Reliability and Validity of Bilateral Thigh and Foot Accelerometry Measures of Walking in Healthy and Hemiparetic Subjects

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Objective. Measures of walking ability in large clinical trials are usually limited to a timed short-distance walk and the distance walked in a fixed time. A new integrated system of 5 accelerometers was tested for reliability and compared to a footswitch system to determine if the accelerometers offered a practical option for the acquisition of spatiotemporal gait parameters. *Methods.* Leg accelerations and decelerations were defined in relation to simultaneous kinematic and electromyographic data acquired from a healthy subject. Eight healthy adults and 6 independent ambulators with hemiparetic stroke walked 15 m at 2 different speeds wearing both the accelerometers and footswitches. Twelve healthy subjects walked at 5 different speeds repeated 3 times on each of 2 days wearing the accelerometers. Walking speed, cadence, stride length, and single- and double-limb support, swing, and stance times were calculated. *Results.* No differences (*t* test, $P > 0.2$) were found between footswitch and accelerometer variables when comparing all left or right legs in healthy subjects and all paretic or unaffected legs in stroke subjects. A 2-way nested ANOVA model (speed, left and right legs, trial, and session) with the accelerometers at walking speeds from 0.5 to 1.8 m/s revealed high reproducibility of all measures. *Conclusions.* The accelerometry system provided reliable and valid spatiotemporal measures of gait for the upper range of speeds likely to be targeted for rehabilitation interventions in ambulatory subjects.

Key Words: *Gait analysis—Accelerometry—Stroke rehabilitation—Locomotion—Walking.*

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Reliable and clinically meaningful assessment and outcome measures are one of the critical needs for trials of rehabilitation interventions to improve walking in patients with stroke, spinal cord injury, multiple sclerosis, degenerative neurological diseases, and other conditions such as frailty and orthopedic diseases.[1-3]

Practical and reliable scales of walking ability after stroke for multicenter clinical trials have been limited to clocked walking speed over 6 to 30 m, distance walked in 2 to 12 min, and ordinal scales of levels of assistance such as the Functional Independence Measure or the Rivermead Mobility Index.[4-7] Walking speed generally correlates with the capability of disabled subjects after stroke to walk in the home or community.[8,9]

Walking speed subsumes many of the observable components of gait and has been considered an overall surrogate for spatiotemporal features of the gait pattern.[10] For single site studies of small numbers of subjects, the spatiotemporal elements of the stance and swing phases of the gait cycle have been specifically assessed using footswitches, walkways that show footprints or are embedded with pressure sensors, and potentiometers attached to each ankle to measure anterior-posterior displacement during stepping. Video-based motion analysis systems add kinematics in 3 planes to spatiotemporal measures and can incorporate kinetics using force plates, but their expense and complexity have limited their utility for large trials. All of these tools must be employed in a laboratory setting over walkways that test only 6 to 12 step cycles. The ecological validity of walking tests performed on a short walkway, rather than during home and community activities, is moot.[9] Step counters and pedometers that are sensitive to vertical movements or stride length are convenient, general measures of walking activity outside of a laboratory, but they are limited to counting the number of strides.[11,12] A low-inertia, triaxial piezoresistant accelerometer placed over the lumbar spine has recorded horizontal-vertical movements that an algorithm converted to measures of walking speed, step length, and balance.[13] Another device used 2 accelerometers, one aligned mediolateral and one craniocaudal over the L3-L4 spinous processes.

In addition to speed and stride length, vertical accelerations were converted to reveal gravitational forces in Gs associated with heel contact, foot flat, toe-off, and midstance components of the gait cycle.[14] The reproducibility of trunk accelerations, however, was noted to decline as walking speed decreased to 0.8 m/s. The reliability of this type of accelerometry for slower walking velocities and for hemiparetic gait has not been published.[13]

For clinical trials with large numbers of subjects, then, investigators face limitations in being able to examine changes that a treatment may induce, such as improved symmetries in single- to double-limb support ratios (SLS/DLS), stride length, and swing times between a hemiparetic and less affected leg. Devices employed on short walkways also may not permit a reliable calculation of stride-to-stride fluctuations in walking parameters because too few steps are taken. The coefficient of variation of step-to-step fluctuations in speed, stride length, and swing time may reflect important aspects of motor control and may serve as sensitive outcome measures for interventional clinical trials.[15]

The Intelligent Device for Energy Expenditure and Activity (IDEEA, MiniSun, Fresno, CA) is an integrated system of lower extremity accelerometers that acquires spatiotemporal data during walking in the laboratory or community. This study proceeded to relate the patterns of accelerations and decelerations produced by the IDEEA during walking to more familiar kinematic and electromyographic data. Then, the accelerometry system (ACS) was compared to a criterion standard, the footswitch-based Clinical Stride Analyzer System (SAS) (B and L Engineering, Los Angeles, CA). Reliability and concurrent validity were assessed across a range of walking speeds and test sessions in healthy controls and for a convenience sample of patients with hemiparetic stroke. This pilot study begins the experimental process of determining whether the ACS can provide outcome measures of value for large clinical trials.

METHODS

Subjects

One healthy subject (age, 56 years) was studied with the ACS, by video-based motion analysis, and with electromyography (EMG). Twelve healthy individuals (7 males, 5 females), ranging from 18 to 69 years of age (mean, 31 ± 17 years), participated in the reliability study. Eight of these volunteers also participated in the concurrent validity study with the ACS and SAS, along with 6 hemiplegic subjects (5 males, 1 female), ages 56 to 70 years (mean, 64 ± 6 years). All subjects signed an informed consent under an institutional review board–approved protocol.

Equipment

Accelerometry system. The ACS was designed to estimate energy expenditure during daily activities outside of a laboratory.[16,17] Its development required reiterative extractions of many parameters from biaxial accelerometers placed on the lower extremities and chest during tests in a metabolic chamber to determine whether a subject walked, climbed or descended stairs, ran, jumped, sat, or crossed the legs. Highly significant correlations were found between energy expenditure determined by the ACS and measurements by calorimetry over periods as long as 24 h.[16]

The ACS employs 5 biaxial capacitive accelerometers manufactured by MiniSun with sampling rates of 32 Hz. They were taped without gel to the upper chest (4 cm below the top of the sternum), each anterior thigh (midpoint between the knee and anterior superior iliac spine), and under each medial forefoot (2 cm below the head of the 4th metatarsal). In this position, the accelerometers were sensitive to forward-back and up-down motion in the sagittal plane. Acceleration per se is related to the force and velocity of the moving segment, either the thigh or foot in this arrangement. A 32-MHz microprocessor with 200 MB of storage and data compression, weighing 60 g, was attached to the subject's belt. Wires connected the 5 sensors that weigh 2 g each to the microprocessor. The software setup required the thigh and foot sensors to be parallel to the ground from a seated position with the chest sensor at the vertical. An error message appeared if the sensors were not in line or the heel of a shoe was higher than 2.5 cm. Weight, height, and age were entered into the software. The entire setup took 5 min. Data acquired from walking tasks were transferred to a PC via a 12-bit AC/DC converter for signal analysis. The software automatically processed 2 h of acquired data in less than 30 s. The analysis program, which was compatible with Excel and ASCI files, allowed manual selection of the steps to be included in calculations, so that outlier steps such as a stumble or the 1st steps taken before a plateau in walking speed was attained could be examined separately and any portion of sequential steps chosen.

The gravitational forces experienced by each trunk, thigh, and foot accelerometer, measured in Gs, are acquired with the ACS. The 2 independent sensing axes (anterior-posterior and up-down) of each accelerometer use a proprietary algorithm that depends especially on height and the acceleration of the thigh and foot during the swing phase to calculate walking speed, cadence, and stride length, as well as single-limb (SLS) and double-limb

Table 1. Variables Automatically Calculated Using the Accelerometry System for a Normal Subject

Variables	Left Foot		Right Foot		Overall	
Number of steps	10		10		20	
Duration (s)	4.9		5.1		10.0	
Distance (m)	7.71		7.91		15.6	
	Mean	*SD*	Mean	*SD*	Mean	*SD*
Single support (ms)	393	16.1	400	13.2	397	14.7
Double support (ms)	109	16.5	96.9	9.9	103	14.7
SLS/DLS	3.6	0.417	4.13	0.576	3.85	0.524
SLS/Total DLS	1.90	NA	1.94	NA	1.92	NA
Swing duration (ms)	400	13.2	394	16.1	397	14.7
Cycle duration (s)	1.00	0.02	0.99	0.02	0.99	0.02
Pulling accel foot (G)	0.72	0.07	0.66	0.07	0.69	0.07
Speed (m/min)	94.3	3.3	93.8	2.3	94.0	2.9
Cadence (steps/min)	122	3.8	118	2.98	120	3.88
Step length (m)	0.77	0.04	0.79	0.03	0.78	0.03
Stride length (m)	1.56	0.04	1.57	0.05	1.56	0.05

SLS = single-limb support; DLS = double-limb support.

(DLS) support, stance, and swing times (Table 1). The ACS also records the duration and frequency of walking-related activities (such as distance traveled and range of velocities) over the course of a day. An animation of the type of leg movement activity carried out by a subject accompanies the accelerometry graphics to help distinguish walking from foot shuffling or stair climbing.

Footswitch system. The SAS is a microprocessor/PC system designed to record foot-floor contact data from footswitches to a recorder, then to the PC. Contact-closing footswitches are placed on the sole at the heel, 1st and 5th metatarsals, and great toe of each foot. A footswitch registers when a change to opened or closed lasts 10 ms or longer. A handheld start and stop switch marks the beginning and end of the test. The SAS had high test-retest reliability in healthy and hemiplegic subjects.[18]

Comparison systems. A triaxial piezoresistive accelerometer (S1586, Entran Devices, Hampton, VA) with an AD converter and 100-Hz sampling rate was employed for a comparison with the ACS's biaxial thigh sensor. A video gait analysis system employed reflective infrared markers attached to each lower extremity segment and 6 cameras (Motion Analysis, Santa Rosa, CA). A biomechanical model was used to compute lower extremity joint kinematics over a short walkway. Surface EMG was acquired with the DataLOG P3X8 (Biometrics, Gwent, UK) system using 8 EMG sensors (preamplifier SX230). The electrodes were secured to the skin over the midpoint of 4 lower limb muscles: tibialis anterior, medial gastrocnemius, biceps femoris, and vastus medialis. Integrated foot switches identified key phases of the step cycle. The raw EMG data were processed using the RMS filter and analyzed using DataLog version 3.0 to allow for rectification and linear envelopes.

Walking Tests

To best appreciate the novel accelerometry information during walking, ACS acceleration and deceleration wave forms were compared with simultaneously recorded kinematic and EMG activity in the healthy subject. Data from representative step cycles were superimposed for purposes of illustration and visual correlation. A triaxial accelerometer was placed over the thigh adjacent to the biaxial ACS thigh sensor to compare waveforms during gait. In addition, the ACS was worn for 10 h to check for signal drift and durability during community activities in 2 healthy subjects.

For the *validity study*, 8 healthy controls and 6 subjects with stroke ambulated over a tiled floor for 15 m, which was the recommended distance for use of the SAS. The healthy subjects were instructed to walk at their "usual comfortable" speed and then "somewhat slower." The subjects with stroke were instructed to walk at their "usual comfortable" walking speed and then "faster, but comfortably safe" walking speed while connected simultaneously to the ACS and SAS. This was repeated 3 times in the same session. The ACS and SAS data were automatically calculated for the 15-m walks using their software. The ACS data for 6 step cycles of every walk were recalculated manually using one of its software programs to define each toe-off and heel-strike event for comparison with the automated calculations.

For the *reliability study*, 12 healthy subjects walked at 5 different speeds 2 to 3 times in 1 test session and at a

2nd test session on a different day. The ACS sensor placement and programming setup were repeated for the test and the retest sessions. Subjects were instructed to walk at their usual speed, then faster, fastest, usual, slower, and slowest speeds, each time over a tiled floor for 30 m. The middle 20 strides were analyzed because walking speed reached a plateau for all subjects.

Statistical Analyses

For the validity study, values for SLS, DLS, swing and stance times, cadence, duration of a gait cycle, and walking speed from the ACS and SAS were determined by calculating the mean and standard deviation for each series of strides and for each leg, comparing left to left and right to right or hemiparetic legs to each other and unaffected legs to each other. The 2-tailed Student *t* test was used for each gait parameter. In addition, the coefficient of variation for each measure was calculated to determine the amount of recorded step-to-step variation that occurred during each walking test. The coefficient of variation represents the ratio of the standard deviation to the mean, which is an estimate of uncertainty. This statistical approach allowed a test of the reliability of each device and their relative reliabilities within the context of the variations in gait patterns that normally occur during walking.[15] A 2-tailed *t* test assessed the significance of differences.

For the reliability study, ACS data were analyzed in the healthy subjects using a 2-tailed *t* test derived from a mixed model to compare subjects walking at different speeds and on different days. The resulting *P* values examined the significance of the differences in the mean of each dependent variable (SLS, DLS, SLS/DLS, swing and stance times, gait cycle, cadence, and stride length) for each speed trial, side (left or right leg), and time of testing. For this 2-way nested ANOVA, speed was the 1st factor tested, then side nested in each speed, trial nested within each side, and session time nested within each trial. In addition, Pearson correlation coefficient was calculated for walking speed and acceleration of the foot from toe-off to heel contact (called pulling acceleration by the developer) across the range of speeds for each subject to confirm whether this acceleration was a critical variable in the ACS calculation of walking speed.

RESULTS

Accelerometry, Kinematics, and Electromyography

Table 1 shows typical spatiotemporal data calculated by ACS software for 10 consecutive step cycles in one of the healthy subjects. The processed recordings of foot accelerations from which these were derived are shown in Figure 1 (top); the *y*-axis is converted to Gs by G = n/15, as determined by the manufacturer. For the healthy subject (top), the accelerations and decelerations for the right and left foots alternated rather symmetrically with similar peaks. For a subject with hemiparesis (bottom), the peak accelerations of the affected left leg were lower and the accelerations-decelerations during the swing phase were more irregular

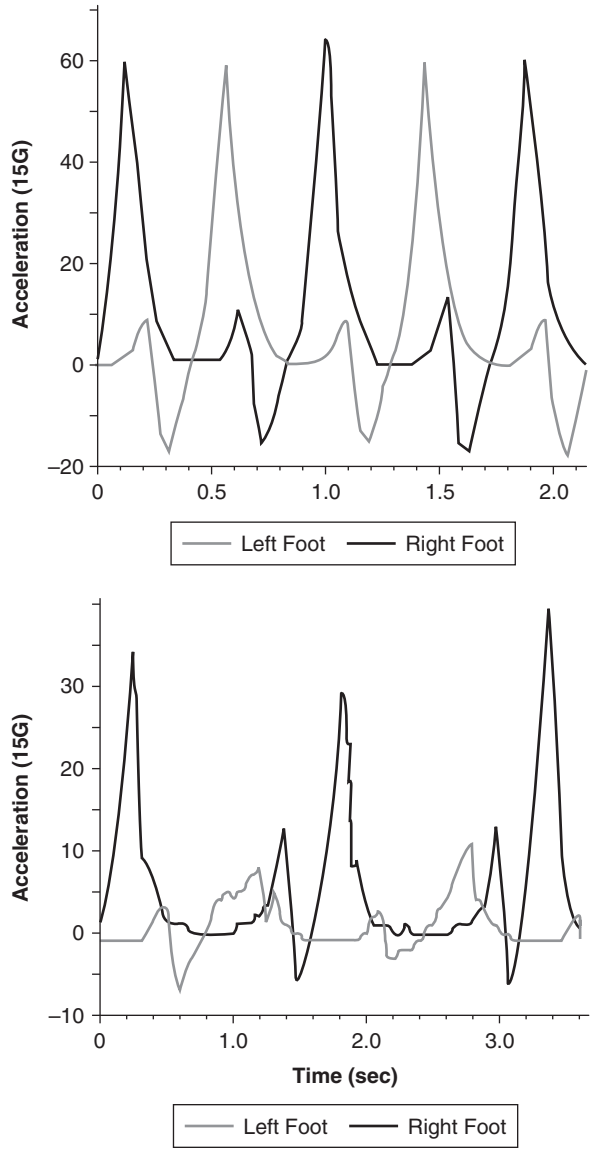

Figure 1. Processed data from right and left foot accelerometry system accelerometers are shown for 3 steps taken by a healthy subject (top) and a subject with a left hemiparetic stroke (bottom). The asymmetry of the left and right lower extremity wave forms is apparent for the patient with stroke, along with lower amplitude irregular accelerations for each leg and slower walking speed compared to the healthy subject.

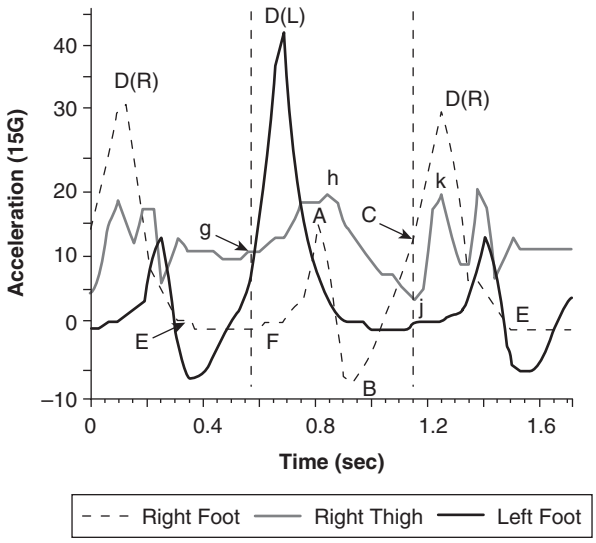

Figure 2. Typical accelerometry system data for 1 step by a normal subject shows the timing of gait-related events for the right foot and thigh. The left foot acceleration curve is shown for reference. Acceleration in gravitational force (Gs) on the *y*-axis is derived from G = n/15. The capitalized letters mark detectable events from the foot sensor, and the small letters refer to the thigh sensor, per the key. Right foot sensor: F-A = from foot flat to heel-off, the foot accelerates into planter flexion; A = toe-off (start of swing); A-B = initial vertical movement of foot; B = initial forward acceleration of foot for swing (start of pulling acceleration of the foot); B-C = mid-swing (ongoing foot acceleration); C-D(R) = terminal swing; DR = initial heel contact at end of swing; D-E = ankle dorsiflexion to foot flat; E-F = mid to terminal stance (no foot acceleration); D(L) = initial heel contact (left foot). Right thigh sensor: g-h = initial thigh acceleration with heel-off; h = active hip flexion with toe-off (pulling acceleration of the thigh); h-j = swing (thigh decelerates in relation to foot); j = end of thigh flexion; k = forward and back accelerations with initial loading of thigh at and after heel contact.

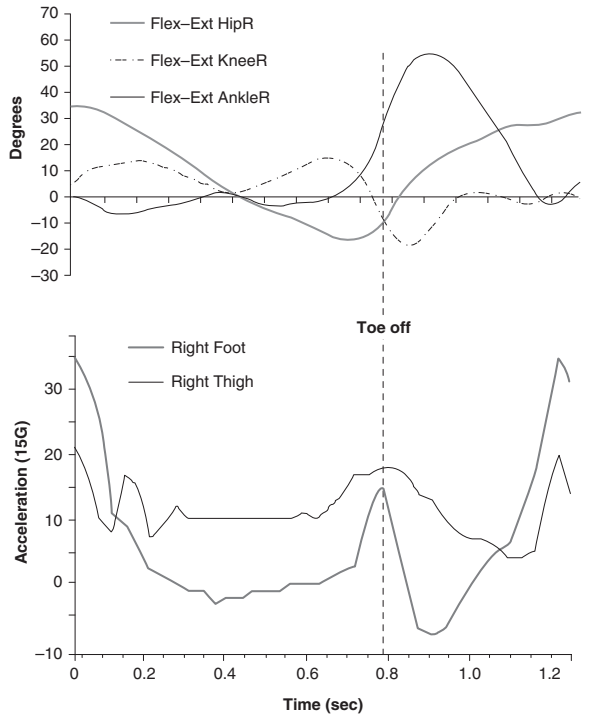

Figure 3. The upper graph shows the kinematics of the right hip (Flex-Ext HipR), knee, and ankle of one leg from heel contact to heel contact recorded during video gait analysis with infrared markers. The *y*-axis is in degrees of flexion or extension. The lower graph is a simultaneous accelerometry system recording of the accelerations of the foot and thigh with the *y*-axis in gravitational force, Gs. The vertical dotted line marks toe-off at the start of the swing phase.

than those of the right leg. Figure 2 defines the gait cycle components of accelerations and decelerations from each foot and the right thigh in a healthy subject for 1 step, from initial heel contact to heel contact of the right leg. Figure 3 relates kinematics (top) to accelerometry (bottom) over time (magnified along the *x*-axis) for 1 step cycle in a healthy subject. Typical phases of the step cycle, including heel contact, foot-flat and mid-stance, heel-off, toe-off, mid-swing, and end-of-swing heel contact, were definable by the ACS. Figure 4 illustrates the EMG bursts (linear envelopes, bottom) from 4 right leg muscles over 3 steps timed to the ACS recording from the right foot sensor in a healthy subject.

A visual comparison (not shown) of the waveforms produced from the thigh by the Entran triaxial accelerometer and the ACS biaxial sensor revealed almost identical waveforms and G forces at toe-off and throughout the swing phase, at heel contact, and during mid-stance in the forward-back sagittal plane for each.

Reliability and Validity

To examine reliability of spatiotemporal parameters, 12 healthy subjects walked at 5 different speeds (ranging from slowest to usual to fastest, from 0.5 to 1.4 m/s) repeated in several trials on 2 test session days, leading to 350 observations used for the nested ANOVA analyses (Table 2). The ACS data revealed *P* values for each dependent variable to be significant for speed ($P < 0.0001$), as expected. No significant effect for left and right leg, trial, or session was found ($P > 0.10$), revealing that ACS data were reproducible for each subject across a wide range of speeds and had high interday reliability. Walking speeds for each subject correlated with the pulling acceleration of the foot during swing (Pearson $r > 0.9$).

Walking speeds ranged from 0.65 to 1.8 m/s for healthy subjects when instructed to walk at casual and

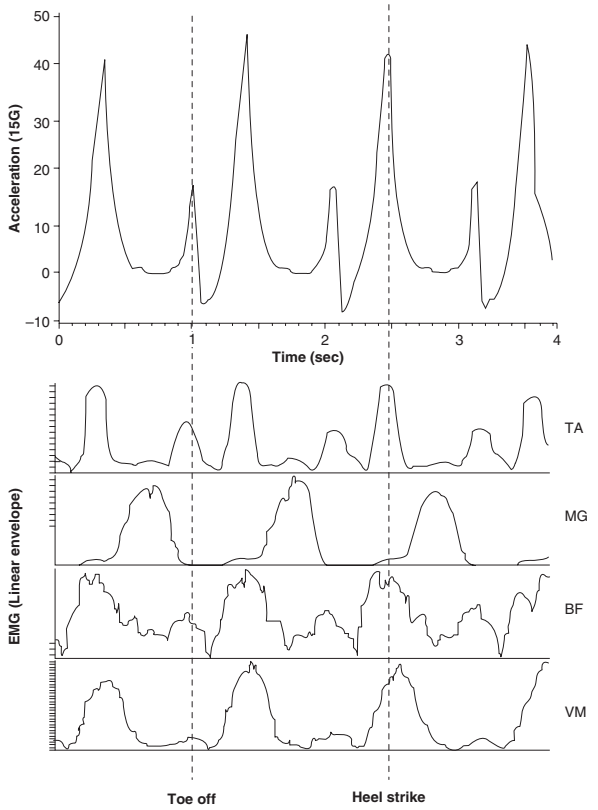

Figure 4. Simultaneous recordings of 4 steps by one leg from the accelerometry system foot accelerometer (top) and surface electromyography (EMG) (bottom) from the tibialis anterior (TA), medial gastrocnemius (MG), biceps femoris (BF), and vastus medialis (VM) muscles. The timing and peak amplitude (mv) of the linear envelopes of the EMG bursts from each muscle coincide with the expected phases of the gait cycle depicted in the timing of the acceleration-deceleration of the foot with each step. The amplitudes of the EMG bursts on the *y*-axis are not shown on the same scale. The 1st vertical dashed line denotes one toe-off event, typically revealed by the smaller of the two acceleration peaks, and coincides with the end of MG activation between heel-off and the start of swing. The 2nd dashed line shows heel contact at the end of the swing phase, exemplified by the higher acceleration peak, which coincides with the higher TA burst.

slower velocities and 0.45 to 0.95 m/s for hemiparetic subjects who were asked to walk at their usual and "safe fastest" velocities. Simultaneous data recorded from the ACS and SAS during a 15-m walk revealed no significant differences in the means calculated by each device for the same leg of each subject ($P > 0.5$ for each parameter). Across healthy subjects, modest differences in stance and swing times, as expected, were found between the left and right legs within subjects for respective phases of the gait cycle. For those with stroke, SLS was significantly less for the affected than unaffected leg by both devices, as anticipated.

The differences in the mean differences for each spatiotemporal variable for each leg, or the coefficients of variation, were not significant ($P > 0.2$; variation <3%) for the respective legs of the healthy subjects (Table 3) or for each hemiparetic leg and unaffected leg of the stroke subjects (Table 4), except for the DLS measure obtained from the SAS.

DISCUSSION

In this pilot study of healthy and hemiparetic subjects, no statistically significant differences for spatiotemporal measures of gait were found between the ACS and SAS. In addition to concurrent validity, the ACS had high test-retest reliability and no significant variations within measures across test sessions and walking speeds for SLS, DLS, stance and swing times, stride length, cadence, and walking speed. Thus, the ACS appeared to offer a valid and reliable measure of the spatiotemporal features of gait for each leg, at least for walking speeds from 0.5 to 1.8 m/s. Further studies will be necessary, perhaps within the context of a large clinical trial of a walking intervention, to determine if ACS measures will be responsive to changes in symmetries in the SLS/DLS ratio, stride length, and swing times between a hemiparetic and less affected leg. The nonsignificant differences in coefficients of variation for spatiotemporal variables using the ACS suggest that it may also be a valuable tool for relating stride-to-stride fluctuations during unconstrained overground walking as an outcome measure for the efficacy of gait training.[15]

The ACS curves fit the timing of expected changes in parameters that are more familiar to investigators—joint angles at the hip, knee, and ankle and the EMG bursts from key muscles in a healthy subject. The ACS thigh sensor and a different manufacturer's commercially available triaxial sensor revealed similar waveforms for specific phases of the gait cycle, confirming that the forward and vertical axes recorded by the proprietary ACS accelerometers are valid for accelerometry calculations for walking. Pulling acceleration of the foot (Figure 2 B–D) correlated linearly with the ACS measure of walking speed, which suggests that this acceleration during the swing phase is a key component in the ACS software's calculation of walking speed.

The literature supported use of the SAS for a concurrent validity study of the ACS. The SAS and the GAITRite walkway (MAP/CIR, Essex, UK), which has sensors embedded in a 3.5-m carpet, were found to have similar reliability for measures of cadence, walking speed, and SLS in a test-retest study of young and older persons.[19] In another investigation, the SAS had high retest reliability for temporal and distance data within a test session and between 2 test sessions for normal

Table 2. Reproducibility of Accelerometry Measures in 12 Healthy Subjects across Multiple Walking Speeds Assessed for Each Leg within Repeated Trials over 2 Sessions

	P Values			
Variable	Speed	Side (Speed)	Trial (Side)	Session (Trial)
SLS	<0.0001	0.96	0.52	0.44
DLS	<0.0001	0.99	0.62	0.10
SLS/DLS	<0.0001	0.95	0.78	0.45
Swing duration	<0.0001	0.97	0.51	0.44
Stance duration	<0.0001	1.0	0.91	0.45
Stride length	0.0001	1.0	1.0	0.33
Gait cycle	<0.0001	1.0	0.98	0.49
Cadence	<0.0001	0.96	0.94	0.49
Acceleration of foot at swing phase	<0.0001	0.16	0.94	0.56

SLS = single-limb support time; DLS = double-limb support (3 periods).

Table 3. Concurrent Validity Comparison of Coefficients of Variation for Accelerometry (ACS) and Footswitch (SAS) Parameters for Healthy Subjects

Leg	Variable	ACS CV	SAS CV	*P* Value for Difference in CV
Left	SLS	2.18	2.47	0.8
	DLS	6.63	39.1	0.01
	Swing duration	1.95	1.46	0.6
	Stance duration	2.85	3.01	0.9
Right	SLS	1.95	0.92	0.3
	DLS	7.88	39.7	0.01
	Swing duration	2.18	2.56	0.7
	Stance duration	2.39	2.61	0.9
Both	Cadence	2.10	2.00	0.9
	Stride	2.22	3.04	0.5
	Gait cycle	2.27	2.03	0.8

CV = coefficient of variation; ACS = accelerometry system; SAS = Stride Analyzer System; SLS = single-limb support; DLS = double-limb support.

Table 4. Criterion Validity Comparison of Coefficients of Variation for Accelerometry (ACS) and Footswitch (SAS) Data for Ambulatory Hemiparetic Subjects across Several Walking Velocities

Side	Variable	ACS CV	SAS CV	*P* Value for Difference in CV
Affected	SLS	1.9	2.3	0.5
	DLS	9.4	20.9	0.4
	Swing duration	5.9	2.4	0.9
	Stance duration	2.5	2.6	0.2
Unaffected	SLS	2.5	2.6	0.5
	DLS	8.0	16.7	0.4
	Swing duration	2.6	2.5	0.3
	Stance duration	1.9	2.5	0.7
Both	Cadence	2.2	1.9	0.7
	Stride length	3.0	3.3	0.8
	Gait cycle	2.1	2.1	0.9

CV = coefficient of variation; ACS = accelerometry system; SAS = Stride Analyzer System; SLS = single-limb support; DLS = double-limb support.

subjects and for hemiparetic subjects who could walk independently.[18] The intraclass correlation coefficient was over 0.85 for each temporal measure. Confidence intervals were large for the small number of hemiparetic subjects in this study, however, so any changes in spatiotemporal asymmetries between the affected and unaffected legs may not be reliably determined with the SAS. The investigators estimated that an initial

asymmetry of 30% may be necessary to detect a change to greater symmetry for SLS time.

Measurement errors with the SAS and ACS may be due to restrictions in gait imposed by the test apparatus, fluctuations that are inherent between gait performances, differences in the placement of sensors, fluxes within testing equipment such as signal drift, and failed sensors or switches. The SAS had a significantly greater coefficient of variation for DLS, which may have reflected a systematic variation in footswitch on and off timing that was magnified by the 3 periods of DLS for each full gait cycle. The SAS calculations rely on the time taken to walk a fixed distance. Total time is determined with a manual switch to mark the initiation and termination of the walk, so timing is subject to small errors. In addition, the SAS does not account for the distance of the step before and after the fixed walking distance (15 m in this study) in its initial and final stride length calculation. During a short-distance walk, these small errors could affect the total DLS time. The SAS may also be less reliable for serial measures in clinical trials of impaired patients who land with the foot flat because the switches are not always activated. In our study, at least 1 data collection from 2 healthy and 2 hemiparetic subjects had to be discarded because the SAS failed to signal heel contact or toe-off for at least several steps. The ACS, unlike the SAS, consistently registered the timing of toe-off and heel contact (or foot flat), regardless of how the foot landed, because vertical and forward-back sagittal plane accelerations and decelerations were always identifiable from the processed wave forms.

The ACS appeared to be durable. Sudden accelerations of the sensors on the sole incurred when running, stubbing the foot, and hopping did not degrade subsequent data collections. We did not encounter error or noise related to other electronic devices. No drift was noted in recordings for up to 10 h. The 32-Hz frequency range for acquisition of accelerations appeared to be adequate for locomotor applications at speeds as low as 0.4 m/s and as high as 1.8 m/s.

The ACS has several shortcomings. The automated software for calculating temporal data was less reliable at slow walking speeds (<0.5 m/s) for an affected leg with short stride length and, thus, little forward acceleration. This problem arose in part because the manufacturer had cut off very slow foot accelerations from being interpreted as walking, so as not to confuse foot shuffling for walking in the original metabolic studies for which the ACS was designed. The manufacturer recently altered the software so that slower step accelerations are recognized as the swing phase of walking. Our results were highly reproducible at velocities as low as 0.4 m/s, however, when we manually selected toe-off and heel-strike. This added a few minutes to the time for analysis. The calculation of walking speed, however, will always be inaccurate for subjects who stop and start before the end of a timed walk or pause between each forward leg acceleration. The device does not calculate speed by distance/time but rather primarily by the acceleration of the swing leg. Thus, full stops during a walk will not be reflected in the calculation for speed over a given distance. In this study, the ACS measure of speed was accurate as long as walking was continuous. For velocity calculations over a short distance in subjects with discontinuous walking, a clocked walking speed should supplement the otherwise accurate ACS spatiotemporal measures. Another issue is that the automated software uses definitions of SLS, as well as DLS, that differ from those ordinarily employed. For example, left-sided SLS by the ACS is calculated from right-sided toe-off to right heel contact. Step cycle duration is defined by heel contact of one leg to heel contact of the other, whereas the conventional definition, used by the SAS, is from heel contact to heel contact of the same foot. This difference alters the calculation of DLS time. The SAS calculates the SLS/DLS ratio by including the usual 3 periods of DLS. The ACS was programmed to incorporate only 2 DLS times for each leg, so that if total DLS time is needed, a manual calculation using the ACS software must be performed. The manufacturer could correct this.

Future studies with the ACS are necessary to determine reliability and validity in patients who are more disabled than those in this study. The ACS should also be evaluated in disabled subjects for its reliability in identifying the type and duration of activities, such as climbing stairs, sitting, and standing over the course of daily activities. These data could be useful for preinterim and posttest of mobility during a clinical trial without the problems associated with relying on subjects to fill out a daily diary about the time they spent walking. The ACS provides other measures that may have value for clinical trials. The thigh accelerometer records the initial passive and then voluntary forward acceleration (*h* in Figure 2) needed to initiate swing at the hip. This level of pulling acceleration could be assessed as a measure of change in voluntary motor control in hemiparetic subjects, who often have difficulty initiating hip flexion for the swing phase. Finally, knowledge of accelerations and decelerations of both thighs and feet may provide therapists with additional insights into how to train patients. Since ACS data can be processed and analyzed within several minutes of acquisition, therapists could use it to make a rapid comparison of the effects of a short-term training strategy on targeted gait deviations. This goal-oriented approach may also help standardize or monitor the gait training efforts during a clinical trial.

CONCLUSION

The bilateral lower extremity accelerometry device was found to be a convenient, reliable, and valid measure of spatiotemporal parameters of the gait cycle across the upper range of walking velocities found in people with neurological diseases and in the frail elderly. Preliminary data showed promise for characterizing clinically important parameters of the gait pattern in hemiparetic subjects. Serial measurements of walking over the time of improvement or decline in ability will be necessary to assess whether the ACS is responsive to changes in the spatiotemporal variables for each leg. If investigators deploy it beyond a laboratory walkway, they can also assess the potential of the ACS as an outcome measure with ecological validity associated with home and community levels of activity and participation.

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