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Comparison of gait symmetry between poststroke fallers and nonfallers during level walking using triaxial accelerometry

A STROBE-compliant cross-sectional study

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Abstract

To compare the degree of gait symmetry of chronic poststroke fallers with that of nonfallers during level walking using triaxial accelerometry.

In this cross-sectional study, a total of 14 patients with chronic stroke were recruited from a community hospital from February 2015 to July 2016. Patient characteristics, including the number of falls in the previous 12 months, were obtained from medical records. The Berg Balance Scale (BBS) and timed up and go (TUG) test were used at the onset of the study. Triaxial accelerometers were attached to the back and bilateral lower extremities of each subject with sampling rates of 120 Hz. The cross-correlation between the acceleration signals of the affected and unaffected feet was measured to assess the degree of gait symmetry. The triaxial acceleration signals of the 5 consecutive and bilateral strides from the middle of each trial were processed to measure the cross-correlation and time delay (T_s) between the magnitude of the acceleration vector of the affected and unaffected foot.

After controlling for possible confounding factors, the mixed-effect models showed that cross-correlation was significantly higher among nonfallers than fallers ($\beta = -0.093$; standard error [SE] = 0.029; *P*-value = 0.002), and that the T_s was significantly longer among fallers than nonfallers ($\beta = -1.900$; SE = 0.719; *P*-value = 0.011).

Cross-correlation and T_s between the affected and unaffected lower extremities may be useful indicators to distinguish poststroke fallers from nonfallers.

Abbreviations: BBS = Berg Balance Scale, BMI = body mass index, Cc_{norm} = normalized cross-correlation, RSS = root-sum-of-squares, SE = standard error, T_s = time delay, TUG = timed up and go.

Keywords: accelerometry, chronic stroke, fall, gait symmetry

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Authorship: WCL designed the study, did statistical analyses, drafted the initial manuscript, and revised important content. YHC and TSK designed the study and drafted the initial manuscript. CHH participated in study design and interpretation of results. YLZ performed data mining and contributed to data analyses. WFW designed the study, contributed to interpretation of results, and revision for important content. WFW is the guarantor of this work, has full access to all the data in the study, and takes responsibility for the integrity of the data and the accuracy of the data analysis.

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1. Introduction

The presence of an asymmetric gait pattern impairs walking performance following stroke.[1,2] A hemiparetic gait is characterized by slow, laborious, and asymmetric limb movements, and is frequently seen in patients with chronic stroke.[3]

Falls are common among chronic stroke patients, occurring in approximately 50% of patients within the 1st year after stroke onset.[4] Falls can have serious consequences in this population. For example, individuals with stroke are much more likely to sustain a hip fracture due to a fall and to lose independent mobility or die after a hip fracture than people without stroke.[5] Falls and their prevention are thus important issues in poststroke care.[6]

Studies on kinetics and kinematics in individuals with stroke have largely focused on gait velocity,[7] ground reaction force profiles,[8] balance,[9] and electrical activity in muscles as measured by electromyography.[10] Few studies, however, have used data on acceleration of the lower extremities to compare differences in gait asymmetry between poststroke fallers and nonfallers. Several recent studies have reported on the use of wireless inertial sensors to measure gait asymmetry during walking in normal subjects[11,12] and stroke patients.[13] Wireless inertial sensors have significant advantages over conventional methods, in that they are smaller in size, lighter in weight, noninvasive, and have been shown to have better validity and reliability.[14] In digital signal processing, cross-correlation has been shown to be a good indicator of the degree to which the 2 signals are similar.[15] The cross-correlation of wireless inertial sensors was recently used to measure gait symmetry and time delay (T_s) during the gait cycle when walking with loads on the lower extremities in healthy individuals.[11] The results of that study showed that during walking with loads on 1 lower extremity the values of the normalized cross-correlation (Cc_{norm}) were lower than the values found in normal gait. In addition, a significant T_s was also found in asymmetrical gait.

We hypothesized that gait asymmetry is associated with fall history in chronic stroke patients. Therefore, in this study we compared the degree of gait symmetry of chronic poststroke fallers with that of nonfallers during level walking using triaxial accelerometry.

2. Methods

2.1. Subjects and experimental design

This was a hospital-based cross-sectional study conducted in the Department of Physical Medicine and Rehabilitation, National Cheng Kung University Hospital, Dou-Liou Branch, from February 2015 to July 2016. All chronic hemiparetic stroke patients who were receiving rehabilitation in the Department of Physical Medicine and Rehabilitation, and who met the following inclusion criteria, were enrolled in this study: 1st time stroke verified through computed tomography or magnetic resonance imaging of more than 1 year duration; age ≥18 years; ability to understand simple instructions; ability to walk without any assistive device for at least 15 m; and stable neurological and functional status after receiving standard physical therapy for more than 6 months after stroke. Individuals with subarachnoidal bleeding or lacunar infarct without apparent hemiparesis were excluded, along with those with comorbidities that would affect gait. A flowchart illustrating the inclusion and exclusion of the chronic stroke patients is shown in Fig. 1.

Gender, age, body mass index (BMI), the type of stroke, the affected side, the period from onset of stroke, Brunnstrom stage for the affected lower extremity,[16] Berg Balance Scale (BBS) score,[17] timed up and go (TUG) test scores,[18] and history of falls in the previous 12 months were recorded upon entry into the study. There were 14 chronic hemiparetic stroke patients (60.4 ± 15.8 years, 6 women) included in this study. There were 6 patients who had falls in the previous 12 months. The demographics of the chronic stroke patients and the falling characteristics[19] of the fallers are shown in Table 1. History of falling is an established risk factor for recurrent falls[20]; therefore, we compared the demographic data, BBS, TUG, and the cross-correlation and T_s between the magnitude of the acceleration vector of the affected and unaffected foot of fallers (1 or more falls) with those of nonfallers.

The control group was composed of 14 nondisabled young adults (6 women) without falls in the previous 12 months, who received the same evaluations as the chronic stroke patients.

2.2. Ethical considerations

This study was approved by the Institutional Review Board of the National Cheng Kung University Hospital, College of Medicine, National Cheng Kung University (Approval No.: B-ER-103-278 and A-ER-104-268). All participants gave written informed consent to take part in this research.

2.3. Instrumentation

Linear accelerations were measured along vertical (V), anterior–-posterior, and medio-lateral axes using 7 triaxial accelerometers (STMicroelectronics LIS3DHTR, range ±16 gravities). The sampling frequency was 120 Hz. The sensors were attached to the back (the level of the 3rd lumbar spinous process), each lateral knee (3 cm above the lateral epicondyle), each lateral ankle (3 cm above the lateral malleolus), and each foot (2 cm below the head of the 4th metatarsal) using custom-made bulk straps (Fig. 2).[13,14] Two tests were conducted in this study to ensure that all participants were familiar with the test procedures. Each test required the subject to walk more than 15 m on a walkway at a self-selected walking speed. Subjects were allowed to rest for 2 minutes between tests. The acceleration signals of the faster 15-m walking test were selected for data acquisition and processing because a higher self-selected walking speed represents the need to adapt to greater upper body accelerations and results in unstable walking in individuals with functional impairments.[21,22] Because contraction of the ankle plantar-flexor results in foot acceleration and plantar-flexor spasticity results in gait asymmetry in stroke patients,[19,23] the triaxial acceleration signals of each foot were selected for data acquisition and processing.

2.4. Data acquisition and processing

All acceleration signals were filtered through a 4th-order low-pass digital Butterworth filter with the cut-off frequency set at 4 Hz. Heel contact events were determined from peak accelerations for the foot in the anterior–posterior axis.[14,24] Although heel contact of the unaffected foot and then the affected foot was determined for each walking trial, the triaxial acceleration signals of the 5 consecutive and bilateral strides from the middle of each trial were extracted, with each stride normalized to N = 120 data points, which corresponds to 100% of the gait cycle (Fig. 2).

We used the magnitude of the 3-dimensional acceleration vector ($\overrightarrow{a_V}$, $\overrightarrow{a_{AP}}$, $\overrightarrow{a_{ML}}$) from each triaxial accelerometer recording to

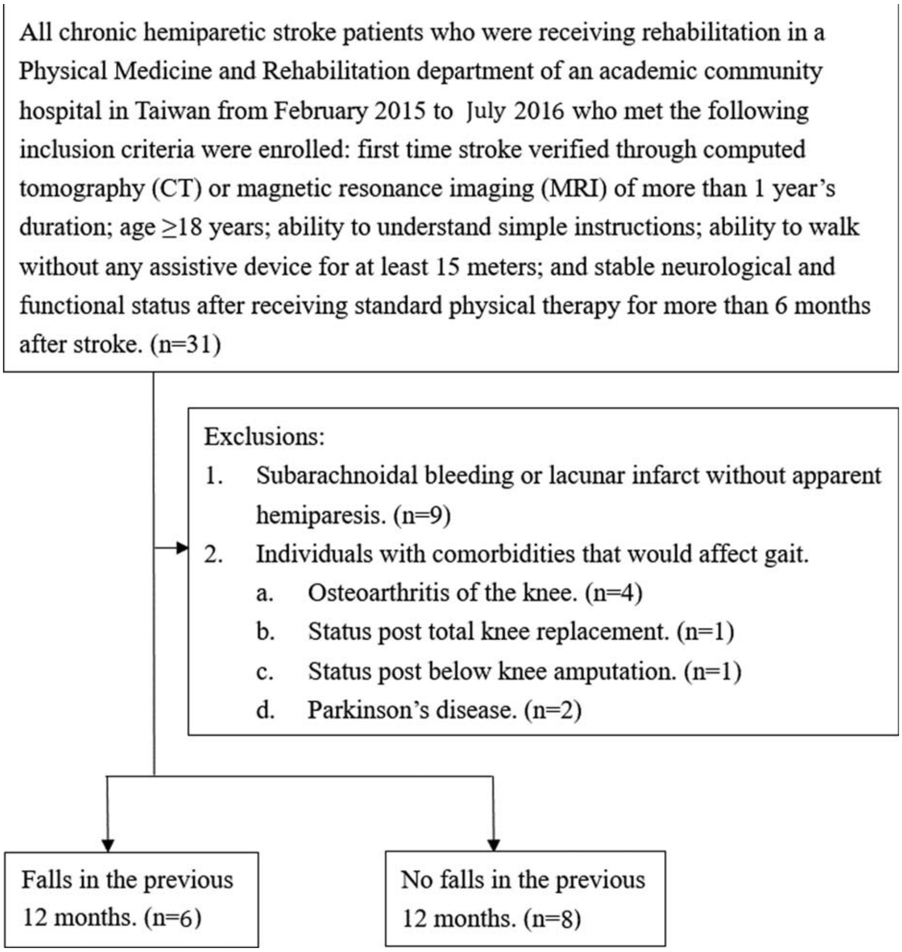


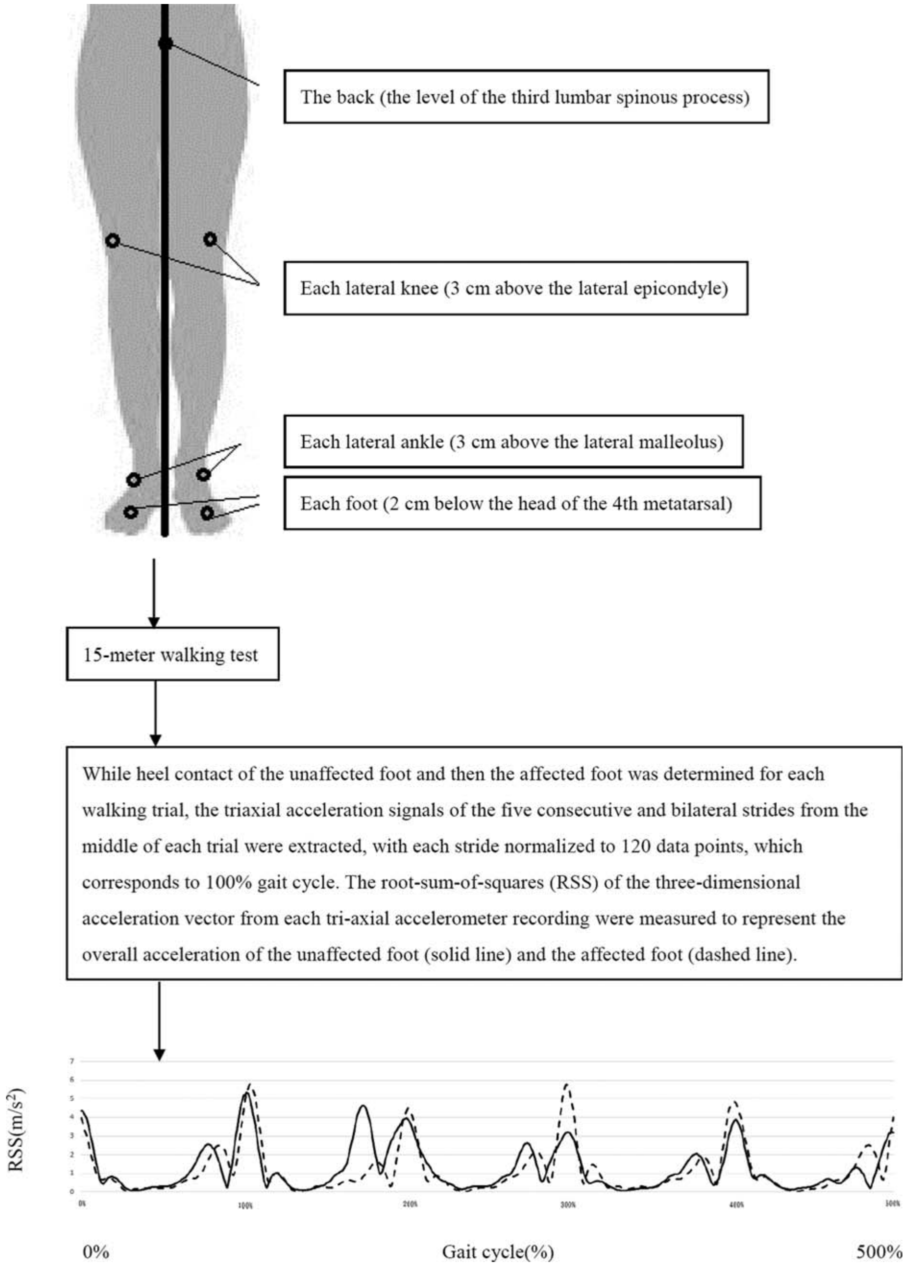
Figure 1. Flow chart illustrating the inclusion and exclusion criteria of the chronic stroke patients.

Table 1

Demographics of the chronic stroke patients and the falling characteristics of the fallers.

Patient	Gender	Age, years	The type of stroke	BMI, kg/m^2	The period from onset of stroke, years	Affected side	Brunnstrom stage for the affected lower extremity	BBS	TUG, secs	Falls in the previous 12 months	Falling characteristics: One-time or repeat faller	Falling characteristics: Place of faller	Falling characteristics: Activity of faller
A	F	74	Cerebral hemorrhage	23.3	1.9	Left	V	46	18.16	Nonfaller	N/A	N/A	N/A
B	M	51	Cerebral infarction	21.3	5.2	Left	V	47	18.56	Nonfaller	N/A	N/A	N/A
C	M	67	Cerebral infarction	19.9	9.6	Left	V	37	33.2	Faller	Repeat faller	Indoors	Walking
D	F	55	Cerebral infarction	19.8	8.1	Right	V	46	19.25	Nonfaller	N/A	N/A	N/A
E	M	48	Cerebral hemorrhage	20.7	10.2	Left	IV	47	19.4	Nonfaller	N/A	N/A	N/A
F	M	43	Cerebral hemorrhage	26.1	3.9	Left	IV	42	26.09	Nonfaller	N/A	N/A	N/A
G	M	80	Cerebral hemorrhage	30.1	8.8	Right	VI	43	28.65	Nonfaller	N/A	N/A	N/A
H	M	77	Cerebral infarction	27.1	8.9	Left	IV	39	32.88	Faller	Repeat faller	Indoors	Walking
I	M	73	Cerebral infarction	27.3	22.1	Left	V	50	16.7	Nonfaller	N/A	N/A	N/A
J	F	50	Cerebral hemorrhage	24.7	1.9	Left	IV	41	28.7	Nonfaller	N/A	N/A	N/A
K	F	79	Cerebral infarction	22.5	2.9	Left	V	31	32.95	Faller	Repeat faller	Indoors	Walking
L	M	71	Cerebral hemorrhage	23.4	1.7	Right	V	41	22.35	Faller	One-time faller	Outdoors	Stairs
M	F	30	Cerebral infarction	18.5	1.2	Right	IV	35	29.04	Faller	Repeat faller	Indoors	Walking
N	F	48	Cerebral hemorrhage	21.5	1.1	Right	V	34	31.74	Faller	One-time faller	Indoors	Walking
Mean		60.4		23.3	6.3			41.4	25.55				
SD		15.8		3.4	5.7			5.6	6.24				

BBS = Berg Balance Scale, BMI = body mass index, F = female, M = male, N/A = not applicable, SD = standard deviation, TUG = timed up and go.



While heel contact of the unaffected foot and then the affected foot was determined for each walking trial, the triaxial acceleration signals of the five consecutive and bilateral strides from the middle of each trial were extracted, with each stride normalized to 120 data points, which corresponds to 100% gait cycle. The root-sum-of-squares (RSS) of the three-dimensional acceleration vector from each tri-axial accelerometer recording were measured to represent the overall acceleration of the unaffected foot (solid line) and the affected foot (dashed line).

Figure 2. Experimental set-up with sensors attached to patient A's back and bilateral lower extremities, and an example of the data processing.

represent the overall acceleration (a) of the unaffected foot (a_{UF}) and affected foot (a_{AF}).[25] The overall acceleration (a) (the root-sum-of-squares [RSS]) was calculated using Eq (1). a_{UF} and a_{AF} are characterized in the following equations:

$$a = RSS = \sqrt{\left(\left|\overrightarrow{a_V}\right|\right)^2 + \left(\left|\overrightarrow{a_{AP}}\right|\right)^2 + \left(\left|\overrightarrow{a_{ML}}\right|\right)^2} \quad (1)$$

$$a_{UF} = a_{UF(1)}, a_{UF(2)}, a_{UF(3)}, \ldots, a_{UF(n)}, \ldots, a_{UF(120)} \quad (2)$$

$$a_{AF} = a_{AF(1)}, a_{AF(2)}, a_{AF(3)}, \ldots, a_{AF(n)}, \ldots, a_{AF(120)} \quad (3)$$

To ensure the uniformity of this study, a_{UF} was regarded as the main reference. The cross-correlation (Cc) between the affected foot (a_{AF}) and unaffected foot (a_{UF}) was computed using the following formula[11,15]:

$$Cc_{(k)} = \sum_{n=1}^{N} a_{UF(n)} a_{AF(n-k)} \qquad \begin{array}{l} k = 0, \pm 1, 2, \ldots, \pm N - 1 \\ \text{if } n - k \le 0 \text{ or } n - k > N \text{ then } a_{AF(n-k)} = 0 \end{array} \quad (4)$$

Cc produces a distinctive peak, which is affected by the magnitude of a_{UF} and a_{AF}. Therefore, auto-correlation of the a_{UF} (AcUF) and a_{AF} (AcAF) are calculated as follows in Eqs (5)–(6) to obtain the normalized Cc (Cc_{norms}). Cc_{norms} produces values ranging from 0 to 1, and values approaching 1 are indicative of a strong correlation between a_{UF} and a_{AF}.

$$\mathrm{AcUF}_{(k)} = \sum_{n-1}^{N} a_{UF(n)} a_{UF(n-k)} \quad \begin{array}{l} k = 0, \pm 1, \pm 2, \ldots \pm N - 1 \\ \text{if } n - k \leq 0 \text{ or } n - k > N \text{ then } a_{AF(n-k)} = 0 \end{array} \tag{5}$$

$$\mathrm{AcAF}_{(k)} = \sum_{n-1}^{N} a_{AF(n)} a_{AF(n-k)} \quad \begin{array}{l} k = 0, \pm 1, \pm 2, \ldots \pm N - 1 \\ \text{if } n - k \leq 0 \text{ or } n - k > N \text{ then } a_{UF(n-k)} = 0 \end{array} \tag{6}$$

$$\mathrm{Cc}_{\mathrm{norms}} = \frac{\max(\mathrm{Cc})}{\sqrt{\mathrm{AxUF}_{(0)} \times \mathrm{AxAF}_{(0)}}} \tag{7}$$

Cc also estimates the T_s between a_{UF} and a_{AF}. T_s is defined as the time it takes for Cc to reach the maximum value.[11,26] Positive T_s indicates that a_{AF} leads a_{UF} and negative T_s indicates that a_{AF} lags behind a_{UF}. T_s was calculated using the following equation and is expressed as the percentage of gait cycle:

$$T_s = \frac{\mathrm{Tmax(Cc)}}{N} \times 100\% \tag{8}$$

For the nondisabled young adults, the overall acceleration of the right foot was regarded as the main reference in order to measure $\mathrm{Cc}_{\mathrm{norm}}$ and T_s.

2.5. Statistical analyses

The clinical characteristics of the chronic stroke patients and the nondisabled young adults, and also the poststroke fallers and nonfallers were compared using the Fisher exact test for categorical variables and the Mann–Whitney *U* test for continuous variables. Mixed-effect models were used to examine the significant differences in $\mathrm{Cc}_{\mathrm{norm}}$ and T_s for repeated measurements within individual subjects. The 5 consecutive measurements of $\mathrm{Cc}_{\mathrm{norm}}$ and T_s were obtained from the 15-m walking test. A *P* value of <0.05 was considered to indicate statistical significance; all tests were 2-tailed. All mathematical calculations were performed with Matlab software (R2015a, Mathworks), and all statistical analyses were performed with the statistical package SAS (Version 9.3, SAS Institute, Cary, NC).

Table 3

Differences in T_s and $\mathrm{Cc}_{\mathrm{norm}}$ between chronic stroke patients and nondisabled young adults using mixed-effects model analysis.

Variable	β	SE	*P*
T_s			
Multivariate			
Chronic stroke patients versus nondisabled young adults	3.449	0.609	<0.001*
Age	−0.039	0.010	<0.001*
Gender	−0.242	0.246	0.328
BBS	0.153	0.068	0.026*
TUG	−0.052	0.059	0.380
$\mathrm{Cc}_{\mathrm{norm}}$			
Multivariate			
Chronic stroke patients versus nondisabled young adults	−0.175	0.034	<0.001*
Age	0.002	0.001	0.002*
Gender	−0.042	0.014	0.004*
BBS	0.013	0.004	<0.001*
TUG	0.007	0.003	0.045*

BBS = Berg Balance Scale, $\mathrm{Cc}_{\mathrm{norm}}$ = normalized cross-correlation, SE = standard error, T_s = time delay, TUG = timed up and go.

* It is significant in statistical comparisons.

3. Results

The mean age of the 14 chronic stroke patients was 60.4 years (range, 30–80 years), and the mean period from onset of stroke was 6.3 years (range, 1.1–22.1 years). The median age of the 14 nondisabled young adults was 24 years (range, 22–27 years). Chronic stroke patient characteristics, BBS scores, and TUG scores are summarized in Table 1. The 5 consecutive measurements of $\mathrm{Cc}_{\mathrm{norm}}$ and T_s (Supplementary file 1, http://links.lww.com/MD/B576) were averaged as the mean $\mathrm{Cc}_{\mathrm{norm}}$ scores and the mean T_s values. Chronic stroke patients had older age, lower BBS scores, higher TUG scores, lower mean $\mathrm{Cc}_{\mathrm{norm}}$ scores, and higher mean T_s values than nondisabled young adults. There were no significant differences between chronic stroke patients and nondisabled young adults in gender and BMI. The clinical characteristics of chronic stroke patients and nondisabled young adults are shown in Table 2. After controlling for age, gender, BBS, and TUG, the mixed-effect model showed that the $\mathrm{Cc}_{\mathrm{norm}}$ in nondisabled young adults was higher than in chronic stroke patients ($\beta = -0.175$; standard error [SE] = 0.034; *P*-value < 0.001). In addition, the model revealed that T_s in chronic stroke patients was higher than in nondisabled young adults ($\beta = 3.449$; SE = 0.609; *P*-value < 0.001). The results of the mixed-effects model for chronic stroke patients and nondisabled young adults are summarized in Table 3. Poststroke fallers had lower BBS scores, higher TUG scores, lower mean $\mathrm{Cc}_{\mathrm{norm}}$ scores, and higher mean T_s values than nonfallers. There were no significant differences between poststroke fallers and nonfallers in age, gender, BMI, the type of stroke, period from onset of stroke, affected side, or Brunnstrom stage. The clinical characteristics of poststroke fallers and nonfallers are shown in Table 4. After controlling for age, gender, BBS, and TUG, the mixed-effect

Table 2

Clinical characteristics of chronic stroke patients and nondisabled young adults.

Variables	Chronic stroke patients (n = 14)	Nondisabled young adults (n = 14)	*P*
Age, years	61 (30–80)	24 (22–27)	<0.001*
Men	8 (57.1%)	8 (57.1%)	1.000
BMI, kg/m^2	22.9 (18.5–30.1)	22.0 (19.2–24.5)	0.427
BBS	42 (31–50)	56 (56–56)	<0.001*
TUG test, secs	27.37 (16.70–33.20)	8.64 (7.13–9.72)	<0.001*
Mean $\mathrm{Cc}_{\mathrm{norm}}$	0.7594 (0.7099–0.9150)	0.9764 (0.9396–0.9913)	<0.001*
Mean T_s, %	−0.0833 (−4.3333–0)	0 (0–0)	0.024*

Data are presented as medians (ranges) or numbers of subjects (percentages). BBS = Berg Balance Scale, BMI = body mass index, $\mathrm{Cc}_{\mathrm{norm}}$ = normalized cross-correlation, T_s = time delay, TUG = timed up and go.

* It is significant in statistical comparisons.

Table 4

Clinical characteristics of poststroke fallers and nonfallers.

Variable	Fallers (n = 6)	Nonfallers (n = 8)	*P*
Age, y	69 (30–79)	53 (43–80)	0.852
Men	3 (50%)	5 (62.5%)	1.000
Cerebral infarction	4 (66.7%)	3 (37.5%)	0.592
BMI, kg/m^2	22.0 (18.5–27.1)	24.0 (19.8–30.1)	0.414
Duration poststroke, years	2.3 (1.1–9.6)	6.7 (1.9–22.1)	0.228
Left hemiparesis	3 (50%)	6 (75%)	0.580
Brunstromm stage IV or below	2 (33.3%)	3 (37.5%)	1.000
BBS	36 (31–42)	46 (41–50)	0.001*
TUG test, secs	32.31 (22.35–33.20)	19.33 (16.70–28.70)	0.005*
Mean Cc_{norm}	0.7335 (0.7099–0.7752)	0.8190 (0.7447–0.9150)	0.020*
Mean T_s, %	−2.2500 (−4.3333–1.0000)	0 (−0.1667–0)	0.001*

Data are presented as medians (ranges) or numbers of subjects (percentages). BBS = Berg Balance Scale, BMI = body mass index, Cc_{norm} = normalized cross-correlation, T_s = time delay, TUG = timed up and go.

* It is significant in statistical comparisons.

model showed that the Cc_{norm} in nonfallers was higher than in fallers ($\beta = -0.093$; SE = 0.029; *P*-value = 0.002). In addition, the model revealed that T_s in fallers was higher than in nonfallers ($\beta = -1.900$; SE = 0.719; *P*-value = 0.011). The results of the mixed-effects model for poststroke fallers and nonfallers are summarized in Table 5.

4. Discussion

Bilateral lower extremity accelerometry has been shown to be a reliable and valid tool for measuring the spatiotemporal parameters of the gait cycle in people with hemiparesis.[14] In this study, we found that the cross-correlation and T_s between the affected and unaffected lower extremities may be useful indicators to distinguish chronic stroke patients from nondisabled young adults and poststroke fallers from nonfallers.

Gait deficits greatly contribute to functional disability after stroke.[27] Gait deficits and balance deficits are important risk factors for falls after stroke.[4] Gait deficits include gait asymmetry, reduced stability during the stance phase, and reduced propulsion at push-off.[4,28] Balance deficits include reduced postural stability during quiet standing and less coordinated responses to both external and self-induced balance perturbations.[4] Cross-correlation between the signals of the inertial sensors of the bilateral lower extremities has been shown to be a good indicator of gait symmetry.[11,15] This could explain our finding that the cross-correlation between the affected and unaffected lower extremities was related to stroke and fall history in stroke patients.

Table 5

Differences in T_s and Cc_{norm} between poststroke fallers and nonfallers using mixed-effects model analysis.

Variable	β	SE	*P*
T_s			
Multivariate			
Fallers versus nonfallers	−1.900	0.719	0.011*
Age	−0.029	0.012	0.016*
Gender	−0.095	0.512	0.853
BBS	−0.048	0.133	0.717
TUG	−0.105	0.081	0.198
Cc_{norm}			
Multivariate			
Fallers versus nonfallers	−0.093	0.029	0.002*
Age	0.002	0.001	<0.001*
Gender	−0.068	0.020	0.002*
BBS	0.008	0.005	0.139
TUG	0.007	0.003	0.029*

BBS = Berg Balance Scale, Cc_{norm} = normalized cross-correlation, SE = standard error, T_s = time delay, TUG = timed up and go.

* It is significant in statistical comparisons.

The propulsive forces required for gait progression are mainly generated during the push-off phase.[4,29] Stroke-related gait deviations during this phase can thus be expected to be responsible for the lack of progression.[29,30] The lack of progression may result in the T_s between the affected and unaffected lower extremities. This could explain our finding that the T_s between the affected and unaffected lower extremities was related to stroke and fall history in stroke patients.

Stroke-related balance deficits comprise reduced postural stability during quiet standing and less coordinated responses to both external and self-induced balance perturbations.[4] BBS is used extensively in geriatric medicine to examine balance and has good discriminative ability to predict falls in elderly persons.[17] The TUG test is also widely used in geriatric medicine to examine gait speed and basic functional mobility.[18] In this study, BBS and TUG scores were measured as confounding factors and controlled for in the multivariate mixed-effect models to evaluate the relationship between gait symmetry and fall history.

Studies have shown that task-specific exercise programs that target balance and gait deficits can drive neural plasticity[31,32] and reduce the number of falls in individuals who have suffered a stroke.[33] Technological advances in assistive devices, such as walking aids and ankle-foot orthosis, have also proven to be valuable in preventing falls after stroke.[27,34] However, more longitudinal studies are needed to provide conclusive evidence of these interventions regarding the prevention of falls in such individuals.

Some limitations of this study must be acknowledged. First, because this was a cross-sectional study, a causal effect of gait symmetry on fall history cannot be claimed. It thus remains unclear whether the faller's gait deficits contributed to falls, or whether the consequence of falling resulted in modified gait patterns. Longitudinal studies are warranted to better understand the relationship between gait symmetry and risk factors for falling in chronic stroke patients. Second, an acceleration vector is a combination of direction or orientation in 3-dimensional geometry and (nonnegative) magnitude, such as RSS as measured in this study. The real T_s from 3 orthogonal axes might be underestimated by transforming 3-dimensional acceleration signals to RSS. Third, the inertial sensors used in this study did not contain gyroscopes as direction indicators, and future research on inertial sensors with gyroscopes is thus suggested.

Fourth, the mobility function is associated with environmental barriers.[35] Future research taking perceived or objective environmental barriers as a confounding factor are therefore warranted to better understand the relationship between mobility function and fall risk in chronic stroke patients. Finally, all of the chronic stroke patients were recruited from a single rehabilitation center at an academic community hospital, which limits the generalizability of our results to other populations of chronic stroke patients.

In conclusion, the cross-correlation and T_s between the affected and unaffected lower extremities may be useful indicators to distinguish poststroke fallers from nonfallers.

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