



Original Article

The relationship between trunk acceleration parameters and kinematic characteristics during walking in patients with stroke

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Abstract. [Purpose] Limited literature has investigated the relationships between acceleration-based gait characteristics and kinematic information from motion analysis systems in gait analysis. The purpose of this study is to determine whether acceleration-based gait characteristics were associated with gait characteristics by motion analysis systems in patients with stroke. [Participants and Methods] Seventeen patients with stroke walked along a 10-m-long walkway at their comfortable speed. Trunk acceleration was measured with an accelerometer. Several reflective markers over bony landmarks on the lower extremities were used to capture movements. We evaluated the correlations of variables calculated between the trunk accelerometers and the motion analysis system. [Results] Walking speed was positively correlated with harmonic ratios along the anteroposterior axis and stride regularity along the vertical and anteroposterior axes. Harmonic ratios were associated with the stance phase percent on the unaffected side. Stride regularity was associated with the stance phase percent on both sides. Smaller interstride variability was associated with smaller peak ankle plantarflexion during both phases and greater peak ankle dorsiflexion during swing phase. Stride regularity is positively associated with maximal knee flexion during swing phase. [Conclusion] Relationships with spatiotemporal and joint kinematic parameters from the motion analysis system support the potential use of accelerometers.

Key words: Gait, Accelerometers, Motion analysis

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INTRODUCTION

Patients with stroke often accompany with gait deficits. Gait analysis, thus, is critical for clinicians to diagnose gait disorders, identify potential neuromuscular impairments, and establish appropriate treatment program. Gait impairments could be detected through gait analysis in spatiotemporal and joint kinematics gait parameters. One way to perform gait performance is observational gait analysis, which is commonly used in the clinic, but several concerns exist: First, observational gait analysis has relatively low accuracy^{1, 2)}. Second, it is difficult for humans to observe high-speed movements¹⁾. Third, participants need to make repeated walks for observers to watch from several views, which requires huge time and sufficient endurance. With the shortcomings of observational gait analysis, objective measurements should be used during gait analysis.

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The accurate objective assessment of gait characteristics can be obtained from advanced instrumented 3-dimensional motion analysis systems, such as VICON, which is considered the gold standard tool for gait analysis. Several studies have used VICON to assess the gait performance in patients with stroke^{3, 4}. In terms of joint kinematics gait characteristics, typical gait deviation involves inadequate propulsion during the pre-swing phase and decreased knee flexion and ankle dorsiflexion of the paretic limb during the swing phase⁵⁻⁷. In spatiotemporal gait characteristics, typical gait deviation involves decreased walking speed, increased swing time, and decreased single limb support time in the paretic limb^{6, 8}. Patients with stroke also have increased variability of spatiotemporal gait parameters compared with healthy controls^{6, 9}. The parameters derived from the motion analysis system have been well recognized in both clinical application and research purpose^{6, 8-10}. Some drawbacks, however, involve complicated application, relatively high cost, and use restricted to a specific laboratory environment.

The portable, inexpensive, and easy-to-use body-worn accelerometers has risen for objective gait analysis¹¹⁻¹⁴. Several studies used data from trunk accelerometers to infer gait characteristics for 3 derivatives: (1) harmonic ratio, (2) interstride variability, and (3) stride regularity. Harmonic ratio is considered a measure of gait stability, symmetry, and smoothness of trunk motion¹⁵⁻¹⁷; Interstride variability indicates gait variation¹⁸; and stride regularity is a measure of gait symmetry¹⁹. These 3 derivatives were reported to differentiate gait characteristics between populations or various conditions of a specific population and associate with clinical performance. For example, harmonic ratios are lower in subacute stroke patients than healthy adults²⁰. Harmonic ratios are correlated with trunk control assessed by the Trunk Impairment Scale²¹. The interstride variability is higher in stroke patients who reported fall history than those who never experienced falls¹⁸. The stride regularity is more irregular in the elderly than young participants²². Furthermore, step regularity is correlated with Tinetti-balance and Tinetti-gait scores in elderly persons²³. Performance in gait speed, step, and stride regularity is better in the elderly than elderly subjects with increased fall risks²³.

Although the trunk accelerometers demonstrate discriminant validity, limited investigations look for the possible relations of trunk accelerometers with kinematic gait characteristics by motion analysis systems. There is only limited literature investigating the relationship between trunk acceleration gait characteristics and kinematic information²⁴. That study shows that the root mean square accelerations are associated with several combinations of kinematic characteristics with lower extremity joints during moderate treadmill running in healthy participants²⁴. Whether similar relationships exist in patients with stroke is unclear. Therefore, this study aimed to determine whether trunk acceleration gait characteristics were associated with any gait characteristics by motion analysis systems in patients with stroke.

PARTICIPANTS AND METHODS

This cross-sectional study was conducted at the Chang Gung Memorial Hospital in Taiwan. The Chang Gung Medical Foundation Institutional Review Board approved the study (IRB No. 103-3564A3), and all participants signed a written informed consent form.

The study recruited 17 patients with unilateral stroke. The inclusion criteria were (1) diagnoses of single unilateral stroke, (2) more than 6 months post-stroke, and (3) able to walk with or without physical assistance. The exclusion criteria were (1) other neurologic or psychiatric diseases, and (2) walking with an ankle-foot orthosis. Walking ability was assessed by Functional Ambulation Classification (FAC)²⁵.

Participants were fitted with the accelerometer apparatus at the level of the L3. Reflective markers were placed over several bony landmarks on the lower extremities, including (1) the seventh cervical vertebra (C7), (2) the fourth lumbar vertebra (L4), (3) bilateral anterior superior iliac spine (ASIS) and posterior superior iliac spines, (4) bilateral medial and lateral epicondyles of the femurs, (5) bilateral tibial tuberosities, (6) bilateral medial and lateral malleoli, (7) bilateral big toes, and (8) bilateral heels. Participants were asked to walk along a 10-m-long walkway at their comfortable walking speed for 3 trials. Adequate rest was allowed between trials. Only the designated 7-m trial distance was analyzed and the first and last 2 to 3 steps were excluded.

A 7-camera VICON MX motion analysis system (VICON MX; Oxford Metrics Inc., Oxford, UK) was used to capture the movement of markers. Movements were recorded at 120 Hz and digitally low-pass filtered at 5 Hz using a second-order dual-pass Butterworth filter.

One accelerometer (Actigraph wGT3X-BT, Pensacola, FL, USA) placed at the level of the L3 was used to measure the linear acceleration of the trunk along vertical, anteroposterior, and mediolateral axes during gait. The data were collected at a sample rate of 100 Hz, and processed with ActiLife software (Actigraph, Pensacola, FL, USA).

Data collected from the motion analysis system and the accelerometer were analyzed using Matlab software (MathWorks, Natick, MA, USA). Parameters calculated are as follows:

Data from the motion analysis system were used to calculate walking speed and the stance phase percentage and step length of the affected and unaffected sides. Average speed was computed as the speed of the marker on the ASIS.

We focused on the sagittal joint kinematic gait parameters of the affected side because the deviations of both stance and swing phase occur mainly on the affected side. The parameters during 1 gait cycle calculated included maximal trunk flexion, maximal trunk extension, maximal hip flexion, maximal hip extension, maximal and minimal knee flexion, maximal ankle dorsiflexion, and plantarflexion during the stance and swing phases of the affected side.

All of the trunk acceleration parameters were calculated during 10 steps. At first, the peak of anteroposterior acceleration

Table 1. Demographic and clinical background characteristics of participants (N=17)

Age (years)	57.8 ± 12.7
Female	5 (29.4%)
Time post-stroke (months)	18.2 ± 11.2
Affected side	
Right	5 (29.4%)
Functional Ambulation Category	
1 (Dependent for physical assistance level II)	1 (5.9%)
2 (Dependent for physical assistance level I)	3 (17.7%)
3 (Dependent for supervision)	0
4 (Independent level surfaces only)	2 (11.8%)
5 (Independent)	11 (64.7%)
Spatio-temporal	
Walking speed (m/s)	0.6 ± 0.2
Stance phase (%) in 1 gait cycle	
Affected side	68.0 ± 7.7
Unaffected side	73.2 ± 6.0
Step length (cm)	
Affected side	46.0 ± 12.5
Unaffected side	42.3 ± 17.0
Joint kinematic (degrees)	
Stance phase of the affected side	
Maximal trunk flexion ^a	10.4 ± 5.7
Maximal trunk extension ^a	4.5 ± 3.2
Maximal hip flexion	17.6 ± 4.3
Maximal hip extension	16.2 ± 7.0
Maximal knee flexion	35.4 ± 12.6
Minimal knee flexion	2.7 ± 4.4
Maximal ankle dorsiflexion	0.4 ± 7.0
Maximal ankle plantarflexion	24.0 ± 9.6
Swing phase of the affected side	
Maximal trunk flexion ^a	9.3 ± 4.5
Maximal trunk extension ^a	6.4 ± 4.1
Maximal hip flexion	20.4 ± 4.3
Maximal hip extension	3.2 ± 6.9
Maximal knee flexion	40.5 ± 15.0
Minimal knee flexion	6.1 ± 4.6
Maximal ankle dorsiflexion	-10.8 ± 6.5
Maximal ankle plantarflexion	21.1 ± 11.9
Accelerometers	
Harmonic ratio	
Vertical	1.0 ± 0.3
Anteroposterior	0.9 ± 0.2
Interstride variability (Coefficient of variability)(%)	
Vertical	9.9 ± 2.6
Anteroposterior	9.3 ± 3.1
Stride regularity	
Vertical	0.3 ± 0.2
Anteroposterior	0.4 ± 0.2

Data are expressed as mean ± SD for continuous variables and as frequency distribution (%) for categorical variables.

^a: Maximal trunk flexion/extension was calculated based on the maximal movements of the marker over C7 in relation to the marker over L4.

was used to identify the start of the step (heel strike)⁵). Trunk acceleration parameters calculated were:

The harmonic ratio indicates the smoothness and rhythm of acceleration patterns^{7, 16}). Briefly, the harmonic ratio is based on the idea that the unit of measurement from a continuous walking trial is a stride (2 steps). By using a finite Fourier series, a harmonic ratio is calculated by dividing the sum of the amplitudes of the first 10 even harmonics by the sum of the amplitudes of the first odd harmonics. We focused on the anteroposterior and vertical harmonic ratios because the mediolateral harmonic ratios do not differ between patients with stroke, age-matched cohorts, and young adults⁵).

The interstride variability is the coefficient of variation of the root mean square acceleration signals of each of the 5 strides¹⁸). A higher coefficient of variation indicates a greater interstride variability.

The stride regularity^{19, 26}) is obtained by the autocorrelation function of trunk acceleration, which defines the correlation between the value at a time $t=i$ and the value $t=i+j$, where j indicates a time lag¹⁹). The second peak was obtained as the stride regularity because 1 step contributes a cycle (phase) and each stride contains 2 steps. The first peak was considered the step symmetry^{19, 22}).

Statistical analyses were performed using SAS 9.4 software (SAS Institute Inc., Cary, NC, USA). Data are expressed as mean ± standard deviation for continuous variables and as frequency distribution (%) for categorical variables. The relationships between the trunk acceleration parameters and spatiotemporal/joint kinematic parameters were evaluated by using the Pearson correlation coefficient. Statistical significance was set at $\alpha=0.05$.

RESULTS

The characteristics of demographic data are summarized in Table 1. Approximately three-fourths of the participants walked independently (FAC Level IV and V). The averaged comfortable walking speed was 0.6 m/s. The averaged percentage of the stance phase in 1 gait cycle was 68.00 for the affected side and 73.2 for the unaffected side. The averaged step length of affected and unaffected sides was 46.0 and 42.3 cm, respectively.

The averaged joint angles during the stance and swing phase of the affected side were shown in Table 1. Here, the average maximal ankle range of motion never reached a neutral position of the ankle joint, with a peak dorsiflexion of -10.8° (equaling plantarflexion of 10.8°). The harmonic ratios, interstride variability, and stride regularity were shown in Table 1.

The correlation between the trunk acceleration and spatio-temporal parameters was shown in Table 2. Walking speed was positively correlated with the harmonic ratio along the anteroposterior ($r=0.62$, $p=0.01$) but not the vertical axis. The stance phase percentage in 1 gait cycle for the affected side was not related to the harmonic ratios. However, the stance phase percentage for the unaffected side was moderately and negatively related to both vertical and anteroposterior ($r= -0.55$ to -0.61 , $p<0.02$ for both) harmonic ratios. The step length of the affected side was

Table 2. The correlation between the trunk acceleration and spatio-temporal parameters

Parameter	Harmonic ratio				Interstride variability				Stride regularity			
	Vertical		Anteroposterior		Vertical		Anteroposterior		Vertical		Anteroposterior	
	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>
Walking speed (m/s)	0.28	0.28	0.62	0.01*	0.18	0.5	0.41	0.1	0.84	<0.01*	0.61	0.01*
Stance phase (%) in one gait cycle												
Affected side	0.03	0.9	-0.06	0.81	0.09	0.72	-0.07	0.79	-0.50	0.04*	-0.51	0.04*
Unaffected side	-0.55	0.02*	-0.61	0.01*	-0.02	0.93	-0.23	0.38	-0.65	0.01*	-0.54	0.03*
Step length (cm)												
Affected side	0.67	0.003*	0.43	0.09	0.04	0.89	0.47	0.06	0.39	0.12	0.46	0.07
Unaffected side	0.25	0.34	0.28	0.28	-0.2	0.44	0.28	0.28	0.65	0.01*	0.57	0.02*

p* value<0.05.Table 3.** The correlation between the trunk acceleration and joint kinematic parameters during the stance phase of the affected side

Parameter	Harmonic ratio				Interstride variability				Stride regularity			
	Vertical		Anteroposterior		Vertical		Anteroposterior		Vertical		Anteroposterior	
	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>
Stance phase of the affected side (degrees)												
Maximal hip flexion	0.37	0.15	0.25	0.34	0.12	0.64	0.09	0.73	0.14	0.60	-0.01	0.97
Maximal hip extension	0.02	0.94	0.04	0.87	-0.11	0.67	0.31	0.23	0.43	0.08	0.47	0.06
Maximal knee flexion	0.38	0.14	0.33	0.2	-0.15	0.57	0.13	0.63	0.46	0.06	0.27	0.29
Minimal knee flexion	-0.06	0.81	-0.02	0.94	-0.02	0.95	-0.16	0.55	-0.05	0.87	-0.12	0.66
Maximal ankle dorsiflexion	-0.31	0.22	-0.2	0.43	-0.37	0.15	-0.21	0.41	0.26	0.31	0.22	0.40
Maximal ankle plantarflexion	0.04	0.88	0.08	0.76	0.63	0.01*	0.27	0.3	0.33	0.20	0.29	0.25

p* value<0.05; All joint angles of affected limbs.Table 4.** The correlation between the trunk acceleration and joint kinematic parameters during the swing phase of the affected side

Parameter	Harmonic ratio				Interstride variability				Stride regularity			
	Vertical		Anteroposterior		Vertical		Anteroposterior		Vertical		Anteroposterior	
	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>	<i>r</i>	<i>p</i>
Swing phase of the affected side (degrees)												
Maximal hip flexion	0.34	0.18	0.52	0.03*	0.42	0.09	0.51	0.04*	0.44	0.08	0.21	0.42
Maximal hip extension	-0.14	0.6	-0.09	0.74	-0.2	0.45	0.12	0.64	0.32	0.22	0.45	0.07
Maximal knee flexion	0.35	0.17	0.43	0.09	-0.06	0.83	0.27	0.29	0.61	0.01*	0.32	0.21
Minimal knee flexion	-0.26	0.32	-0.21	0.42	-0.21	0.42	0.01	0.96	0.15	0.58	0.04	0.88
Maximal ankle dorsiflexion	-0.12	0.65	-0.14	0.59	-0.63	0.01*	-0.3	0.24	0.18	0.50	0.05	0.86
Maximal ankle plantarflexion	0.2	0.45	0.16	0.54	0.6	0.01*	0.34	0.18	0.20	0.43	0.12	0.66

**p* value<0.05; All joint angles of affected limbs.

moderately and positively correlated with harmonic ratio along the vertical axis ($r=0.67$, $p=0.003$) but not the anteroposterior axis. The step length of the unaffected side, however, was not correlated with the harmonic ratio. Moreover, walking speed, stance phase percentage in 1 gait cycle, and step length were not related to the interstride variability ($p>0.06$ for all). Furthermore, walking speed was positively correlated with stride regularity along both the vertical ($r=0.84$, $p<0.01$) and anteroposterior axis ($r=0.61$, $p=0.01$). The stance phase percentage of both sides was correlated to stride regularity ($r=-0.50$ to -0.65 , $p<0.04$ for both). In addition, the step length of the unaffected side ($r>0.57$, $p<0.02$) but not the affected side was significantly related to the stride regularity.

The correlation between the trunk acceleration and joint kinematic parameters during the stance phase of the affected side was shown in Table 3. The harmonic ratio was not correlated to any angles. The interstride variability was not correlated to any joint angles, with the exception of the significant relationship between maximal ankle plantarflexion and the interstride variability along the vertical axis ($r=0.63$, $p=0.01$). Further, the stride regularity was not correlated to any joint angles.

The correlation between the trunk acceleration and joint kinematic parameters during the swing phase of the affected

side was shown in Table 4. Harmonic ratios were not correlated to the any joint angles, with the exception of the positive relationship between maximal hip flexion and harmonic ratio along the anteroposterior axis ($r=0.52$, $p=0.03$). Moreover, the interstride variability along the anteroposterior axis was positively correlated to the maximal hip flexion ($r=0.51$, $p=0.04$) but not the maximal hip extension. The interstride variability and maximal minimal knee flexion were not correlated. However, maximal ankle dorsiflexion ($r= -0.63$, $p=0.01$) and plantarflexion ($r=0.6$, $p=0.01$) were separately related to the vertical interstride variability. Furthermore, the stride regularity was not correlated to any joint angles, with the exception of the relationship between the maximal knee flexion and stride regularity along the vertical axis ($r=0.61$, $p=0.01$).

DISCUSSION

This study provides more evidence of the potential use of accelerometers for gait analysis by showing that the spatiotemporal and joint kinematic parameters were related to parameters calculated from trunk accelerations. First, walking speed was positively correlated with the harmonic ratio along the anteroposterior axis and stride regularity. The stance phase percentage in 1 gait cycle of the unaffected side was negatively correlated with harmonic ratios along both axes separately. The stance phase percentage of both sides and stride regularity were negatively correlated. The step length of the affected side was associated with the harmonic ratio along the vertical axis, whereas that of the unaffected side was associated with stride regularity along both vertical and anteroposterior axes. Second, during the stance phase of the affected side, the maximal ankle plantarflexion was associated with the interstride variability along the vertical direction. Furthermore, during the swing phase of the affected side, the maximal hip flexion was positively associated with the harmonic ratio along anteroposterior axis and with interstride variability along anteroposterior axis. A higher stride regularity was associated with greater peak knee flexion. The maximal ankle dorsiflexion and plantarflexion was associated with interstride variability along the vertical axis.

The existed relationships between parameters from motion analysis systems and parameters from trunk accelerations in patients with stroke not only support the potential use of accelerometers but provide the possibility of prediction of the spatiotemporal and joint kinematic parameters by trunk accelerations. For example, 70% of the variance in walking speed can be explained by stride regularity along the vertical axis (Table 2, $r=0.84$, $r^2=0.7$). Furthermore, 36–40% of the variance in peak ankle range of motion can be explained by the interstride variability along the vertical direction (Maximal ankle dorsiflexion: $r= -0.63$, $r^2=0.4$; Maximal ankle plantarflexion: $r=0.6$, $r^2=0.36$).

Harmonic ratios are lower in stroke patients with low function than the age-matched group and healthy young people⁵. This study indicated that harmonic ratios along the anteroposterior axis but not the vertical axis were positively related to walking speed, consistent with previous findings⁵. Harmonic ratios were not correlated with any joint angles, except for the maximal hip flexion during the swing phase of the affected side. This could be because the harmonic ratio is a global measure during a period and does not count for specific phases¹⁵. In addition, the participants in this study with higher harmonic ratios tended to demonstrate greater maximal hip flexion during the swing phase. The peak hip flexion occurs in midswing as a peak of about 19° of flexion²⁷. A decrease of the peak hip flexion in the swing phase could contribute to the decreased step length of the affected side, supported by a tendency of the positive relationship between the harmonic ratio and the step length of the affected side ($p=0.09$, Table 2). A decrease of the peak hip flexion can be compensated by trunk and pelvic movements, which are commonly seen in patients with stroke. Because the harmonic ratio is also a predictor of falls 1 year after the assessment²⁸), we suggest that the decreased hip flexion and the following compensatory strategies may be related to the falls.

We intended to assess interstride variability with the coefficient of variation but not the standard deviation because the coefficient of variation has been normalized to its mean and does not depend on the measurement unit. The interstride variability in this study was not related to any spatial-temporal parameters but was related to several joint kinematic parameters with hip and ankle joints, including the maximal ankle plantarflexion during the stance phase and the maximal hip flexion, ankle dorsiflexion, and plantarflexion during the swing phase. The most relationships existed with the ankle joint. More specifically, an increase of interstride variability (poor gait stability) was associated with an increase of maximal ankle plantarflexion during the stance phase, a decrease of maximal ankle dorsiflexion, and an increase of maximal ankle plantarflexion during the swing phase. Excessive ankle plantarflexion/decreased ankle dorsiflexion may cause drop foot and gait disturbances. Furthermore, ankle impairments are shown to be linked to decreased gait velocity and gait asymmetry in stroke patients²⁹). We suggest that interventions targeted at the ankle joint may be necessary to apply to people with higher interstride variability after stroke.

Participants in this study demonstrated poorer stride regularity than those in previous studies¹⁸). Participants with better stride regularity demonstrated faster walking speed, decreased stance phase percentage in 1 gait cycle, and enlarged step length of the unaffected side, all of which indicate better gait and balance performance. The capacity to walk with a shorter stance phase and larger step length of the unaffected side may imply that participants can demonstrate longer single limb support time of the affected side. Furthermore, trunk accelerations and joint kinematic parameters during the stance phase of the affected side was not related. The only relationship with joint kinematic parameters occurred with maximal knee flexion during the swing phase. One commonly observed swing phase kinematic deviation is decreased peak knee flexion during the swing phase²⁷), and the potential cause of decreased knee flexion involves decreased peak hip extension in late stance phase²⁷). Our results showed a tendency of a positive relationship between stride regularity along both axes and maximal hip extension in the stance phase ($p=0.06$ to 0.08), supported by the previous findings²⁷).

We did not observe relationships between parameters calculated from trunk acceleration signals and joint kinematic parameters with hip and knee joints during the stance phase of the affected side. The reason could be due to minimal gait deviations over these 2 joints in our participants (Table 1). The other reason could be because different abnormal gait patterns in opposite movement directions can occur during stance phase, making it difficult to have a linear relationship. For example, abnormal gait patterns with excessive knee flexion or knee hyperextension can be observed during the stance phase because weak quadriceps can induce inability to full knee extension or a hyperextended knee.

This study has a few limitations. First, our sample is relatively small. Further, our participants did not show knee flexion deviations or demonstrate decreased plantarflexion of the affected side during the stance phase, both of which are commonly observed in patients with stroke. Future studies should include a large number of patients with various gait patterns³⁰). Second, joint kinematic parameters focused on joint angles of the affected side. Further studies are needed to investigate whether the similar relationships exist in the unaffected side. Third, approximately 25% participants needed physical assistance during ambulation (FAC Level I and II), which may have influenced the accelerations measured during gait. Further studies should investigate whether the relationships differ in stroke patients who need different levels of assistance.

Relationships with spatiotemporal and joint kinematic parameters support the potential use of accelerometers in patients with stroke. Our findings might prove to be helpful for clinicians who wish to use accelerometers to assess gait performance in patients with stroke. Accelerometers could be provided to clinicians or researchers as a quantitative human gait analysis tool for routine use.

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Conflicts of interest

None.

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