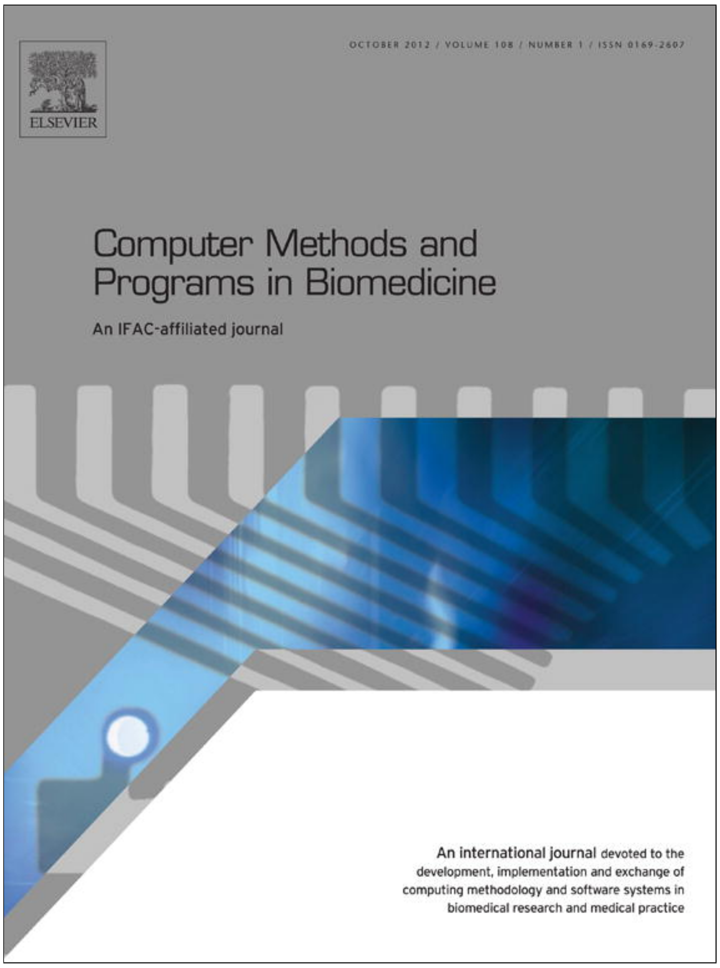
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Estimation of spatial-temporal gait parameters in level walking based on a single accelerometer: Validation on normal subjects by standard gait analysis

F. Bugané [a,b], M.G. Benedetti [a], G. Casadio [b], S. Attala [a], F. Biagi [a], M. Manca [c], A. Leardini [a,]*

[a] *Movement Analysis Laboratory, Istituto Ortopedico Rizzoli, Italy*
[b] *LorAn Engineering Srl, Castel Maggiore, Bologna, Italy*
[c] *Laboratorio Analisi del Movimento, Ospedale San Giorgio, Ferrara, Italy*

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ABSTRACT

This paper investigates the ability of a single wireless inertial sensing device stuck on the lower trunk to provide spatial-temporal parameters during level walking. The 3-axial acceleration signals were filtered and the timing of the main gait events identified. Twenty-two healthy subjects were analyzed with this system for validation, and the estimated parameters were compared with those obtained with state-of-the-art gait analysis, i.e. stereophotogrammetry and dynamometry. For each side, from four to six gait cycles were measured with the device, of which two were validated by gait analysis. The new acquisition system is easy to use and does not interfere with regular walking. No statistically significant differences were found between the acceleration-based measurements and the corresponding ones from gait analysis for most of the spatial-temporal parameters, i.e. stride length, stride duration, cadence and speed, etc.; significant differences were found for the gait cycle phases, i.e. single and double support duration, etc. The system therefore shows promise also for a future routine clinical use.

1. Introduction

Human mobility is a fundamental requirement for a satisfactory quality of life. The World Health Organization in the recent ICF – International Classification of Functioning Disability and Health – gave prominence to the analysis of functional motor aspects such as activity level and participation. It follows the importance of monitoring quantity and quality of motor activities in rehabilitation, to define therapeutic intervention setting and outcome evaluation.

Level walking is a basic requirement for many daily activities, therefore modern gait analysis provides essential information on the functional capabilities of subjects [1,2]. This is obtained by measuring the kinematics and kinetics of the main body segments and joints using stereophotogrammetry and dynamometry in well-instrumented and specifically designed laboratories. Among the many relevant measurements, spatial-temporal parameters are widely used in the clinical context. These describe quantitatively the main events of gait, and therefore reflect the ability of the patient to fulfill the general requirements of gait, i.e. the

* *Corresponding author at*: Movement Analysis Laboratory, Istituto Ortopedico Rizzoli, Via di Barbiano 1/10, 40136 – Bologna, Italy. Tel.: +39 51 6366522; fax: +39 51 6366561.
E-mail address: leardini@ior.it (A. Leardini).
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weight acceptance, the single limb support and the swing limb advancement [3]. An asymmetric gait, a prolonged stance or double stance phases, the lack of the physiological sequence of foot rockers and the reduction of speed of progression are all relevant parameters to diagnose pathological gait and to assess functional outcome after treatments. There are a number of reliable methods for measuring these gait parameters, such as force plates, plantar pressure systems and optoelectronic stereophotogrammetry [4].

According to the modern concept of ecological validity [7], wireless inertial sensing devices are being developed recently also for the assessment of spatial-temporal parameters in unobstructed environment outdoors, thus overcoming the typical limitations of measurements in indoor laboratory settings. Several applications in rehabilitation and in recovery of patient mobility have been reported already using these devices [4–6].

2. Background

Different techniques have been proposed to detect cadence and walking variability by means of accelerometers, using autocorrelations [8,9] or peak detection algorithms [10]. Validation studies have shown that successive foot contacts during gait can be detected by accelerometers attached to the foot [11,12], shank [13,14], thigh [15], and thorax [16]. Other studies [17–25] have shown that during walking a consistent pattern of trunk acceleration occurs with fixed relationships to spatial-temporal parameters. In particular, it has been shown [17,18] that three-dimensional (3D) displacements of the lower trunk during walking are predicted well by an inverted pendulum model of the body's centre of mass trajectory. In agreement with the model predictions, the amplitude and timing of the displacement were correlated to spatial-temporal parameters. In a first study, Zijlstra and Hof [17] examined the feasibility of spatial-temporal parameters estimation based on one 3-axis accelerometer positioned at L5 by a waist belt. This estimation was compared with corresponding events measured exactly by force platforms, both on a treadmill and during over-ground walking. A subsequent study by the same author [18] described these parameters during over-ground walking obtained by the same technique from a large population of healthy young and healthy elderly subjects. The validation study showed that the anterior–posterior component of the acceleration has a maximum value precisely at the foot contact instant, although the author did not investigate thoroughly the significance of the relevant double bump and minimum values. Menz et al. [19] reported a very similar and consistent pattern for this component, supporting the occurrence of heel strike at its positive peak. Mansfield and Lyons [20] analyzed the fundamental component of the anterior–posterior acceleration of the lumbar spine to identify the instants of foot contacts. In a population of active healthy adults and the elderly Auvinet et al. [21] analyzed the vertical component of the acceleration of a sensor apparently in the same lumbar spine position to seek for possible specific parameters influenced by gender and age. Gonzalez et al. [22] analyzed vertical and anterior–posterior components and identified the main gait events, which were also supported by a combined measure of ground reaction force. Although the latter two studies claimed a smooth and clear pattern also for the vertical component, its complexity makes it very difficult to implement an algorithm that can automatically identify gait cycle events especially in the case of pathological subjects. Medial-lateral acceleration has rarely been analyzed [17–19], with the purpose of distinguishing between right and left cycles.

3. Design considerations

There is therefore ample literature on the identification by lower trunk acceleration of the main general events of level walking and relevant typical gait spatial-temporal parameters, such as walking speed, stride frequency and stride length. However, the duration of the separate phases within the gait cycle, such as stance, swing, double and single supports, has rarely been addressed by this technique [21]. Clinical assessment of pathological gait would benefit greatly from this information obtained by using a single sensor, particularly in the context of rehabilitation to assess quantitatively the outcome of specific interventions on gait with the minimum encumbrance for the patient and the minimum cost for the hospital. The present study was aimed at investigating whether a single sensor on the lower trunk can provide reliably spatial-temporal parameters in level walking. Relevant reliability was assessed initially only in a population of healthy subjects, as a necessary first step before analyses and validation in a series of pathological populations.

The specific scope of the present study was to assess the relevant performance of a novel technique based on a wireless inertial sensing device, Free4Act (F4A – LorAn Engineering, Bologna, Italy), in particular to validate the estimation of the spatial-temporal parameters. The measurements obtained with this device and relevant original algorithms were compared with corresponding ones obtained by state-of-the-art gait analysis (GA). In particular, the instants of foot contacts were assessed to distinguish them between right and left leg cycles and also to estimate the specific gait cycle phases.

4. Description of method/system

The system and the method used to perform the measurements and validate the results are as follows:

4.1. *The inertial sensing and the gait analysis systems*

The new portable F4A consists of a wireless network of inertial sensors for human movement analysis. The sensors are controlled by a data logger unit (up to 16 elements) by a ZigBee radio type communication. Each sensor is sized 62 mm × 36 mm × 16 mm, has a weight of 60 g, and is composed of a 3-axis accelerometer (max range ± 6 g), a 3-axis gyroscope (full scale ± 300°/s) and a 3-axis magnetometer (full scale ± 6 gauss). However, in the present study, only the accelerometer is used. This sensing device is calibrated with the gravitational acceleration immediately after manufacturing. The F4A sensor is attached to the subject's waist with a semi-elastic belt, covering the L4–L5 inter-vertebral space, in a way acceleration is collected about the three orthogonal anatomical axes, i.e. the anterior–posterior, medio-lateral and vertical axes. From the collected acceleration signals, the following typical spatial-temporal gait parameters [26] are then obtained:

- Step length [m], the distance between the ipsilateral and contralateral heel strikes;
- Stride length [m], the distance between two consecutive heel strikes of the same foot;
- Stride length/height [%], the stride length normalized by subject height;
- Stride duration [s], the time between two consecutive heel strikes of the same foot;
- Step duration [s], the time between ipsilateral and contralateral heel strikes;
- Foot symmetry [%], the step duration as percentage of gait cycle;
- Stance duration [%], the foot support phase, i.e. from heel strike to toe off of the same foot, duration as percentage of gait cycle;
- Swing duration [%], the foot swing phase, i.e. from toe-off to heel strike of the same foot, duration as percentage of gait cycle;
- Double support duration [%], the duration of the phase of support on both feet as percentage of gait cycle;
- Single support duration [%], the duration of the phase of support on one foot as percentage of gait cycle;
- Speed [cm/s], the average instantaneous speed within the gait cycle as integration of acceleration;
- Cadence [strides/min], the number of strides in a minute;
- Normalized speed [%], the speed as percentage of the subject's height.

The overall F4A inertial sensing system, made of all relevant instrumentation including the sensors, plus all relevant algorithms and software, is able also to recognize automatically whether the starting step is of the right or left leg.

The validation of this system was performed by using a GA system, which included a stereophotogrammetric unit with eight M2-cameras (Vicon 612, Vicon Motion Capture, Oxford, UK) and two dynamometric platforms (Kistler Instruments, Einterthur, Switzerland), sampling respectively at 100 and 1000 Hz. For marker positioning and lower limb 3-D kinematics, a recently established protocol for clinical analysis was used [26,27].

4.2. Experimental setting

Each subject was fitted with the relevant reflective markers and the F4A sensor. The latter was placed over the L4–L5 inter-vertebral space so that its reference coordinate frame had the z-axis oriented to the front, x-axis oriented vertically upward and y-axis orthogonal to the other two, towards the right. This motion analysis was performed with a sensitivity for the F4A accelerometer of 3G and a sampling frequency of 50 Hz. The subjects were asked to stand up and remain in the up-right posture for a few seconds, and then to walk barefoot along a 10-m pathway, at a self-selected speed. These conditions were recommended for standard GA; however, extensive tests on different conditions (different shoes, pathway lengths, speeds of progression, etc.), had revealed that very similar patterns for the acceleration signal are obtained. This entailed 8–12 steps, according to the subject's natural cadence; the central three, for the right and left full gait cycles, were analyzed also by the GA system. This exercise was repeated 5 times, i.e. trials, for each participant. To test the right and left step identification procedure, the subjects were asked to start walking always with the right foot.

In a few subjects, at the beginning of the study, two additional F4A sensors were placed on the dorsal aspect of the forefeet.

4.3. Data processing

To estimate the spatial-temporal parameters, the anterior–posterior (z-axis) accelerometer signal of F4A was found to be the most revealing. Relevant raw data were first filtered by a first-order Butterworth low-pass filter with a cut off frequency of 2 Hz. From the typical curve (Fig. 1) with two positive and one negative peaks, the relevant timing was calculated; the second positive peak was taken as the instant of foot contact [17,18]. Therefore, step and stride were identified respectively between two and three of these events. The first and last steps were removed from any following calculation to avoid transitional phases, i.e. gait initiation and termination. Velocity and displacement were then calculated with single and double integration of the acceleration. To avoid drift, integration was performed at each step interval by assuming that speed at foot contact was zero. All spatial-temporal parameters derived immediately from these quantities.

To discriminate automatically between left and right steps, the medial-lateral (y-axis) acceleration was analyzed. Following the concepts implied in the established inverted pendulum model for level walking [28], and assuming the F4A sensor is very close to the centre of mass, acceleration to the left was taken as that during the right leg support phase and vice versa.

In the present study, as in standard GA, the first event of heel strike for each foot was taken from the relevant force platform, the previous and the following corresponding events from the comparison of that kinematics configuration (foot position, hip/knee/ankle flexion angles, etc.) over the GA results along the entire data collected. From these three events, two full gait cycles for each side were analyzed. All spatial-temporal parameters were calculated for the relevant right and left cycles.

4.4. Statistical analysis

To assess the performance of this F4A system, the estimated spatial-temporal parameters were compared with those obtained from standard GA using the paired *t*-test. When the assumptions related to the Gaussian distribution of the data and to the correlation between the compared measures were not verified by the One-Sample Kolmogorov–Smirnov test, the Wilcoxon non-parametric test was used. For both the paired t-test and the Wilcoxon test, significance was assumed for p-value smaller than 0.05. For each of the five repetitions,

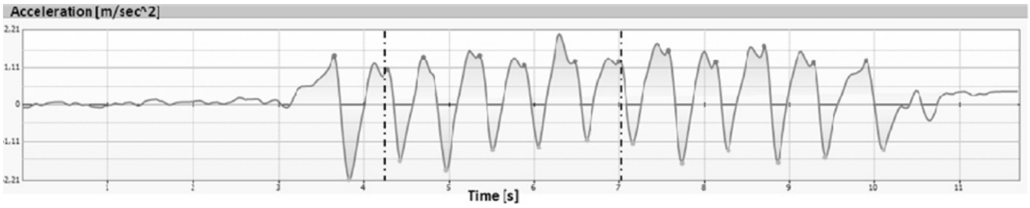

Fig. 1 – Anterior–posterior acceleration signal from the F4A sensor in a representative subject (#2) during the full exercise of standing and walking, after Butterworth low-pass filtering. The period when the GA system collects data is shown between vertical lines (dash-dot).

this comparison was based on the specific two cycles analyzed by GA, and also on all the cycles measured by the F4A.

5. Status report

Ten women and twelve men volunteered for the validation experiments. The subjects were recruited among students of the University of Bologna and none had a previous history of muscle-skeletal, neurological, or generic gait disorders. Their age ranged from 20 to 35 years (mean and standard deviation – SD – of age for women: 24.1 years SD 1.29; for men: 27.4 years SD 3.77), their body mass ranged from 51 to 95 kg (for women: 55.8 kg SD 5.07; for men: 79.4 kg SD 8.42) and their height ranged from 160 to 187 cm (for women: 166.8 cm SD 4.92; for men: 176.8 cm SD 6.20). All these subjects gave informed consent to participate in the study, which was authorized by the local Scientific committee.

Anterior–posterior acceleration at L5 (Fig. 1) was cyclic, with two close positive peaks and a single negative peak. Corresponding signals from the sensors on the forefeet and the preliminary comparison with GA data suggested associating the second positive peak to heel strike. Each step was identified between two consecutive of these peaks, and after the right/left discrimination was obtained successfully for each trial (Fig. 2).

For the identification by the single sensor on L5 of the phases within the gait cycle, acceleration signals from those sensors on the forefeet were initially analyzed (Fig. 3). These signals did not show peaks in correspondence of relevant peaks in L5 sensor acceleration, apart from the toe-off instants, in the vicinity of negative peaks in the L5 sensor. In the present inertial sensing system therefore, the second positive peak of L5 was taken as the instant of foot contact.

Based on this identification, the spatial-temporal parameters were determined; for two of these parameters, values obtained from all subjects are provided (Tables 1 and 2). The values obtained were consistent over subjects, in the range of physiological gait, and in particular these compared well with corresponding values from GA-based measurements. Less reliable results were obtained for the identification of the phases within the gait cycles (Table 3), where significant differences were found for some subjects. In these three tables, most *t*-test values of the two comparisons with the GA gold standard appeared to be smaller when all cycles were included, showing that the two cycles exactly corresponding to those from GA compared better than all possible cycles all together. Overall, when analyzed over the entire subject population, all the duration measurements of the gait cycle phases were found to be statistically different (Table 4).

The good intra- and inter-subject consistency among the subjects of the parameter estimations was also supported by the direct observation of the anterior–posterior acceleration patterns (Fig. 4); the thin standard deviation band demonstrates the small variability. The curve shows a maximum peak precisely at 50% of the cycle, in correspondence of the heel contact of the contralateral, i.e. left, foot. Minimum peaks are at about 10% and 60%, in correspondence of the toe-off events, respectively of the left and right feet. The three positive and the two negative peaks of acceleration have a very similar timing and value over the subjects.

These parameters can be used to provide a variety of information, in particular references from a control group for a better description of abnormalities in the clinical context. In Fig. 5, for example, the symmetry in step timing is illustrated effectively, from the data obtained in the present study. Corresponding values from single or populations of patients can be superimposed for a valuable graphical representation.

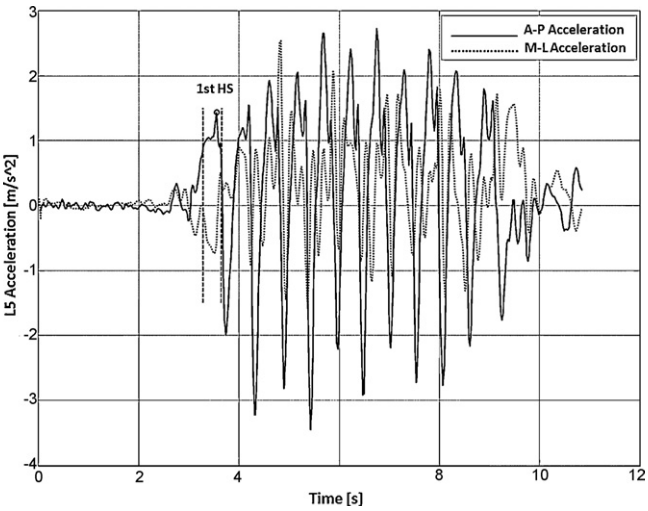

Fig. 2 – Superimposition of anterior–posterior (solid line) and medial-lateral (dotted) acceleration signals from the F4A sensor in a representative subject (#2) during the full exercise of standing (initial 3 s) and walking, after Butterworth low-pass filtering. The interval of the medio-lateral signal (between dashed vertical lines) around the first heel strike (HS) is used for right and left discrimination.

Table 1 – Right stride length as calculated with the GA system and estimated with the F4A system for each subject. For the former, mean and standard deviation are over the 5 repetitions of the two cycles collected, i.e. over 10 values (left columns). The same is reported for F4A only from the corresponding two cycles (central columns), and also from all the collected cycles (right columns). The corresponding *p*-values from the *t*-test are also reported, those with statistical significance are marked with *.

Subjects	Right stride length [m]							
	GA data; two GA cycles		F4A data; two GA cycles		*p*-Values	F4A data; all cycles		*p*-Values
	Mean	Std	Mean	Std		Mean	Std	
#1	1.42	0.01	1.41	0.03	0.681	1.42	0.02	0.584
#2	1.39	0.07	1.35	0.16	0.655	1.42	0.16	0.790
#3	1.54	0.04	1.58	0.07	0.292	1.59	0.10	0.312
#4	1.24	0.03	1.29	0.04	0.060	1.27	0.03	0.106
#5	1.39	0.02	1.38	0.07	0.797	1.37	0.05	0.532
#6	1.44	0.02	1.48	0.15	0.601	1.41	0.09	0.561
#7	1.33	0.02	1.34	0.06	0.919	1.28	0.05	0.065
#8	1.53	0.01	1.61	0.22	0.397	1.63	0.05	0.083
#9	1.35	0.02	1.35	0.06	0.060	1.29	0.05	0.060
#10	1.34	0.02	1.46	0.09	0.056	1.43	0.09	0.063
#11	1.49	0.04	1.48	0.13	0.846	1.48	0.07	0.527
#12	1.36	0.04	1.31	0.13	0.337	1.33	0.07	0.313
#13	1.25	0.03	1.24	0.08	0.993	1.22	0.08	0.422
#14	1.45	0.03	1.55	0.07	0.125	1.50	0.05	0.145
#15	1.43	0.02	1.48	0.07	0.144	1.48	0.09	0.102
#16	1.41	0.02	1.49	0.08	0.059	1.42	0.03	0.489
#17	1.41	0.07	1.25	0.19	0.133	1.40	0.15	0.950
#18	1.40	0.01	1.34	0.09	0.206	1.28	0.10	0.055
#19	1.39	0.01	1.30	0.14	0.242	1.42	0.03	0.093
#20	1.37	0.03	1.33	0.05	0.205	1.32	0.08	0.112
#21	1.25	0.01	0.97	0.10	0.004*	1.21	0.07	0.060
#22	1.29	0.04	1.38	0.18	0.360	1.39	0.09	0.074
Mean	1.38	0.03	1.38	0.10	0.371	1.39	0.07	0.295

Table 2 – Stride duration as calculated with the GA system and estimated with the F4A system for each subject. For the former, mean and standard deviation are over the 5 repetitions of the two cycles collected, i.e. over 10 values (left columns). The same is reported for F4A only from the corresponding two cycles (central columns), and also from all the collected cycles (right columns). The corresponding *p*-values from the *t*-test are also reported.

Subjects	Stride duration [s]							
	GA data; two GA cycles		F4A data; two GA cycles		*p*-Values	F4A data; all cycles		*p*-Values
	Mean	Std	Mean	Std		Mean	Std	
#1	1.00	0.02	0.99	0.02	0.091	1.00	0.03	1.000
#2	1.07	0.03	1.07	0.05	0.876	1.08	0.03	0.335
#3	1.07	0.01	1.06	0.01	0.389	1.09	0.02	0.055
#4	1.11	0.01	1.12	0.02	0.144	1.11	0.02	0.867
#5	1.07	0.01	1.06	0.02	0.172	1.08	0.02	0.744
#6	1.09	0.03	1.08	0.03	0.573	1.09	0.04	0.327
#7	1.14	0.02	1.15	0.02	0.680	1.16	0.01	0.150
#8	1.17	0.02	1.15	0.04	0.130	1.19	0.01	0.096
#9	1.08	0.01	1.08	0.02	0.212	1.08	0.01	0.212
#10	0.93	0.02	0.95	0.02	0.240	0.95	0.03	0.248
#11	1.09	0.02	1.09	0.03	0.914	1.08	0.03	0.630
#12	1.37	0.04	1.28	0.08	0.069	1.32	0.05	0.056
#13	1.06	0.02	1.04	0.02	0.461	1.05	0.02	0.884
#14	1.06	0.02	1.06	0.03	0.921	1.08	0.03	0.310
#15	1.08	0.02	1.07	0.04	0.506	1.09	0.03	0.100
#16	0.96	0.02	0.96	0.02	0.679	0.99	0.05	0.132
#17	1.06	0.02	1.07	0.03	0.823	1.10	0.01	0.123
#18	1.14	0.02	1.13	0.02	0.332	1.16	0.04	0.098
#19	0.99	0.01	0.99	0.01	0.493	1.00	0.02	0.115
#20	1.16	0.02	1.15	0.02	0.849	1.14	0.02	0.303
#21	1.10	0.04	1.08	0.10	0.592	1.10	0.04	0.940
#22	1.23	0.06	1.20	0.05	0.150	1.18	0.07	0.064
Mean	1.09	0.02	1.08	0.03	0.468	1.10	0.03	0.354

Table 3 – Right stance duration as calculated with the GA system and estimated with the F4A system for each subject. For the former, mean and standard deviation are over the 5 repetitions of the two cycles collected, i.e. over 10 values (left columns). The same is reported for F4A only from the corresponding two cycles (central columns), and also from all the collected cycles (right columns). The corresponding *p*-values from the *t*-test are also reported, those significant are marked with *.

Subjects	Right stance duration [s]							
	GA data; two GA cycles		F4A data; two GA cycles		*p*-Values	F4A data; all cycles		*p*-Values
	Mean	Std	Mean	Std		Mean	Std	
#1	58.73	1.94	60.33	2.37	0.412	60.94	1.13	0.112
#2	60.16	1.81	60.26	0.97	0.906	60.94	0.95	0.242
#3	58.08	1.00	62.42	0.55	0.003*	62.70	1.35	0.005*
#4	60.25	0.71	62.57	3.14	0.132	62.21	1.83	0.059
#5	59.00	1.23	65.07	1.87	0.001*	63.06	1.85	0.003*
#6	59.59	0.73	62.29	2.17	0.079	62.69	2.26	0.052
#7	59.44	1.00	61.53	3.82	0.256	63.39	2.66	0.026*
#8	59.38	1.30	63.11	3.14	0.081	62.43	3.15	0.169
#9	57.05	0.93	59.20	2.00	0.131	60.54	2.54	0.015*
#10	58.50	1.11	62.61	2.49	0.057	62.94	1.52	0.153
#11	57.67	0.88	61.90	3.46	0.097	59.11	1.58	0.061
#12	57.40	1.63	60.51	3.61	0.177	60.41	0.44	0.148
#13	59.14	1.14	65.30	4.37	0.084	65.07	1.34	0.051
#14	59.13	0.42	62.29	2.39	0.087	63.42	0.61	0.058
#15	57.53	1.06	59.53	1.29	0.102	62.01	0.85	0.005*
#16	58.00	0.90	60.55	2.35	0.093	61.68	1.76	0.080
#17	59.25	1.40	60.32	0.55	0.382	60.24	0.92	0.506
#18	58.87	0.88	57.28	1.18	0.108	58.67	2.03	0.810
#19	57.07	1.12	60.31	1.83	0.055	62.92	1.21	0.001*
#20	61.79	0.65	61.88	1.49	0.927	63.02	1.31	0.057
#21	62.81	1.48	61.23	2.84	0.241	64.42	2.42	0.128
#22	60.56	0.61	62.11	1.92	0.170	62.91	0.77	0.057
Mean	59.06	1.09	61.48	2.26	0.21	62.08	1.57	0.13

6. Lessons learned

The aim of this study was to introduce and test a new wireless system used to identify standard gait spatial-temporal parameters during level walking. For this purpose, walking at self-selected speed was analyzed in a large group of healthy young subjects by the single sensor F4A and by a complete state-of-the-art GA system. To test the accuracy of the typical gait cycle phase identification, spatial-temporal parameters from the GA system were compared with those

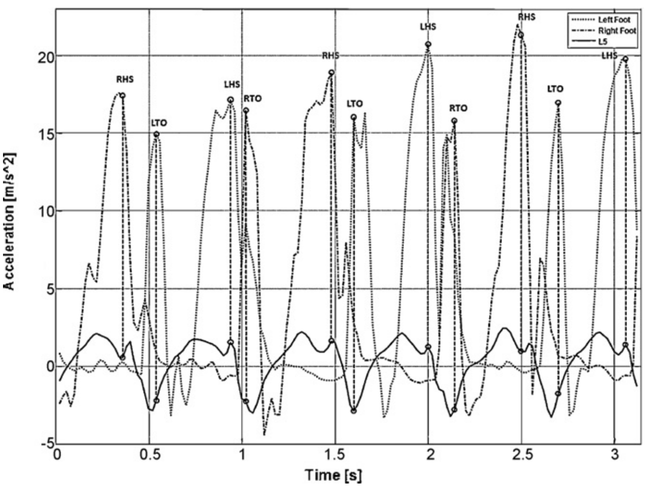

Fig. 3 – Superimposition of typical anterior–posterior acceleration signals (subject #2), from the sensor on L5 (solid line), and from the sensors on the forefeet (dash-dotted for the right and dotted for the left) along a few steps. The timing of the gait-related events as assumed in most of the previous literature is also shown (dashed vertical lines,): right foot heel strike (RHS) and toe-off (RTO), left foot heel strike (LHS) and toe-off (LTO).

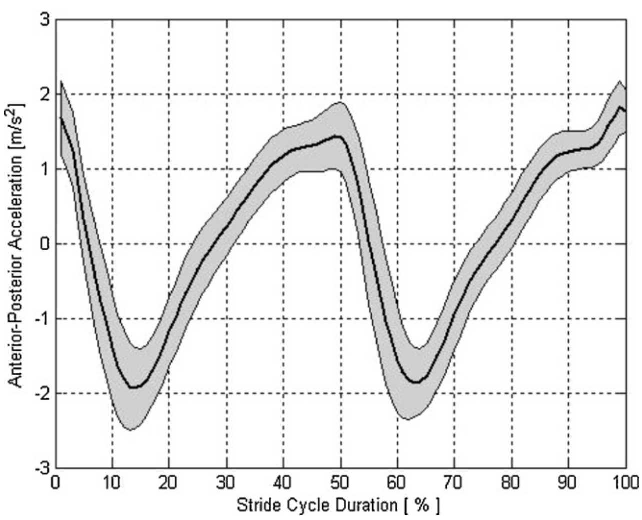

Fig. 4 – Anterior–posterior acceleration over the normalized cycle (0–100%) of the right leg; the cycle taken was the one observed also by GA. Mean (solid curve) and one standard deviation (grey band) are shown as calculated over 5 repetitions of all 22 subjects.

Table 4 – Spatial-temporal parameters as calculated with the GA system and estimated with the F4A system for all the subjects. For the former, mean and standard deviation are over the 5 repetitions of the two cycles collected, i.e. over 10 values (left columns). The same is reported for F4A only from the corresponding two cycles (central columns), and also from all the collected cycles (right columns). The corresponding p-values from the paired t-test or from the Wilconxon non parametric test (the latter in case marked with the subscript $_w$) are also reported, those significant are marked with *.

All subjects spatial-temporal parameters	GA data; two GA cycles		F4A data; two GA cycles		*p*-Values	F4A data; all cycles		*p*-Values
	Mean	Std	Mean	Std		Mean	Std	
Right step length [m]	0.69	0.02	0.69	0.07	0.758_w	0.68	0.05	0.522
Left step length [m]	0.7	0.02	0.73	0.07	0.051	0.72	0.06	0.139
Mean stride length [m]	1.39	0.03	1.40	0.09	0.517	1.40	0.07	0.542
Right stride length [m]	1.38	0.03	1.38	0.10	0.821	1.39	0.07	0.698
Left stride length [m]	1.39	0.03	1.42	0.10	0.188	1.40	0.08	0.685
Stride length/height [%]	80.10	1.66	81.15	5.28	0.362	80.96	4.16	0.379
Right stride length/Height [%]	80.31	1.64	76.40	5.85	0.709_w	80.55	4.16	0.732
Left stride length/height [%]	80.41	1.94	78.77	5.82	0.338_w	80.88	4.42	0.715
Mean stride duration [s]	1.09	0.02	1.08	0.03	0.088	1.10	0.03	0.220
Right step duration [s]	0.54	0.01	0.53	0.03	0.056	0.56	0.06	0.101_w
Left step duration [s]	0.54	0.02	0.54	0.03	0.989	0.57	0.06	0.220_w
Right foot symmetry [%]	50.29	1.27	49.43	2.80	0.101_w	49.96	1.63	0.592_w
Left foot symmetry [%]	49.87	1.31	50.92	2.91	0.06_w	50.11	1.63	0.757
Right stance duration [% stride]	59.06	1.09	61.48	2.26	$<0.001^*_w$	62.08	1.57	<0.001*
Left stance duration [% stride]	58.99	0.98	63.55	2.22	<0.001*	63.05	1.57	<0.001*
Right swing duration [% stride]	40.95	1.09	37.85	2.52	<0.001*	37.44	1.65	$<0.001^*_w$
Left swing duration [% stride]	41.01	0.98	36.79	2.59	<0.001*	36.88	1.62	<0.001*
Double support duration [% stride]	8.96	0.74	12.69	1.41	<0.001*	13.14	1.03	$<0.001^*_w$
Single support duration [% stride]	40.96	1.19	38.31	1.92	$<0.001^*_w$	37.96	1.15	$<0.001^*_w$
Speed [cm/s]	129.39	3.90	125.49	14.59	0.304	127.24	14.16	0.452
Cadence [strides/min]	56.07	1.07	56.33	1.55	0.091	55.54	1.41	0.062
Normalized speed [%]	74.95	2.25	72.95	8.50	0.343	73.74	8.26	0.454

detected by the new system, both based on the two corresponding gait cycles and all gait cycles measured. Data analysis started with the detection of foot contacts from the anterior–posterior acceleration, which was used to calculate all spatial-temporal parameters; medial-lateral acceleration was used only to recognize left and right steps. The comparison gave very encouraging results, with very few significant differences between the two systems. These results reveal that the present spatial-temporal parameter estimation from a single sensor in L5 is the most accurate in the known literature [17–22].

As reported in the literature, the medio-lateral acceleration curve gave valuable information to discriminate between the right and left gait cycles within the analyzed exercise; these cycles were recognized correctly in fact also in the present work for each trial in all subjects. The relevant medial and lateral displacements from double integration of the acceleration were not found equivalently revealing. However, a clear and consistent peak at heel strike, positive or negative, was found in the medio-lateral acceleration curve only at the beginning of the very first gait cycle. Therefore, a criterion based on the sign of the area under the signal curve close to the instant of first foot contact was examined and here proposed, and this was able to discriminate successfully the starting, and the following, step sides. This automation is particularly of value in future clinical assessments, where the starting side cannot be forced in a large number of pathological populations.

To obtain a correct analysis from the present F4A system, the subject must stand still for a few seconds before starting and for a few seconds after stopping to walk to allow automatic detection of the walking part of the exercise. This is necessary because the algorithm subtracts the offset automatically from to the original signal, calculated as the mean value over the first second of time, and also because a certain threshold for the signal is expected to be passed before the first step is recognized as such. In addition, this threshold should be assessed carefully when patients with pathologies affecting severely stability and mobility are assessed. These and other controls are implemented to avoid false identification of walking steps at the beginning and end of the exercise; however, when possible, i.e. when a very large number of steps can be measured with the F4A system, a few steps among the first and last ones should be removed from the following analysis of the spatial-temporal parameters.

The values obtained for the present spatial-temporal parameters compare well with corresponding reference data

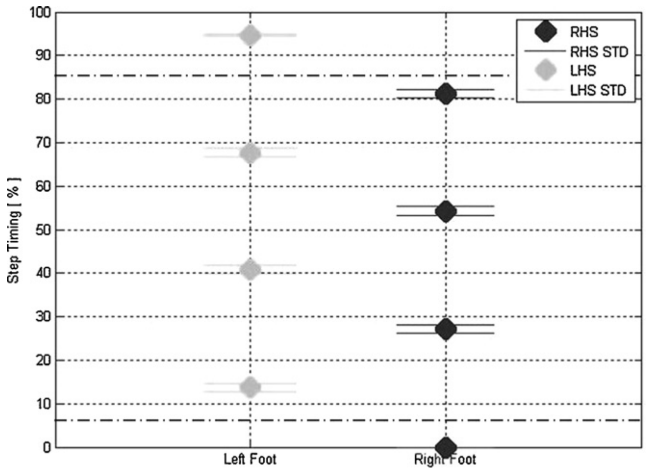

Fig. 5 – A representation of the progression symmetry for clinical applications: percentage duration of each of the first seven steps analyzed as averaged over the 5 repetitions of all 22 subjects; relevant means (circle) and the standard deviations (horizontal segments) are shown. The corresponding GA cycles are also shown (between horizontal dash-dot).

reported in the literature [5,9,29] from large populations of healthy subjects and by a variety of comprehensive instruments. Furthermore, the present results are in agreement also with those reported in previous works where lower trunk accelerometer signals were analyzed [18,19,21]. However, there are a few previous studies reporting these parameters for the gait cycle phases [21]. For these phases, the calculation of the parameters is critical, due to a difficult timing identification for the gait cycle events, associated to the presence of the double positive peaks. This problem might be limited by increasing the present acquisition frequency, for which the typical two-peak curve can no longer be flattened. Furthermore, since the GA system uses a semiautomatic procedure for calculating the gait cycle phases, it might be valuable in the future to compare these parameters given by the F4A system with those obtained by a mat of electronic sensors, acquiring simultaneously the walking performance by the two measurement systems. Finally, the statistically significant difference observed for a number of these measurements does not diminish the overall value of the present novel technique; all spatial-temporal parameters including the various phases within the gait cycle can be obtained reliably despite the simple and cheap instrumentation. This would allow that such measurements can be taken routinely also in standard clinical settings, though relevant validation should be performed.

For most of the parameters calculated from the accelerometer signals, the double comparison as in Tables 1–3, between the two corresponding cycles and between the all cycles collected from the two systems, reveals that the new F4A system is able to represent the gait performance of the subject in the precise cycle association better than in the overall mean values. This can be appreciated by the generally smaller p-values of the second comparison for most of the subjects, and also by the larger number of subjects with statistically significant differences: for stance duration, for example (Table 3), these were two in the precise comparison, six with all cycle comparison. This can be explained in different ways, but especially by irregular patterns at the beginning and end of the exercise, associated to the walking initiation and termination; these can certainly alter the mean calculations over all cycles. Further evidence is the smaller accuracy with which the F4A system estimates the spatial parameters compared with the temporal ones, which is revealed with statistically significant difference particularly by the larger number of subjects. This can be explained by the inevitable errors implied in the necessary double integration of acceleration. Another possible cause is the change in sensor orientation over walking, which results in variable spurious acceleration due to gravity.

F4A proved to be quick and easy to use, the sensors do not impede subject movements and this new overall system provides relevant information on walking performance automatically without the need for expensive gait laboratory instrumentations. The present gait analysis, though limited to spatial-temporal parameters, can be performed in a variety of settings, with very little encumbrance and in a short time, so that large number of subjects and conditions can be assessed and, if necessary, monitored over time. A number of similar systems have been proposed, but the present one is the first based on lower trunk acceleration that is only able to detect automatically spatial-temporal parameters also of the separate gait cycle phases, although in this case the accuracy is a little less. With a single sensor the costs and complexity of the analysis are considerably reduced. The system seems to be especially suitable for a number of clinical studies, designed for example to identify possible pathological gait impairments or evaluate the effects of therapeutic interventions.

7. Future plans

In the near future, some of the algorithms devised for the present system will be investigated for possible further improvements; in particular the ability to detect the timing of the various phases within the gait cycle will be analyzed, by looking at all three acceleration components of the sensor on L5. However, the results obtained for the gait parameter estimation are already very encouraging, therefore the applicability of present system can be exploited in various fields of human movement. In the analysis of the physiology of gait, a much larger normal population size can be analyzed, and the effect of gender, age, weight etc. on gait performance can be addressed quantitatively. The system also has the potential to be of value for future quantitative assessments within routine clinical analyses of pathologies affecting locomotion. In particular, the quality of level walking before and after relevant treatments, as well as over the recommended rehabilitation program, can be monitored with these measurements, which do not require expensive instrumentation, large and special rooms, long patient preparation, and challenging clinical interpretation. Other more complex motor tasks are also going to be investigated, for which specific new algorithms shall be devised. The present automatic and easy calculation of important spatial-temporal parameters can also be used in combination with other measurement systems for physiological assessment of locomotion, such as electromyography and oxygen consumption.

Conflict of interest

One of the authors (GC) is the owner of the company manufacturing the device. This has employed one another author (FB).

Acknowledgments

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