J Rehabil Med 2002; 34: 278–283

Taylor&Francis healthsciences

ANKLE DORSIFLEXION DELAY CAN PREDICT FALLS IN THE ELDERLY

Gilles Kemoun[1], Philippe Thoumie[2], Dominique Boisson[3] and Jean Daniel Guieu[4]

From the Physical Medicine and Rehabilitation Departments, [1]General Hospital, Wattrelos, [2]Rothschild University Hospital, Paris, [3]University Hospital, Lyon and [4]Neurophysiologic Explorations Department, University Hospital, Lille, France

The aim of this study was to investigate the kinematic and kinetic characteristics of walking in healthy non-faller elderly in order to develop predictive parameters for falls. A 1-year prospective trial was completed on a walking circuit with two integrated force platforms and an optoelectronic system for three-dimensional movement analysis. Gait was investigated in 54 volunteers who were healthy people over 60 who had not fallen in the previous year. The subjects were contacted 2-monthly over a period of 1 year. The results showed that 16 of the 54 people tested had fallen. There was no significant age difference between the group of fallers and the group of non-fallers. Fallers walked more slowly and tended to use a double support for a longer period of time. Fallers were less powerful but mainly showed fewer power and moment variations. The range of motion at the ankle and the hip was reduced. We noticed a change in the walking pattern, showing a delay in the dorsiflexion of the ankle at the swing phase. In conclusion, subclinical gait parameters occur in older people. The advent of neuromotor pattern alterations when walking is related to the tendency to fall. Ankle dorsiflexion delays, in particular, appear to be predictive of falls.

Key words: gait analysis, falls, older people.

J Rehabil Med 2002; 34: 278–283

Correspondence address: Docteur Gilles Kemoun, Service de Rééducation et de Réentraînement à l'effort, Centre Hospitalier de Wattrelos, Rue du Docteur Alexander Fleming, 59393 Wattrelos Cedex, France. E-mail: gkemoun@nordnet.fr

Submitted April 26, 2001; accepted May 15, 2002

INTRODUCTION

Gait impairments can be related to a perceptible organic or psychological pathology. Disorders are often idiopathic and are usually multifactorial. In many studies, attempts have been made to show the changes in gait as people age, but results are often conflicting and consequently it is difficult to know exactly what changes are due to ageing and what changes are due to unassociated pathologies.

Falling is one dramatic consequence of walking troubles. With regard to prevalence, between 30% and 50% of people over the age of 65 fall at least once a year (1). In addition, it appears that cross-sectional studies may underestimate the actual number of falls that are demonstrated by longitudinal studies (2).

The aetiology of the falls is most often multifactorial, but two causal categories can be distinguished: extrinsic (those caused by obstacles) and intrinsic (those caused by pain or disease in the person concerned) (1, 3). The consequences of these falls are sometimes dramatic (4).

Frequent disorganization of neuromotor patterns in the elderly has been demonstrated by Woollacott et al. (5). They observed an increase in the latency of the muscular response and an activation of the hip muscles before those of the ankle, a pattern that is inverted in young subjects. The authors suggest that deterioration of postural control is the reason for the balance modifications in older people.

Some authors have analysed the risk factors of falls (6, 7). Many studies analysing older people's gait have been published; these studies include both clinical and experimental protocols, involve a variety of different procedures, and have provided explanations for some gait characteristics (8–12). In addition, a few studies have stressed the potential relationship between falling and standing (13–15). However, as far as we can tell, few studies have been made of the dynamic characteristics of the gait of the elderly that could predict falls (16).

The aim of this study was to use a method of three-dimensional movement analysis to discover the kinematic and kinetic characteristics of walking in voluntary, healthy, non-faller older people and to determine whether there are specific parameters which could predict falls. Assuming such parameters exist, prevention protocols designed to correct these parameters could be developed.

Our principal hypothesis is that ageing leads to a gait modification that affects the neuromotor pattern of dynamic postural control and the range of motion for lower limb joints. Given that, the presence of muscular co-contractions would cause a decrease in the articular range of motion, moments of force and powers during gait. A modification of the kinematics and the kinetics in relation to falls should also be observed.

EQUIPMENT AND METHOD

This study concerned 54 volunteers, 38 men and 16 women. All were healthy people over 60 who had not fallen in the previous year. The mean age was 66.72 ± 4.88 years (60 to 77). These persons lived at home and were completely independent. They had been asked to respond to medical questionnaires in past years and were familiar with noting down all events concerning their health. Each participant gave his/her informed consent, and the research protocol was approved by the local research ethics committee.

 ISSN 1650–1977

Table I. *Basic kinematic data (median (interquartile range)) comparing fallers and non-fallers*

	Fallers	Non-fallers	
Cadence (step/min)	99.00 (89.50 to 102.50)	108.00 (106.00 to 115.00)	$p = 0.059$
Speed (m/s)	0.96 (0.69 to 1.09)	1.29 (1.11 to 1.36)	$p = 0.026$*
Stride length (m)	1.12 (1.01 to 1.28)	1.31 (1.27 to 1.42)	$p = 0.108$
Step length (m)	0.57 (0.50 to 0.64)	0.65 (0.64 to 0.73)	$p = 0.120$
Stride duration (s)	1.20 (1.16 to 1.34)	1.11 (1.04 to 1.12)	$p = 0.058$
Double support duration (%)	27.78 (25.69 to 29.38)	23.21 (22.92 to 25.93)	$p = 0.024$*
Single support duration (%)	36.99 (35.35 to 37.6)	38.15 (35.71 to 40.0)	$p = 0.108$
Step duration (%)	49.31 (48.75 to 50.83)	50.00 (48.08 to 51.78)	$p = 0.671$
Single support start (%)	13.51 (12.39 to 14.88)	13.55 (9.68 to 14.29)	$p = 0.524$
Double support start (%)	50.68 (49.17 to 51.24)	50 (48.21 to 51.92)	$p = 0.671$
Swing start (%)	64.64 (62.40 to 65.37)	62.10 (60.66 to 64.28)	$p = 0.077$

* Statistically significant differences between fallers and non-fallers.

Each participant had a detailed clinical examination by the same physician to verify that the individual had no perceptible neurological, locomotor or cardiovascular pathologies and was taking no medication known to increase falls. All subjects were retired and lived at home. All of them led an active life, but no one participated in sporting activities.

A 10 m indoor walking circuit was used for this study. The circuit was composed of two BIOVEC 1000 force integrated platforms (0.5 m × 0.5 m), each with 250 Hz sampling rate, placed one behind the other (Advanced Mechanical Technology Inc., Newton, MA, USA), and an optoelectronic system for three-dimensional analysis of movement (Vicon System, Oxford Metrics Ltd, Oxford, England). This system recorded the trajectory of the reflecting markers placed on the subject's body via 5 infrared stroboscopic cameras (50 Hz sampling rate), 2 located on each side of the walking path and one in front, providing a viewing volume of 2.8 m length by 1.0 m width and 1.8 m height. Thirteen spherical retroreflective markers (2.5 cm in diameter) defined the different segments of the subject's pelvis and lower limbs. The markers were placed bilaterally in accordance with the Vicon Clinical Manager user's guide on anatomically well-defined points of the lower limbs: anterosuperior iliac spine, thighs, knees, heels, lateral malleoli, toes, and sacrum.

The subject performed the trials barefoot, clothed in his or her underclothes. Following the identification of anatomic points, markers were placed on the subject's body, always by the same operator. Five dynamic recordings of the subject's gait over the entire circuit were made. The subject stood at the beginning of the circuit and started spontaneously, moving at his/her own speed. The recording began as soon as the subject entered the viewing volume, and it ended when the subject left the viewing volume. The subject stopped at the end of the circuit and came back to the starting point. In order to ensure the accuracy of the measurements, the machines were calibrated prior to every session, thus maintaining a minimal precision of 1.5 mm in the three-dimensional markers' position.

For a period of 1 year following the session, the participants recorded their falls in a special diary. Another physician, completely uninvolved with the rest of the study, called each participant every 2 months to collect the new data, specifically as it pertained to falls. For the purposes of this study, fall is defined as an unexpected event when a person fell to the ground from an upper level or on the same level, including falls on stairs and onto a piece of furniture (17).

These phone calls permitted an official record of new symptoms that could provide evidence of an organic or psychological pathology, as well as records of any new drugs that could have an effect on the incidence of falling. From the information pertaining to falls, the subjects were divided into two groups: fallers and non-fallers.

A gait analysis of walking was completed using VCM software (Vicon Clinical Manager). This software provides its own biomechanical model of the neuro-musculo-skeletal system of the pelvis and of the legs during walking, which allows an estimation of variables which cannot be measured directly (18). Graphical representations are constructed using 51 frames of the VCM software. These frames situate the movement in time and allow a frame-by-frame comparison of the phases of the stride.

The analysis dealt with joint amplitudes (peak and range of motion), moments of force and powers. In our model, the moments of force in the articular centre represent the elastic loads of the soft tissues, mostly provided by the muscles controlled by the neuro-muscular system (19). These moments give a good indication of the muscle activity at a specific joint and allow an evaluation of the agonist/antagonist muscle balance. The powers confirm the result of the moments of force and angular speeds. The generation of power confirms the concentric activity of a muscular group and the absorption agrees with the eccentric activity (9). These variables reveal the functional role of the anatomic structures and can be undervalued when there is simultaneous activity of the agonists and the antagonists (co-contractions) (9, 19). For this reason it is important to compare all the data in order to ensure a functional, rational interpretation.

The data collected in the sagittal plane were the only ones processed; first, because during walking most work is done in the sagittal field, since the aim of locomotion is to move the body against gravity by realizing the movements which propel it ahead in the plane of progression (19). Second, because given the technology used the data collected in the frontal or horizontal plane may not be reliable, data in these planes were not processed. The use of skin markers has the disadvantage of shifting position in relation to the anatomic marks, despite scrupulous procedure. Some authors have established the effect of these uncertainties in the definition of "embedded axes", according to Euler, as described by Kadaba et al. (18).

The statistical analysis concerned the descriptive criteria (age, sex, weight, height), the kinematic data and the numerical data concerning the sagittal plane. Comparison of the numeric, descriptive, kinematic and kinetic data between the two groups was done using the non-parametric test of Mann and Whitney, as adapted to our small group. Data are presented with median values and interquartile range.

RESULTS

The information gathered during the follow-up phone calls revealed that 16 persons out of the 54 participants had fallen. The other 38 did not describe any noticeable event, either medical, therapeutic or accidental, apart from some benign winter illnesses.

None of the 16 fallers had had a change of treatment since the first interview and no new clinical symptoms were mentioned before, after or in conjunction with the falls and hence all falls were judged to be accidental.

Eight patients described forward falls at home caused by an obstacle (carpet or pet) and 8 patients described backward falls on slippery ground. The falls were judged to be benign without any somatic or psychic consequences. Thus two groups could be determined: a group of fallers (F) including 16 subjects (12

Table II. *Kinematic and kinetic data in the sagittal plane for the ankle, knee and hip (median (interquartile range)) comparing fallers and non-fallers*

	Fallers	Non-fallers	
Ankle:			
Power peak (W/kg), P+	2.53 (2.42 to 2.82)	3.12 (2.53 to 3.65)	$p = 0.120$
T1 (U/51)	28.00 (27.00 to 29.00)	26.00 (26.00 to 27.00)	$p = 0.077$
Moment peak (Nm/kg), M+	1.58 (1.53 to 1.75)	1.54 (1.47 to 1.64)	$p = 1.00$
T2 (U/51)	25.00 (25.00 to 25.00)	23.00 (22.00 to 23.00)	$p = 0.006$*
A1–A2 (deg)	18.50 (17.50 to 19.00)	23.00 (19.75 to 24.00)	$p = 0.040$*
A2–A3 (deg)	6.50 (5.25 to 8.00)	13.00 (10.38 to 15.00)	$p = 0.020$*
T3 (U/51)	33.00 (33.00 to 34.00)	31.00 (31.00 to 31.75)	$p = 0.016$*
Knee:			
Power peak (W/kg)	−0.81 (−0.95 to −0.70)	−1.35 (−1.48 to −1.25)	$p = 0.057$
Power variation (W/kg)	0.91 (0.88 to 1.01)	1.42 (1.34 to 1.57)	$p = 0.057$
Moment peak (Nm/kg)	−0.17 (−0.20 to −0.07)	−0.04 (−0.27 to 0.08)	$p = 0.671$
Moment variation (Nm/kg)	0.74 (0.46 to 0.82)	0.74 (0.61 to 0.84)	$p = 0.524$
Range of motion peak (deg)	62.00 (53.00 to 63.50)	63.00 (56.00 to 66.00)	$p = 0.777$
Range of motion variation (deg)	50.10 (40.50 to 53.50)	52.00 (44.25 to 55.00)	$p = 0.358$
Hip:			
Power peak (W/kg), P-	−0.93 (−0.94 to −0.69)	−1.23 (−1.34 to −0.89)	$p = 0.157$
Power variation (W/kg)	1.02 (0.81 to 1.70)	2.04 (1.80 to 2.27)	$p = 0.016$*
Moment peak (Nm/kg), M-	−0.54 (−0.48 to −0.74)	−0.97 (−1.12 to −0.77)	$p = 0.024$*
Moment variation (Nm/kg)	0.88 (0.85 to 0.90)	1.60 (1.67 to 1.71)	$p = 0.011$*
Range of motion peak (deg)	57.20 (45.80 to 61.25)	60.48 (55.30 to 64.35)	$p = 0.258$
Range of motion variation (deg), ΔA	40.28 (38.25 to 44.35)	47.26 (42.59 to 50.36)	$p = 0.007$*

* Statistically significant differences between fallers and non-fallers.

T1 = power peak time, T2 = moment peak time, T3 = dorsiflexion peak time, U/51 = number of frames of the gait cycle recording at the peak time, A1–A2 = plantar flexion during the second phase of double support time, A2–A3 = dorsiflexion at the beginning of the swing.

males and 4 females), and a group of non-fallers (NF) including 38 subjects (26 males and 12 females). There were no significant differences with regard to age, weight and height. We assumed there was no difference between men and women regarding falls and tested our assumption with Fisher's exact test for small samples. The results of our test showed that, in our population, we could not assume no difference due to sex ($p = 0.210$).

The basic kinematic data are detailed in Table I and we can see that the F walk more slowly with a tendency to reduce cadence. Their steps are shorter and they favour double support over single support, thus delaying the swing phase.

The kinematic and kinetic data in the sagittal diagram concern the ankle, the knee and the hip.

Ankle (Table II and Fig. 1): The dorsiflexion of the ankle during the second phase of double support (A_1–A_2) and the plantar flexion of the ankle at the beginning of the swing (A_3–A_2) were significantly lower for the F ($p = 0.040$ and $p = 0.020$) than for NF. Moreover a significant delay of the dorsiflexion peak for the F in relation to the NF was observed ($p = 0.016$). A significant delay in the moment peak for the F in relation to the NF was observed. The generation of power (P+) at the beginning of the phase of second double support was lower for the F but was not significant ($p = 0.120$). A delay of the power peak was observed for the F in relation to the NF, but this difference was not significant.

Knee (Table II and Fig. 2): The results are statistically non significant.

Hip (Table II and Fig. 3): The range of motion in flexion (ΔA) was significantly lower for the F ($p = 0.007$) than for NF. The maximal flexion moment (M-) at the end of the single support phase and at the beginning of the second double support phase was significantly lower for the F ($p = 0.024$). The change of power between the minimum peak (P-) at the back step at the end of the single support phase and the maximum peak at the beginning of the swing phase was also significantly lower for the F ($p = 0.016$) than for NF.

On the whole, the F were less powerful. The force variations and the moment variations were also less important. The range of motion during gait was reduced and, finally, we noticed a change in the walking pattern with a delay in the dorsiflexion of the ankle at the beginning of the swing phase.

DISCUSSION

The number of subjects in our study is unusual considering the technical modalities. In the literature, the published studies with similar technical modalities were based on smaller samples. Elble et al. explored step initiation in 2 groups of 4 people (20) and Winter et al. studied biomechanical changes of walking in a group of 15 elderly people (21). The difficulties in our study were chiefly due to the constraining character of the test, which required an experienced operator, complicated equipment and adjustments, a long testing time and a very long period of data analysis. However some non-significant results in our study with relatively small sample size may be due to a type II error.

Fall data were obtained via diaries and phone interviews. The value of diaries to clinical research has been documented (17, 22). Keeping a ledger diary becomes part of everyday life

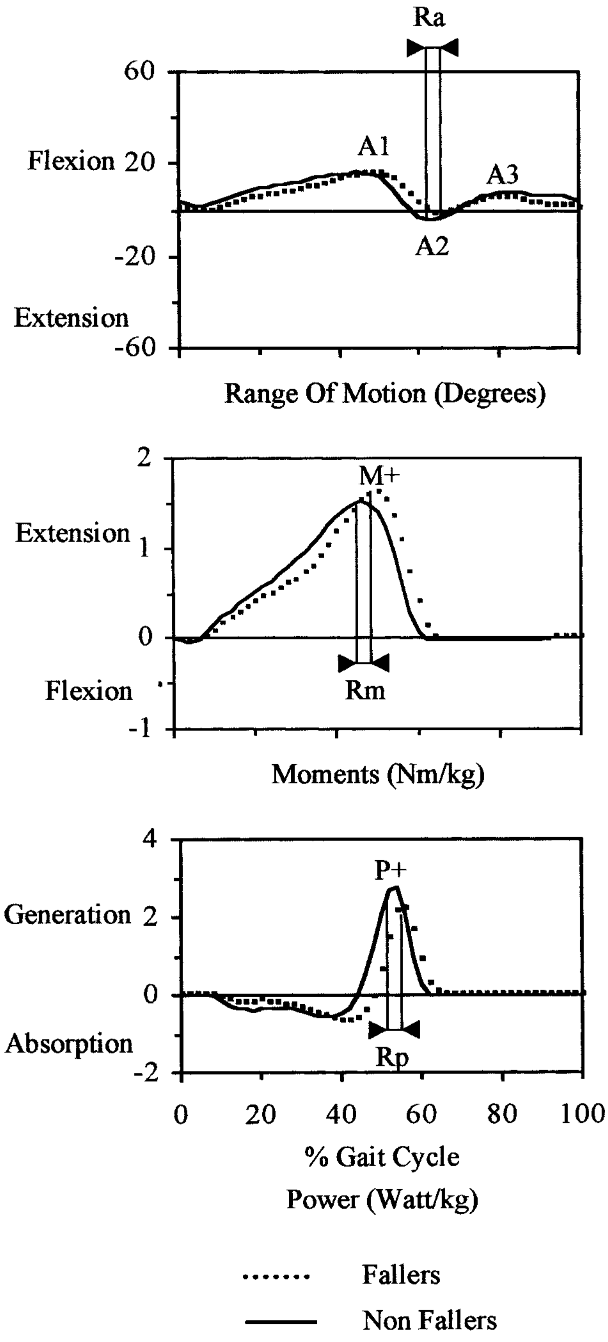

Fig. 1. Ankle kinematic and kinetic indices (range of motion, moments and power) during the gait cycle in the sagittal plane for fallers and non-fallers. A1 = dorsiflexion peak during support phase; A2 = plantar flexion peak at preswing; A3 = dorsiflexion peak during swing; A1–A2 = plantar flexion during the second phase of double support time; A2–A3 = dorsiflexion at the beginning of the swing; Ra = dorsiflexion delay; M+ = moment peak; Rm = moment peak delay; P+ = power peak; R*p* = power peak delay.

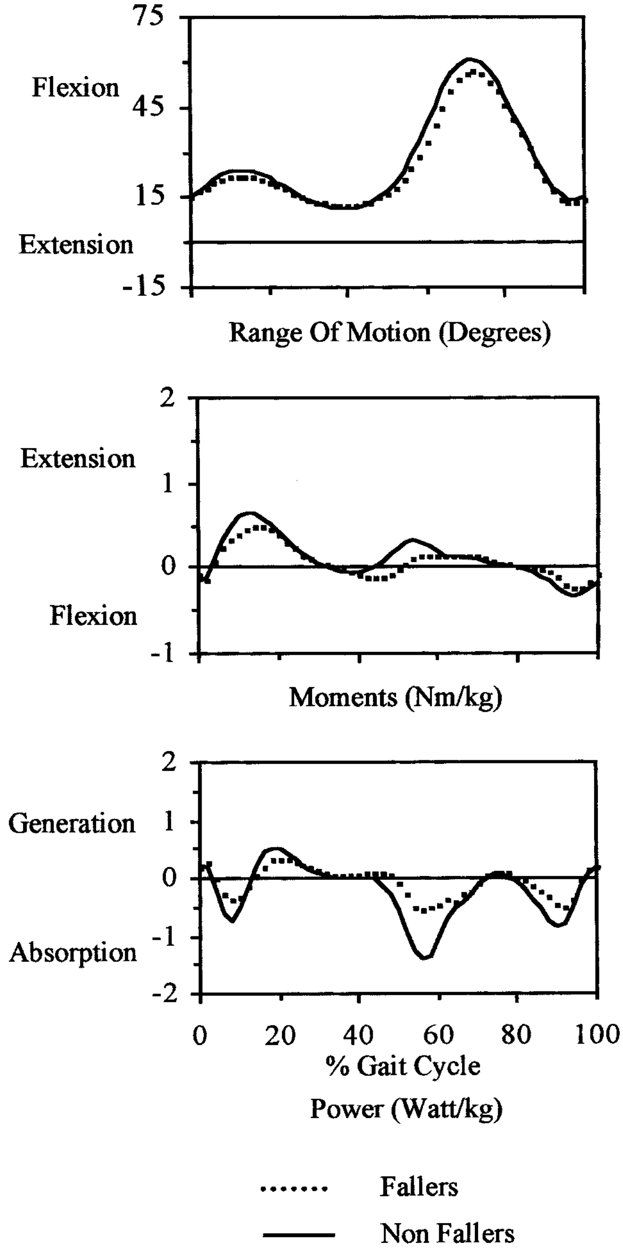

Fig. 2. Knee kinematic and kinetic indices (range of motion, moments and power) during the gait cycle in the sagittal plane for fallers and non-fallers.

and each event is recorded when it happens. It is not dependent on recall during a clinic visit (23).

Most studies show an increase in falls with age (3). In their prospective study concerning balance and falls, Maki et al. did not find any significant age differences between the group of fallers and the group of non-fallers (15). In our study, the average age was lower than the age of maximal incidence of falls (>80 years old). This is due to the fact that it is very difficult to find aged volunteers who qualify for the test (no perceptible pathology which could change walking, and no known treatment favouring falls).

Moreover, the proportion of fallers in our population (29.6%) was close to the known epidemiological data for frequency of falls (estimated to be 30% after 65) (24). In our study, the lack of difference in age may be due to the fact that inclusion criteria

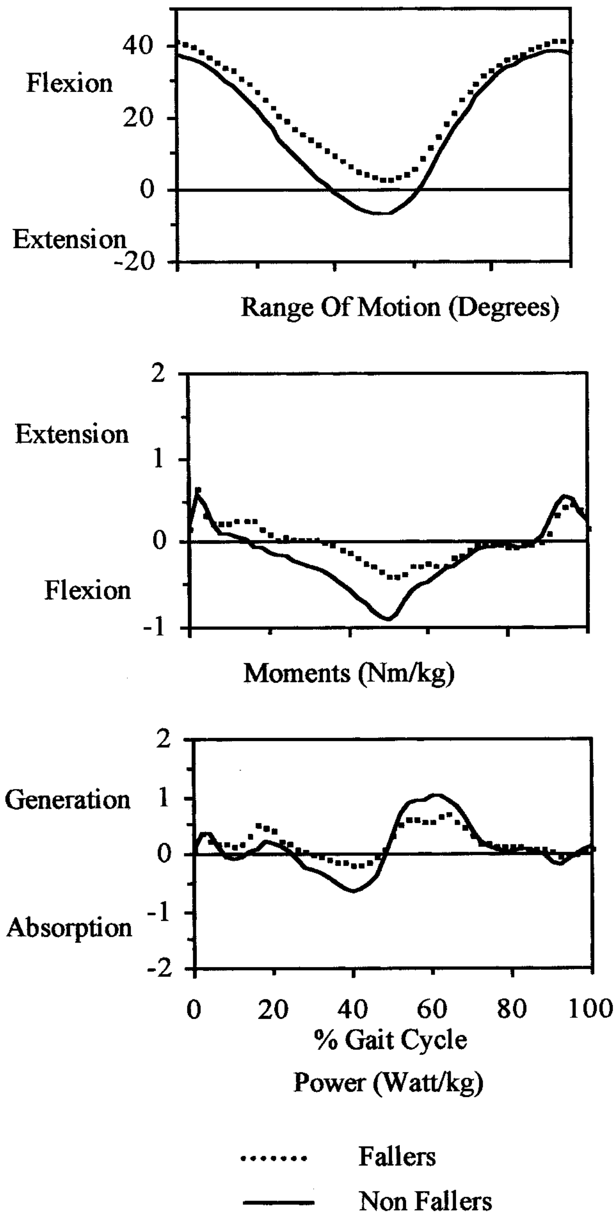

Fig. 3. Hip kinematic and kinetic indices (range of motion, moments and power) during the gait cycle in the sagittal plane for fallers and non-fallers.

were based on functional data, independent of age, and allowed cancellation of the effect due to age. In the studies of Fernie et al. on a homogeneous sample of subjects living in an institution, there was also no difference in age between fallers and non-fallers (14). It is known that the incidence of pathologies and functional deterioration increases with age; however, there is considerable individual variability. Thus the influence of age should be tempered and it should only be considered as an indirect factor.

Fernie et al. carried out a double-control prospective study concerning the relationship between falls and postural oscillations for 205 aged subjects. They found 46% of falls in women, and 30% in men, and did not note any difference with regard to posture between men and women (14). Hageman et al. did not notice any differences with regard to gender on postural control measures in the healthy elderly while the measures studied were sensitive to age-related changes (25). In our study, the inclusion criteria provided a homogeneous group from a medical and functional point of view. Consequently, our results cannot be completely generalized for all older people who fall.

In the literature, there are fall studies comparing aged fallers and non-fallers in the static field (5, 13–15, 26, 27) and studies which estimate the modification of walking for old people in relation to young people (21, 28) but, to our knowledge, few have studied the elderly's gait changes in relation to falls (10, 16, 18).

At the level of the hip, we observed a consistent reduction in range of motion during walking for fallers (while the clinical assessment did not show lower extremity limitations), as well as a less important variation of moments of force, chiefly appearing in the second part of the cycle. These differences are caused by a variation in the neuromotor pattern between the agonist muscles (psoas and quadriceps) and the antagonist muscles (hamstring and gluteus) with a tendency towards co-contraction for fallers. These results are consistent with those of Kerrigan et al. that show an isolated and consistent reduction in hip extension while walking in the elderly; this tendency is exaggerated in fallers (10).

In addition, for the fallers, there is a significant reduction of power partly caused by a less important variation of moments of force, but also by a less important variation of angular speed. Therefore, for the non-fallers, there is more absorption of power which can be returned during the swing phase, allowing dynamic balance, thus making walking less costly on the energetic level (29). Devita & Hortobagyi (8) show that the altered motor pattern is manifested as a redistribution of joint torques and powers, which alters the relative contributions from the various muscle groups to the total performance. The results of Judge et al. also support these studies as do our own (9).

At the level of the ankle, we did not notice any significantly different variation of power and moment between fallers and non-fallers, although there was a tendency towards lower levels of power and moment for the fallers.

We observed that the peak moment of plantar flexion was significantly delayed for fallers, which indicated delayed propulsion. Moreover, the dorsiflexion occurred late for fallers, which could explain the catching phenomena as the step proceeded. It is well known that these phenomena facilitate falls. As far as we know, this has not been demonstrated before.

Moreover, in our study as well as others in the literature, fallers exhibited a significant decrease in range of motion during gait for the plantar or dorsiflexion, whereas the analytical, clinical examination did not reveal any particular anomalies. These lower amplitudes as well as the delayed starting of dorsiflexion were accompanied by an imbalance of the soleus–tibialis anterior couple. This neuromotor pattern, which occurred during the swing phase, therefore generated the dynamic modifications that facilitated catching the foot as the step proceeded.

Thus it is not only the deficiency of the dorsiflexors that causes falling but also the continued contraction of plantar flexors. This observation is very important in rehabilitation, where the tendency is to reinforce dorsiflexors for old people who fall often, whereas it seems also important to modify the neuromotor pattern.

In conclusion, in this study, we have described the features of older people's walking; some subjects fell and we were able to show parameters in relation to the falls. We found several elements that confirm our hypothesis concerning the alteration of neuromotor pattern and range of motion during gait. Our results are consistent with those in the literature. A "senile" walking pattern beginning at a subclinical stage truly exists and we were able to analyse this. Range of motion and muscle strength, which were studied here with moments of force and power, decrease for fallers. However, neuromotor patterns are of prime importance in the dynamic balance. The advent of neuromotor pattern alterations by co-contractions, when walking, is related to falls and we showed that ankle dorsiflexion delay during gait could predict falls.

ACKNOWLEDGEMENT

The authors wish to thank Lisa Spencer for her help in revising the English version.

REFERENCES

1. Sheldon JH. On the natural history of falls in old age. Br Med J 1960; 4: 1685–1690.
2. Perry BC. Falls among the elderly: a review of the methods and conclusions of epidemiologic studies. J Am Geriatr Soc 1982; 30: 367–371.
3. Hindmarsh J, Harvey Estes E. Falls in older persons: causes and interventions. Arch Intern Med 1989; 149: 2217–2222.
4. Murphy J, Isaacs B. The post-fall syndrome: a study of 36 elderly patients. Gerontology 1982; 28: 265–270.
5. Woollacott MH, Vonhosten C, Rosblad B. Aging and postural control: changes in sensory organisation and muscular coordination. Int J Aging Human Dev 1986; 23: 23–81.
6. Robbins AS, Rubinstein LZ, Josephson KR, Schulman BL, Osterweil D, Fine G. Predictors of falls among elderly people. Arch Intern Med 1989; 149: 1628–1633.
7. Tinetti ME, Speechley M, Ginter SF. Risk factors for falls among elderly persons living in the community. N Engl J Med 1988; 319: 1701–1707.
8. Devita P, Hortobagyi T. Age causes a redistribution of joint torques and powers during gait. J Appl Physiol 2000; 80: 1804–1811.
9. Judge JO, Ounpuu S, Davis RB. Effects of age on the biomechanics and physiology of gait. Clin Geriatr Med 1996; 12: 659–678.
10. Kerrigan DC, Lee LW, Collins JJ, Riley OP, Lipsitz LA. Reduced hip extension during walking: healthy elderly and fallers versus young adults. Arch Phys Med Rehabil 2001; 82: 26–30.
11. Kerrigan DC, Todd MK, Della Croce U, Lipsitz LA, Collins JJ. Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. Arch Phys Med Rehabil 1998; 79: 317–328.
12. Grabiner PC, Biswas ST, Grabiner MD. Age-related changes in spatial and temporal gait variables. Arch Phys Med Rehab 2001; 82: 31–35.
13. Keshner EA, Allum JH, Honegger F. Predictor of less stable postural responses to support surface rotations in healthy human elderly. J Vestib Res 1993, 3: 419–429.
14. Fernie GR, Gryfe CI, Holliday PJ, Llewelyn A. The relationship of postural sway in standing to the incidence of falls in geriatric subjects. Age and Ageing 1982; 11: 11–16.
15. Maki BE, Holliday PJ, Topper AK. A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. J Gerontol 1994; 49: M72–M84.
16. Maki BE. Gait changes in older adults: predictors of falls or indicators of fear? J Am Geriatr Soc 1997; 45: 313–320.
17. Luukinen H, Koski K, Laippala P, Kivela SL. Factors predicting fractures during falling impacts among home-dwelling older adults. J Am Geriatr Soc 1997; 45: 1302–1309.
18. Kadaba MP, Ramakrishnan HK, Wootten ME. Measurement of lower extremity kinematics during level walking. J Orthop Res 1990; 8: 383–392.
19. Eng JJ, Winter DA. Kinetic analysis of the lower limbs during walking: what information can be gained from a three dimensional model? J Biomechanics 1995; 28: 753–758.
20. Elble RJ, Moody C, Leffler K, Sinha R. The initiation of normal walking. Movement Disorders 1994; 2: 139–146.
21. Winter DA, Patla AE, Franck JS, Walt SE. Biomechanical walking pattern changes in the fit and healthy elderly. Phys Ther 1990; 70: 340–347.
22. Stack E, Ashburn A. Fall events described by people with Parkinson's disease: implications for clinical interviewing and the research agenda. Physio Res Int 1999; 4: 190–200.
23. Gray P, Hildebrand K. Fall risk factors in Parkinson's disease. J Neurosci Nurs 2000; 32: 222–228.
24. Melton LJ, Riggs BL. Risk factors for injury after a fall. Clin Geriatr Med 1985; 1: 525–539.
25. Hageman PA, Leibowitz JM, Blanke D. Age and gender effects on postural control measures. Arch Phys Med Rehabil 1995; 76: 961–965.
26. Maki BE, Holliday PJ, Topper AK. Fear of falling and postural performance in the elderly. J Gerontol 1991; 46: M123–M131.
27. Woollacott MH, Manchester DL. Anticipatory postural adjustments in older adults: are changes in response characteristics due to changes in strategy? J Gerontol 1993; 48: M64–M70.
28. Blanke DJ, Hageman PA. Comparison of gait of young men and elderly men. Phys Ther 1989; 6: 144–148.
29. Dillingham TR, Lehman JF, Price R. Effect of lower limb on body propulsion. Arch Phys Med Rehabil 1992; 73: 647–651.