compared between fallers and nonfallers during walking, showing significantly greater variability for the fallers.[22,42]

The higher STF and reduced ROMs of the fallers in the present study also could be the result of a stiffer gait pattern. Using a mechanical oscillator spring model, Holt et al[12] showed that, at a comfortable walking speed of 1.25 m/s, the subjects' minimal variability took place at an average STF of 0.98 Hz, which coincides with the resonant frequency of the lower extremity during the swing phase.[34] The results for the nonfallers in the present study support these findings, showing that, at 1.30 m/s, an average STF of 1.01 Hz was observed. However, in the fallers, an average STF of 1.12 Hz was found, which suggests a greater stiffness in their walking patterns, resulting in higher resonant frequencies and reduced flexibility.

In summary, it is difficult to declare whether the fallers' gait variability itself predisposes them to falls, whether this unsteadiness indicates another phenomenon that put these individuals at risk for another fall, or whether the increased gait variability is a direct result of their fall experience. To better evaluate the present results and the potential clinical utility of gait variability measures, it will be helpful to understand what gives rise to the gait unsteadiness of elderly fallers.[35,36]

Conclusions

A number of changes occurred in the walking patterns of the elderly fallers such as a reduced push-off and a shorter STL, urging them to perform a higher STF. We hypothesize that these changes were performed in order to try and lessen their walking pattern variability. However, the fallers still showed significantly greater variable gait patterns than the nonfallers, particularly at the higher walking speeds. It may be argued that the history of falls itself and a fear of another fall may have contributed to difficulties of the fallers to alter gait patterns required when coping with changes in walking conditions, causing them to perform different kinematics adaptations. We believe that, in addition to other common factors, variability of kinematic variables as well as the ability to adapt to different walking conditions should be important gait risk factors for falls and thus deserve serious consideration when assessing the causes and symptoms associated with elderly fallers.

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