J Med Syst
DOI 10.1007/s10916-009-9311-8

ORIGINAL PAPER

Portable Activity Monitoring System for Temporal Parameters of Gait Cycles

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Received: 22 February 2009 / Accepted: 30 April 2009

Abstract A portable and wireless activity monitoring system was developed for the estimation of temporal gait parameters. The new system was built using three-axis accelerometers to automatically detect walking steps with various walking speeds. The accuracy of walking step-peak detection algorithm was assessed by using a running machine with variable speeds. To assess the consistency of gait parameter analysis system, estimated parameters, such as heel-contact and toe-off time based on accelerometers and footswitches were compared for consecutive 20 steps from 19 individual healthy subjects. Accelerometers and footswitches had high consistency in the temporal gait parameters. The stance, swing, single-limb support, and double-limb support time of gait cycle revealed ICCs values of 0.95, 0.93, 0.86, and 0.75 on the right and 0.96, 0.86, 0.93, 0.84 on the left, respectively. And the walking step-peak detection accuracy was 99.15% (±0.007) for the proposed method compared to 87.48% (±0.033) for a pedometer. Therefore, the proposed activity monitoring system proved to be a reliable and useful tool for identification of temporal gait parameters and walking pattern classification.

Keywords Activity monitoring · Accelerometer · Gait analysis · Temporal parameter · Walking · Step-peak

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Introduction

Gait analysis techniques have been used in clinical research to investigate the features of normal or abnormal gait, to quantify subsequent improvement after operation, and to assess balance and mobility monitoring. Objective measures of the motion analysis are necessary for better management of rehabilitation. Measurements of temporal parameters of gait are used for the evaluation of lower-extremity disorders and for the quantification of their subsequent improvement after therapeutic exercise. Various instruments have been developed to assist in the study of human gait. Motion analysis systems and force plates contain kinematics and kinetics in three planes to spatio-temporal measures of gait analysis, but their cost and intricate setting have limited those to use them for large trials and daily life [1]. Foot switches or pressure sensors attached to the sole are used temporal parameters [2, 3]. These techniques provide unsatisfactory results for abnormal walking and sensor attachment. Inexpensive and miniature accelerometer sensors have been used in a

Published online: 16 June 2009

clinical setting [4]. Accelerometer is typically lightweight and wearable which facilitate unencumbered movement of the subject. And it can be used to detect simple temporal features of gait, such as the stride time. A uni-axial accelerometer to identify each heel strike was used by Evans et al. [5]. They demonstrated the possibility of detecting steps and stride time from trunk acceleration. Temporal parameters were measured using two accelerometers attached on thigh by Aminian et al. [6]. According to Zijlstra and Hof [7], a basic pattern of trunk acceleration with fixed relationships to spatio-temporal gait parameters could be expected during walking. Still, however, a portable and wireless acceleration data acquisition system to measure acceleration signals during walking and a gait analysis algorithm which can estimate temporal parameters has not been introduced. Also, the possibility to identify the main peaks of output signals such as heel strike and toe off from two accelerometers attached on each ankle compared with the standard reference based on footswitches has not been proposed.

In this study, we have developed a small, wireless, and portable three-axis acceleration data acquisition system to measure acceleration signal during walking and a gait analysis algorithm which can estimate temporal parameters of gait. We have also assessed its consistency by comparing its temporal parameters of gait with those of footswitches.

Methods

Subjects

The ten subjects (six men and four women, age: 25± 3 years) volunteered to participate in the validation study of gait cycles and walking step-peak detecting algorithm. Their average weight of the subjects 65 kg (range 48~85 kg), and their average height of the subjects was 166 cm (range 154~176 cm). Subjects with lower limb abnormalities, previous lower limb surgery, and low back pain as well as with neurological or surgical abnormalities on clinical examination were excluded from this study. Procedures were approved by the Human Investigation Committee of the Institution and all subjects signed informed consent form before participation.

Implementation of accelerometer system

Small and wireless system was built using three-axis accelerometer (MMA7260, Freescale, TX, USA). The sensitivity of accelerometer is calibrated by rotating device with variable speed, and measured to calibrate the accelerometer signals in gravitational constant g (=9.8 m/s^2) as a unit. With the advantages of small-package size (6×6×1.45 mm; QFN-16 package) and low power consumption, the system could be powered by a small 3.7 V Li-Polymer cell battery (300 mAh). In order to detect heel strike and toe off time of ankle, small and sensitive pressure sensors (foot switch/ IESF-R-5L, CUI Inc., OR, USA) were connected to the proposed system and located in one's shoes. So, we could observe contact time of heel and toe, also, acceleration of ankle at the same time. Overall specifications of the developed system are summarized in Table 1 and system configuration diagram and photos are shown in Figs. 1 and 2.

Data processing

Both the accelerometer data and foot switch data from the pair of ankle devices were A/D converted and transmitted wirelessly to the PC-side receiver at the rate of 60 Hz. The Acknowledge 3.7.3 program (BIOPAC, USA) was used to

Table 1 Specification of the proposed portable activity monitoring system

		Specification
Degree of freedom (DOF)		3
Accelerometer (MMA7260, Freescale)		3-axis (800 mV/g)
Foot switch		2 (heel & toe)
Pressure Sensor (IESF-R-5L, CUI)		Flexible PCB (Max. Load 4 kg)
A/D converter (embedded with MSP430F149)	Resolution	12 bits
	Sampling rate	60 Hz
Wireless Comm. (nRF2401, Nordic, Norway)	Carrier Freq.	2.4 GHz
	Modulation	GFSK
	Date Tx/Rx rate	1 Mbps (Shock-Burst)
Power	Battery-powered	Li-Ion cell (3.7 V 300 mAh)
Physical Characteristics	Size (W × H × D)	25×160×25,mm
	Weight	50 g
Current consumption		<12 mA at 3.3 V

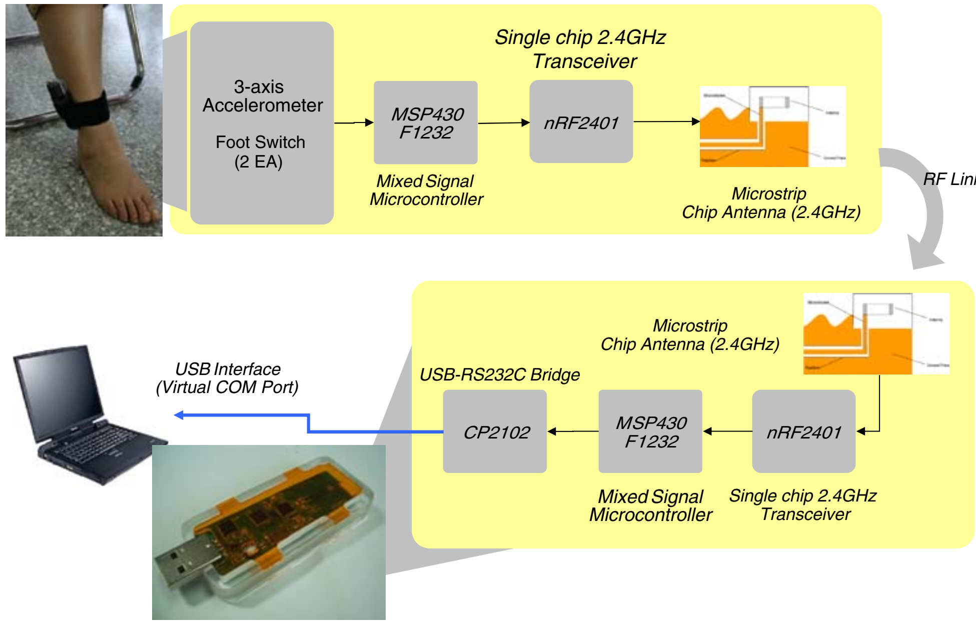

Fig. 1 System block diagram of portable activity monitoring system

store transmitted data. Then, three-dimensional acceleration data is transformed into a "signal vector magnitude (SVM) [8] which is represented as:

$$\mathrm{SVM} = \sqrt{a_x^2 + a_y^2 + a_z^2} \tag{1}$$

Then, in order to extract features points which are associated with gait cycle, LPF (f_{pass}=6 Hz, f_{stop}=10 Hz) was designed.

$$y[n] = \sum_{i=0}^{10} b_i x[n] \tag{2}$$

where $x[n]$ is the input signal after SVM calculation and $\{b_i\}$ are the coefficients of low-pass filter. Frequency response of LPF is shown in Fig. 3. After low-pass filtering, in order to extract peak candidate points, least-square polynomial derivative approximations filter is applied. The difference equation for the five-point filter is:

$$y[n] = \frac{1}{10}(2x[n] + x[n-1] - x[n-3] - 2x[n-4]) \tag{3}$$

Then, step peak detection algorithm, based on syntactic peak-valley detection algorithm was applied. From the detected step peaks of each subject, consecutive 20 steps were selected for the gait analysis. Figure 4 shows the flowchart of walking step-peak detection algorithm.

Gait analysis

The proposed activity monitoring system was attached to the lateral side of the both ankles. Two footswitch sensors were placed inside each sole of shoes, under the heel and the first metatarsal. This way, the moments of heel strike and toe-off were directly detected. The subjects walked

Fig. 2 Photos of the portable activity monitoring system and foot switches

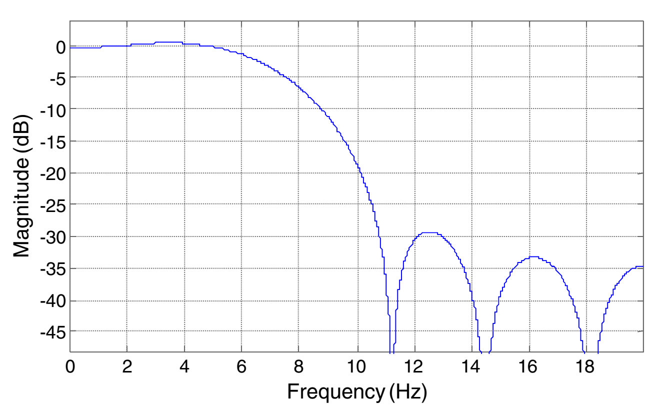

Fig. 3 Frequency response of low-pass filter

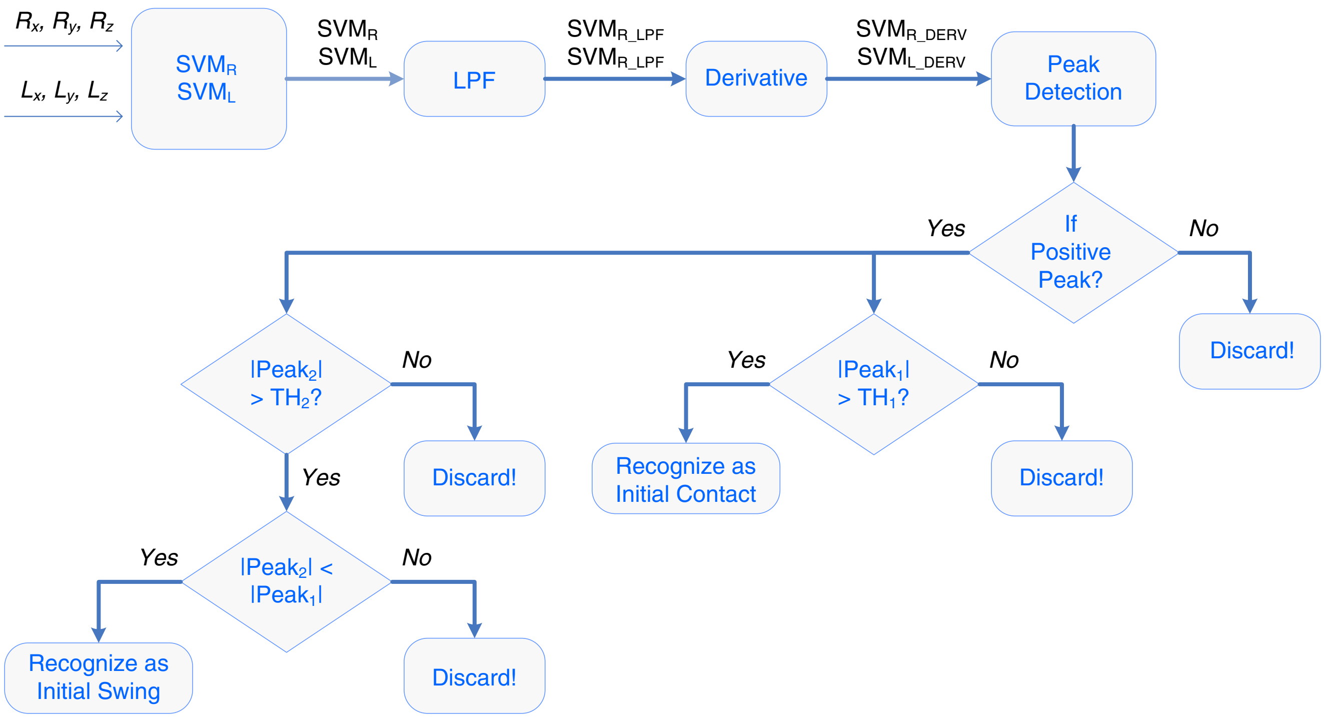

Fig. 4 The flowchart of walking step-peak detection algorithm

over a 5 m-straight corridor for 5 min. They were instructed to walk at their preferred speed while monitored wirelessly by the proposed activity monitoring system. Then from the data for the 20 consecutive steps, prominently repeating peaks of step-peak points were marked to calculate the temporal parameters of gait cycle. The Acknowledge 3.7.3 program was used for the visual inspection while comparing it with simultaneous footswitch data, and the following gait parameters were determined for the each gait cycle: stance time(ST), swing time(SW), single-limb support time (SS), and double-limb support time(DS; Fig. 5). From the 20 values for the each parameter in a single leg, statistical mean and standard deviation were calculated and used for the further statistical test.

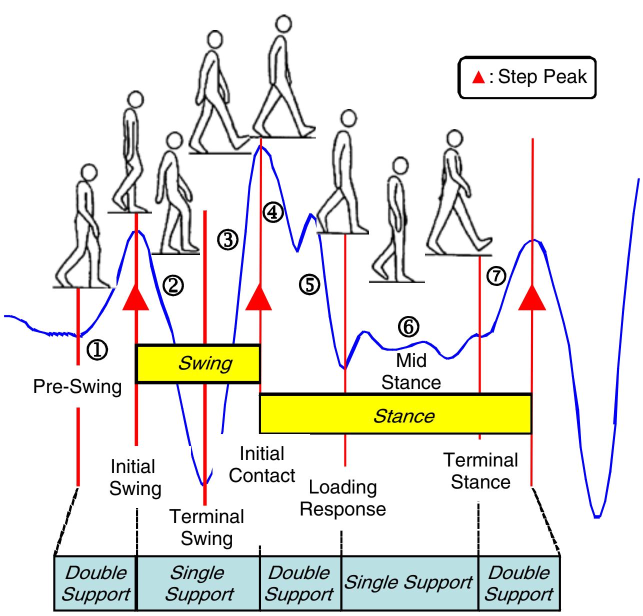

Fig. 5 The relationship between the ankle acceleration during gait cycles and the temporal gait parameters for each gait cycle: Stance time (*ST*), Swing time (*SW*), Single-limb support time (*SS*), and Double-limb support time (*DS*)

Statistical analysis

For the consistency study, the ICCs (Intraclass correlation coefficients) were used for each gait parameter. The ICCs value less than 0.75 were considered as 'poor to moderate', greater than 0.75 as 'good', and bigger than 0.90 as 'excellent' [9].

Experiments and results

The walking step-peaks which appear when left and right foots contact the ground, regarded as reference points in analyzing temporal parameters. Figure 5 shows the relationship between the ankle acceleration during gait cycles

Table 2 Experiment protocol of running machine for the validation of walking step-peak detection algorithm.

Running machine speed (km/h)	2	4	6	8	10
Duration (sec)	10	10	10	10	10

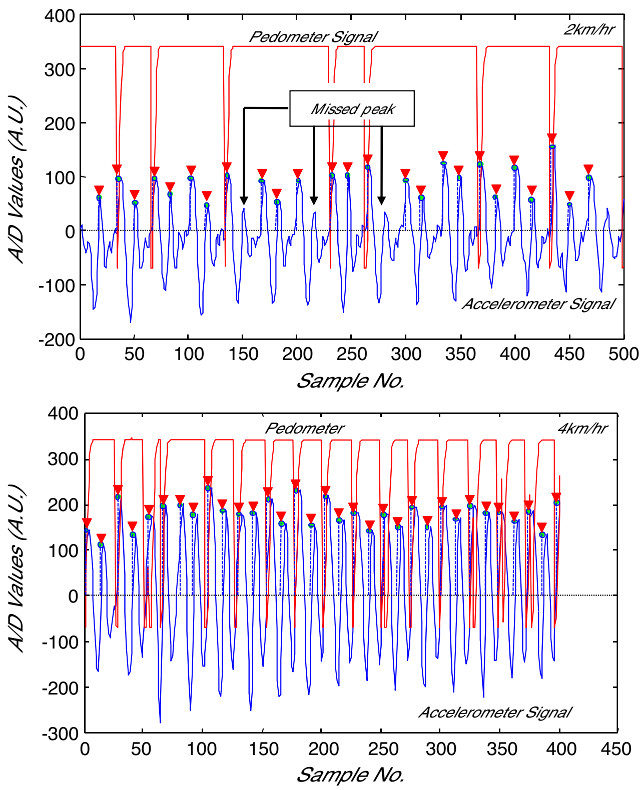

Fig. 6 An example of walking step-peak detection result at speed of 2 and 4 km/h: the *red triangle* indicates automatically detected step-peak

and the temporal gait parameters for each gait cycle. The red squares indicate step-peaks which are detected in gait cycles. As a validation study of walking step-peak detection algorithm, ten young subjects (25±3 years) were instructed to walk on running machine at various predefined revolution speeds as summarized in Table 2. In order to compare detection accuracy, pedometer (HJ-150, OMROM, JAPAN) were used to all subjects, and real walking steps are countered by hand counter manually. Figure 6 shows examples of step-peak detection result at the speed of 2 and 4 km/h. When subject walked at the speed of 2 km/h, the intensity of acceleration (normal to ground plane) is not enough to trigger wire-spring of pedometer, but sensor has enough sensitivity to trace upward and downward acceleration of the body. At the speed of 4 km/h, a wire-spring of pedometer was triggered when the foot wearing the proposed monitoring system contacts the ground, but not triggered when the opposite foot contacts the ground. Even in this case, the accelerometer was enough to sense the small change of acceleration, and the proposed algorithm could detect correctly step-peaks as well. Figure 7 shows a whole step- peak detection result from one of our subjects. Over the speed of 6 km/h, the proposed algorithm and pedometer both could detect step-peaks correctly. Most of errors for pedometer were from the low intensity of vertical

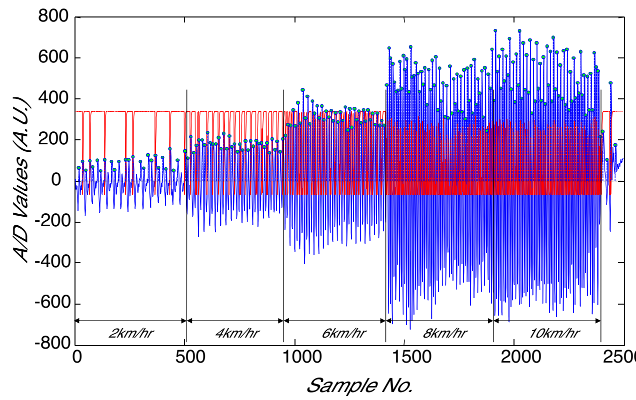

Fig. 7 The result of walking step-peak detection over various walking speeds

acceleration which might not enough to trigger wire-spring of pedometer. Overall accuracy of step-peak detection is summarized in Table 3. The accuracy of the proposed algorithm is 99.15% (±0.007) and 87.48% (±0.033) for a pedometer.

A typical example of ankle acceleration during gait cycles was plotted in Fig. 8. The results showed that the temporal parameters detected by the footswitches and those from the accelerometers matched well. The average temporal parameters of gait cycles were summarized in Table 4. The mean duration of the right and left stance, swing, single-limb support, and double-limb support phases showed the differences in the accelerometers and foot-switches for 20 steps (Table 4). Temporal gait parameters measured with our system revealed high ICCs (Table 5).

Table 3 The accuracy comparison between the proposed walking step-peak detection and pedometer from a verification experiment

Subjects	Total steps (manual counter)	Pedometer (HJ-150)	The proposed algorithm	Accuracy of the proposed algorithm	Accuracy of pedometer
1	236	214	232	98.31%	90.68%
2	248	229	245	98.79%	92.34%
3	243	216	240	98.77%	88.89%
4	235	202	235	100.00%	85.96%
5	227	195	223	98.24%	85.90%
6	227	199	227	100.00%	87.67%
7	227	193	226	99.56%	85.02%
8	189	156	189	100.00%	82.54%
9	236	199	233	98.73%	84.32%
10	222	203	220	99.10%	91.44%
			Average	99.15%	87.48%
			STD	0.007	0.033

The detection accuracy of each subject is average accuracy over various speeds, 2, 4, 6, 8 and 10 km/h

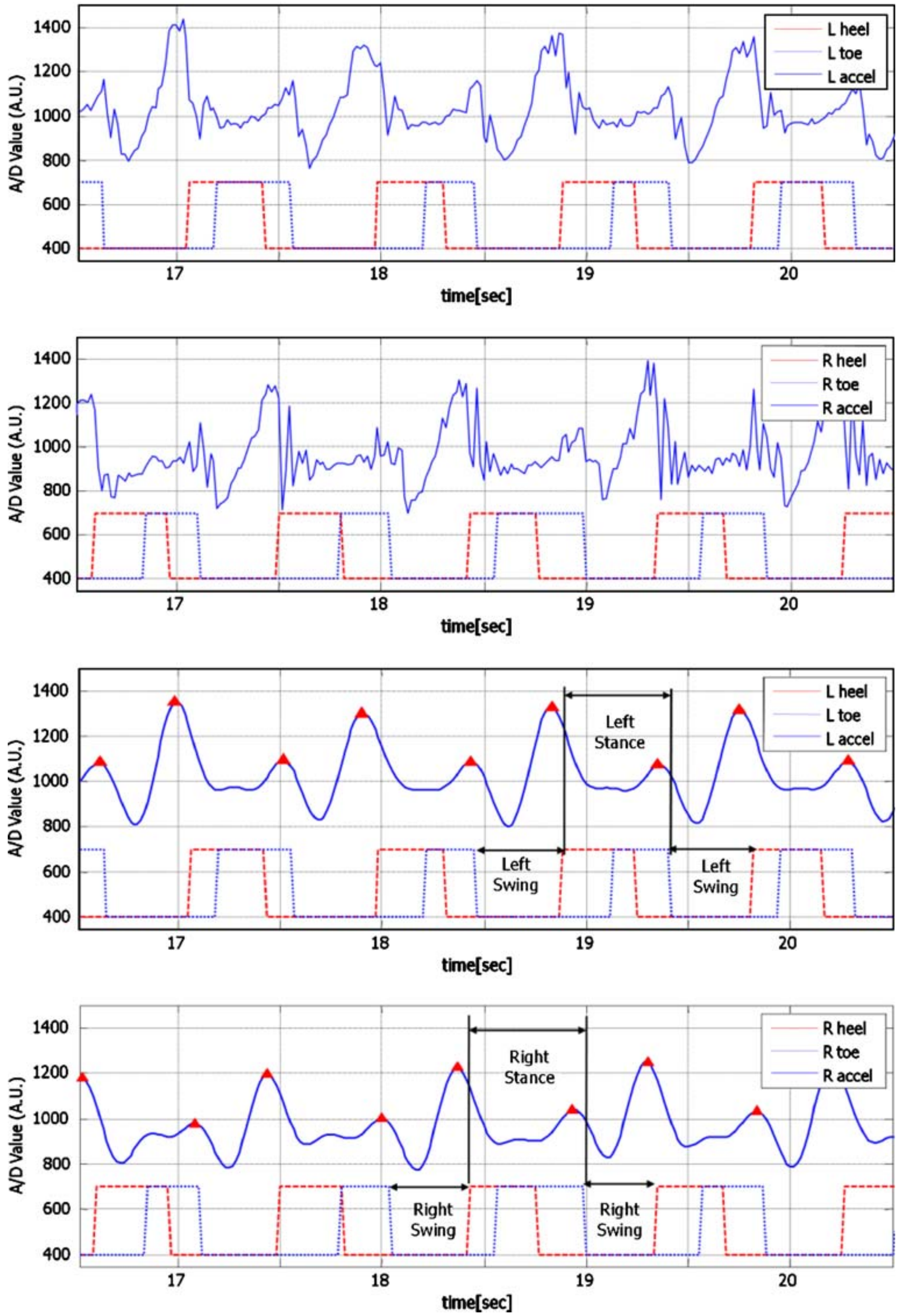

Fig. 8 The result of temporal gait parameter analysis. The vector sum of left and right foot ankle acceleration were plotted with output of footswitch over three strides for the comparison. *Upper waveforms* indicate raw signals of accelerometers and footswitches and bottom for filtered signals

The ICCs of the stance time was greater than those of the swing, single support, and double support time.

Discussion

The purpose of this study was to develop and to verify a novel, portable and wireless gait cycle analysis system. To verify the proposed system, footswitches were selected as standard criteria because footswitches have been generally used to measure the beginning and end of walking [10, 11]. The result of the study showed that two peaks of acceleration for each gait cycle from the accelerometers could be found in normal gait. The first accelerometer peak meaning heel contact and the second peak meaning toe off matched well with footswitch actions, thus showing that

Table 4 Gait temporal parameters calculated using the accelerometers placed on each ankle accelerometer and footswitch placed under each sole of foot

Ankle	Footswitches (s)				Accelerometers (s)			
	ST Mean (SD)	SW Mean (SD)	SS Mean (SD)	DS Mean (SD)	ST Mean (SD)	SW Mean (SD)	SS Mean (SD)	DS Mean (SD)
Right	0.672 (0.028)	0.449 (0.022)	0.427 (0.024)	0.245 (0.025)	0.676 (0.038)	0.438 (0.028)	0.399 (0.027)	0.277 (0.034)
Left	0.694 (0.027)	0.427 (0.024)	0.449 (0.022)	0.245 (0.025)	0.715 (0.037)	0.399 (0.027)	0.438 (0.028)	0.278 (0.033)

ST stance time, *SW* swing time, *SS* single limb support time, *DS* double-limb support time

gait phases were related to main peaks of acceleration. This result supported the study by Saremi et al. [1], which showed gait phases were related to main peaks of acceleration. In addition, the temporal parameters of gait from footswitches and those from the accelerometers were consistent enough that our system could be used as a valuable tool for gait analysis (Table 5).

This paper has focused on the pattern of accelerometer signal vector sum (SVM). The signals from accelerometers can provide useful information for gait analysis not in each of vertical, anterior-posterior, and medio-lateral axis but in the sum of three vectors. Acceleration is significantly influenced by the direction. But it may be not in one direction but in the sum of three directions. Walking is a natural locomotion and can be influenced by three directions.

In addition, the signals from the accelerometers on the ankle showed a higher consistency. This may be due to the activity of both ankles during forward progression. Walking is a distal lower-extremity performance which is primarily controlled by the ankle and foot.

ICCs were used to analyze the consistency in the temporal parameters of gait from footswitches and that of the accelerometers. The higher ICCs for temporal measures suggested that it could become a valuable tool for measuring phases of a gait cycle. Like other reliability coefficients, the ICC ranges from 0.00 to 1.00 [12]. It is calculated using variance estimates obtained through an analysis of variance. Therefore, it reflects both degree of correspondence and agreement among ratings.

The acceleration signal shape from our study was different from other studies, suggesting that the accelerometer position could affect the signal shape. The influence difference from the different accelerometer position on the body could be considered in the further studies.

The limitations of this study were small number of steps; only 20 steps from each subject were selected by automatic classification algorithm. Therefore, for generalization of our findings, further study is warranted with large number of subjects and automatic signal analysis program. Despite this limitation, our portable and wearable activity monitoring system was proved to be a valuable tool to calculate temporal parameters of gait. The 3D acceleration measurement system is inexpensive and comfortable to wear, and it can partially replace or complement the complex and expensive motion analysis.

Conclusion

In this study, portable and wireless activity monitoring system based on three-axis accelerometer was developed and its validity for detection of walking step-peak and calculating temporal gait parameters was confirmed. The main peaks of the acceleration data for heel strike and toe off could be accurately detected. This system was found to be a highly consistent tool to measure gait detection in healthy subjects in daily life for functional gait analysis. The system demonstrated the feasibility for being used for mass trials and functional gait monitoring.

Acknowledgments This work was supported by the Korea Science and Engineering Foundation grant funded by the Korea government (MEST; No. R01-2008-000-21000-0).

Table 5 Consistency comparison of ICC of variation for accelerometers and footswitches parameters for healthy subjects

Ankle	Variable	ICC(3, k)	95% Confidence interval	*P*
Right	ST	0.950	0.798~0.992	0.000
	SW	0.932	0.661~0.986	0.001
	SS	0.861	0.307~0.972	0.009
	DS	0.746	-0.267~0.949	0.045
Left	ST	0.956	0.780~0.991	0.000
	SW	0.861	0.307~0.972	0.009
	SS	0.932	0.661~0.986	0.001
	DS	0.837	0.186~0.967	0.014

ICCs intraclass correlation coefficients, *ST* stance time, *SW* swing time, *SS* single limb support time, *DS* double-limb support time

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