Northumbria Research Link

Citation: Storm, Fabio A., Buckley, Christopher and Mazzà, Claudia (2016) Gait event detection in laboratory and real life settings: Accuracy of ankle and waist sensor based methods. Gait & Posture, 50. pp. 42-46. ISSN 0966-6362

Published by: Elsevier

URL: https://doi.org/10.1016/j.gaitpost.2016.08.012 <https://doi.org/10.1016/j.gaitpost.2016.08.012>

This version was downloaded from Northumbria Research Link: http://nrl.northumbria.ac.uk/id/eprint/43894/

Northumbria University has developed Northumbria Research Link (NRL) to enable users to access the University's research output. Copyright © and moral rights for items on NRL are retained by the individual author(s) and/or other copyright owners. Single copies of full items can be reproduced, displayed or performed, and given to third parties in any format or medium for personal research or study, educational, or not-for-profit purposes without prior permission or charge, provided the authors, title and full bibliographic details are given, as well as a hyperlink and/or URL to the original metadata page. The content must not be changed in any way. Full items must not be sold commercially in any format or medium without formal permission of the copyright holder. The full policy is available online: http://nrl.northumbria.ac.uk/policies.html

This document may differ from the final, published version of the research and has been made available online in accordance with publisher policies. To read and/or cite from the published version of the research, please visit the publisher's website (a subscription may be required.)

Northumbria University
NEWCASTLE

CUSTOMER SERVICE EXCELLENCE CSE UniversityLibrary

Gait & Posture 50 (2016) 42–46

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

Gait event detection in laboratory and real life settings: Accuracy of ankle and waist sensor based methods

Fabio A. Storm[a,b,*], Christopher J. Buckley[a,b,c], Claudia Mazzà[a,b]

[a] Department of Mechanical Engineering, The University of Sheffield, Sheffield, UK
[b] INSIGNEO Institute for in silico Medicine, The University of Sheffield, Sheffield, UK
[c] MRC-Arthritis Research UK Centre for Integrated Research into Musculoskeletal Ageing (CIMA), The University of Sheffield, Sheffield, UK

ARTICLE INFO

Article history:
Received 11 February 2016
Received in revised form 28 July 2016
Accepted 11 August 2016

Keywords:
Inertial sensor
Algorithm
Gait events
Free walking
Temporal parameters

ABSTRACT

Wearable sensors technology based on inertial measurement units (IMUs) is leading the transition from laboratory-based gait analysis, to daily life gait monitoring. However, the validity of IMU-based methods for the detection of gait events has only been tested in laboratory settings, which may not reproduce real life walking patterns. The aim of this study was to evaluate the accuracy of two algorithms for the detection of gait events and temporal parameters during free-living walking, one based on two shank-worn inertial sensors, and the other based on one waist-worn sensor. The algorithms were applied to gait data of ten healthy subjects walking both indoor and outdoor, and completing protocols that entailed both straight supervised and free walking in an urban environment. The values obtained from the inertial sensors were compared to pressure insoles data. The shank-based method showed very accurate initial contact, stride time and step time estimation (<14 ms error). Accuracy of final contact timings and stance time was lower (28–51 ms error range). The error of temporal parameter variability estimates was in the range 0.09–0.89%. The waist method failed to detect about 1% of the total steps and performed worse than the shank method, but the temporal parameter estimation was still satisfactory. Both methods showed negligible differences in their accuracy when the different experimental conditions were compared, which suggests their applicability in the analysis of free-living gait.

1. Introduction

The interest in objective daily monitoring of physical activity in habitual environments is growing for both clinical and research purposes. Among activities of daily living, gait is a major marker of disease progression [1], and the step-by-step determination of gait parameters is required for the analysis and characterization of quasi-periodic motions [2], both in terms of absolute values and of their variability [3].

To avoid altering a subject's natural movement, a necessary requirement during daily physical activity monitoring is that the smallest number of sensors should be positioned in minimally cumbersome locations. Thanks to recent technological advances, wearable sensors based on inertial measurement units (IMUs) have become an ideal choice to capture continuous gait data, playing a crucial role in the transition of gait analysis from traditional assessment carried out in specialised gait laboratories to daily life monitoring [4].

To determine temporal gait parameters, the accurate detection of gait events, such as initial foot contact (IC) and final foot contact (FC) is required. Methods to obtain IC and FC timings from a single IMU positioned on the lower trunk have been proposed in both normal and pathologic gait [5,6]. Several authors have also proposed the use of two synchronized IMUs on the lower limbs, with the shanks being the most popular location [7,8]. The validity of these methods has generally been tested in laboratory settings, during straight walking, and against references such as instrumented mats [9], force platforms [5], and motion capture systems [8], often relying on a limited number of consecutive strides. However, controlled steady-state straight walking conditions that are obtained in a laboratory may not reproduce real life behaviour. Currently it is not known whether the acceleration and angular velocity patterns generated during real life behaviour can affect the accuracy of algorithms tested in the controlled laboratory conditions. Indeed, the variability of stride velocity and gait cycle time during scripted straight walking has been shown to be higher over longer distances (>20 m) in comparison to short distances

---

[*] Corresponding author at: Department of Mechanical Engineering, The Pam Liversidge Building, Sir Frederick Mappin Building, Mappin Street, Sheffield, S1 3JD, UK.

E-mail address: fastorm1@sheffield.ac.uk (F.A. Storm).

http://dx.doi.org/10.1016/j.gaitpost.2016.08.012

(<10 m) [10], and repeated single walking protocols also generated lower variability in gait parameters with respect to continuous overground walking [11]. A recent study using a wearable accelerometry-based pendant showed that variability of step duration during activities of daily living performed in a semi-controlled environment was higher and did not correlate with laboratory gait [12]. These findings suggest that walking strategies may be affected by different experimental conditions, and that this might reflect into different patterns of the signals used to estimate IC and FC event. However, to the best of the authors' knowledge, the accuracy of the estimates of both IC and FC events in free living gait, i.e. carried out in a urban environment has not been yet assessed.

The aim of this study was to test the performance of two different IMU-based methods for gait temporal parameters estimation during gait in free living conditions. One method is based on the use of two shank-worn IMUs [8], and the other on a single waist-worn IMU [9]. These algorithms were selected for their previously reported robustness to changes in IMU attachments and to an individual's gait speed, and for their reported high accuracy [6]. The algorithms were applied to gait data of ten healthy subjects walking in different daily life environments, both indoor and outdoor, and completing protocols that entailed both straight and free walking, and their outputs were compared to data obtained from pressure insoles.

2. Materials and methods

Ten healthy volunteers (3 females, 7 males, age 28 ± 3 y.o.) were recruited for the study. Ethical approval from the University of Sheffield's Research Ethics Committee was obtained, and the research was conducted according to the declaration of Helsinki. All participants provided informed written consent.

Each participant was asked to wear three IMUs (Opal^{TM}, APDM; weight 22 g, size 48.5 mm × 36.5 mm × 13.5 mm) containing a 3-axis accelerometer, a 3-axis gyroscope, and a 3-axis magnetometer. One IMU was positioned on the lower trunk on the fifth lumbar vertebra (L5) [9], with its sensing axes X, Y and Z pointing downward, to the left, and forward, respectively. The other two IMUs were positioned at each ankle, just above the malleoli [8], with X, Y and Z pointing downward, to the right, and backward, respectively. The devices measured accelerations and angular velocities at a sampling frequency of 128 Hz, and the accelerometer range was set at ±6 g. Two pressure-sensing insoles (F-Scan 3000E, Tekscan) were used to obtain IC and FC reference timings. The insoles were cut to fit tightly into each participant's shoe. They were calibrated using a step calibration technique according to manufacturer instructions. Sampling frequency was set at 128 Hz and the gait events were obtained using the ground reaction force (10 N threshold) [13]. A vertical jump was used as a synchronizing event between the IMUs and the insoles in order to realign the two signals coming from both instruments at the beginning of each trial. The equivalency of the nominal sampling frequency of the two instruments was verified on three separate 20-min recordings, where at 1 min intervals a series of impacts clearly detected by both instruments were generated, and showed a consistent mismatch between signals of one sample each two minutes recording (7.8 ms). This mismatch was corrected for in the 15-min free outdoor walking data by realigning the signals each two minutes. This procedure was not needed in the other walking conditions, which lasted less than two minutes.

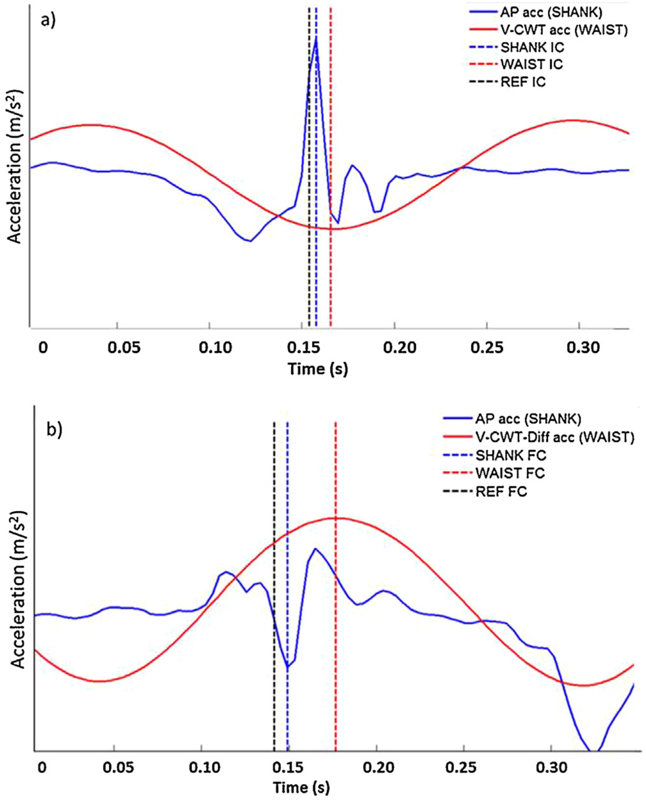

Fig. 1. Gait event detection for the tested algorithms. (a) Anterior-posterior acceleration signal of the shank (*AP acc*, solid blue line), with corresponding IC timings (*SHANK IC*, dashed blue vertical line). Wavelet-filtered pelvis acceleration signal in the vertical axis (*V-CWT acc*, solid red line), with corresponding IC timings (*WAIST IC*, red dashed vertical line). Reference IC timings are also shown (*REF IC*, black dashed vertical line). (b) Anterior-posterior acceleration signal of the shank (*AP acc*, solid blue line), with corresponding FC timings (*SHANK FC*, blue dashed vertical line). Derivative of the wavelet-filtered pelvis acceleration signal in the vertical axis (*V-CWT-Diff acc*, solid red line), with corresponding FC timings (*WAIST FC*, red vertical lines). Reference FC timings are also shown (*REF FC*, black dashed vertical line). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Fig. 1 shows typical signals collected at the shank and pelvis, and the corresponding IC and FC instants for both methods used to compute the temporal gait parameters. In the shank-based method (SHANK), the peak in the angular velocity signals in the sagittal plane during mid-swing is used to identify windows in the signal where no gait events can occur. When coupled with the alternate shank, these intervals allow the identification of search windows for IC and FC events. The IC is identified as the instant of minimum angular velocity in the sagittal plane between the beginning of the IC search window and the instant of maximum anterior-posterior acceleration. The FC is identified as the instant of minimum anterior-posterior acceleration in the FC search window [8]. For the waist-based method (WAIST), data is collected from a single IMU positioned on the lower trunk at L5 level. A first Gaussian continuous wavelet transformation is applied to the vertical acceleration signal, and the minima are identified as the IC timings. The resulting signal is then differentiated and the FC timings are identified as the instants of its maxima [9].

Subjects completed four walking tasks in the conditions detailed in Table 1, and the IMU and pressure insoles data were collected during each task. A stopwatch was used to measure walking time and compute average walking speed during the indoor and outdoor straight walking conditions.

For the outdoor free walking task, participants were instructed to walk freely in the city centre without any restrictions regarding route or walking speed, and avoiding stairs. Both the indoor free walking and outdoor free walking conditions had the potential of recording the participant's turns in addition to straight line walking, both of which were included in the analysis. On the contrary, data recorded during resting or transitory periods were

Table 1
Summary of the walking conditions performed during the experimental protocol, with acronym, a brief description, and the duration or repetition.

Condition	Acronym	Description	Duration/Repetitions
Indoor scripted walking	ISW	Walking at preferred speed along a 20.0 m long walkway.	Eight repetitions.
Outdoor scripted walking	OSW	Walking at preferred speed along a 50.0 m long walkway.	Six repetitions.
Indoor free walking	IFW	Walking along corridors within a university building.	Two minutes.
Outdoor free walking (Short)	OFWS	Walking along footpaths open to the public in the city centre without any restrictions in route or walking speed, avoiding stairs.	Two minutes selected from a fifteen minute walk.
Outdoor free walking (Long)	OFWL	Walking along footpaths open to the public in the city centre without any restrictions in route or walking speed, avoiding stairs.	Fifteen minutes.

excluded from the analysis. To this purpose, once the swing phase of a step was detected from the shank angular velocity signal, all the following steps were retained for the analysis only until when the time distance between subsequent steps was lower than 1 s. The pressure insoles pattern was then visually inspected to verify the absence of anomalies.

For each condition and method, the IC and FC timings were obtained from the IMUs, and used to compute stride, step and stance time. Mean values and their coefficient of variation (CV) were computed. The coefficient of variation is a standardized measure of dispersion and is the ratio of the standard deviation to the mean for each temporal parameter.

For the statistical analysis, the outdoor free walking data was split into two datasets. To allow a comparable amount of strides and a fair comparison with the other tested walking conditions, the OFWS data included two minutes of arbitrarily selected outdoor free walking data. This analysis period was chosen for each participant by selecting an interval of two consecutive minutes starting from a randomly identified sampled instant of time between the beginning of the trial and the end of the 13th minute of test. In addition, to reflect the entire outdoor free walking condition, all the fifteen minutes of outdoor free walking were included in the OFWL condition. Missing and extra gait events were also counted and included in the study.

For each method, the absolute error for each estimated parameter (IC, FC, stride time (mean and CV), step time (mean and CV), and swing time (mean and CV)) was determined as follows:

$$|E| = |p - p_r| \tag{1}$$

Where p_r is the reference value of the parameter p.

Descriptive statistics for |E| (mean and standard deviation values) were determined for each subject, and the resulting group averages and standard deviations were finally computed.

A Shapiro–Wilk test was performed to check for data normality. For each method and each parameter a Friedman Test for non-normal distribution was then used to compare the |E| values obtained in the different walking conditions, with a significance level of 0.05. Post-hoc tests with Bonferroni correction were also performed to test if there were significant differences between indoor controlled walking (ICW) and the remaining walking conditions.

3. Results

The total number of gait cycles analysed in the ISW, OSW, IFW, OFWS, and OFWL conditions were 94 ± 17, 121 ± 11, 188 ± 16, 132 ± 40, and 767 ± 119, respectively. The participants completed a median of 120 consecutive strides during the OFWL condition, while during the indoor free walking task, the median number of consecutive strides was 30. The SHANK method detected 100% of both IC and FC events. The WAIST method showed 29 missing IC events in each of both OFWS and IFW condition, corresponding to 1.3% of the total number of analysed steps. In the OFWL condition, a total number of 124 missing IC events over the 10 participants were detected, corresponding to 0.7% of the total analysed steps. The missing events were evenly distributed across participants, with the exception of one outlier, adding up 58 missing IC events. No missing events were found in the OSW and ISW conditions. Furthermore, no missing FC events were found for the WAIST method in any of the investigated walking conditions. Average recorded walking speeds during indoor and outdoor scripted walking were 1.44 ± 0.10 m/s and 1.51 ± 0.11 m/s, respectively. The descriptive statistics for stride time, step time and stance time as estimated by the pressure insoles used as reference are shown in Table 2. Descriptive statistics (mean and SD) for gait events (IC and FC) and temporal parameters absolute error (|E|) are listed in Table 3.

For the SHANK method, the Friedman test showed that the absolute errors associated with FC timing, stride time, step time and stance time were significantly different between conditions ($p < 0.05$). Pairwise comparisons showed that |E| were significantly smaller during indoor scripted walking (ISW) than those obtained in the outdoor free condition for stride time (both OFWS and OSWL) and step time (only OSWS). For FC timing and stance time, errors were significantly larger in the indoor scripted condition

Table 2
Mean and SD values of temporal gait parameters for all walking conditions.

Parameter	ISW	IFW	OSW	OFWS	OFWL
Stride Time (s)	1.05 ± 0.06	1.06 ± 0.06	1.03 ± 0.05	1.05 ± 0.07	1.06 ± 0.08
Step Time (s)	0.53 ± 0.03	0.53 ± 0.03	0.52 ± 0.02	0.52 ± 0.03	0.53 ± 0.04
Stance Time (s)	0.64 ± 0.05	0.64 ± 0.05	0.63 ± 0.04	0.64 ± 0.05	0.64 ± 0.06
Stride Time CV (%)	1.54 ± 0.37	2.88 ± 1.08	2.21 ± 0.30	3.02 ± 0.95	3.99 ± 1.21
Step Time CV (%)	2.58 ± 0.91	3.87 ± 1.40	3.21 ± 0.57	4.32 ± 1.09	5.11 ± 1.33
Stance Time CV (%)	2.44 ± 0.84	3.58 ± 1.29	2.91 ± 0.44	3.94 ± 1.27	4.99 ± 1.31

Table 3
Mean (±SD) values of the absolute error |E| for IC timing, FC timing, and temporal parameters (mean and CV) of both methods (SHANK and WAIST). *Statistically significant difference between walking conditions ($p < 0.05$).

Method	Condition	ISW	IFW	OSW	OFWS	OFWL	p<0.05 between conditions
SHANK	**IC (ms)**	12 ± 11	11 ± 9	12 ± 7	14 ± 9	14 ± 8	
	FC (ms)	51 ± 21	50 ± 19	37 ± 16	41 ± 22	41 ± 20	ISW vs OSW
	Stride Time (ms)	6 ± 2	6 ± 2	7 ± 3	9 ± 4	9 ± 4	ISW vs OFWS; ISW vs OFWL
	Step Time (ms)	9 ± 4	9 ± 3	10 ± 5	14 ± 8	13 ± 6	ISW vs OFWS
	Stance Time (ms)	44 ± 13	43 ± 14	28 ± 12	32 ± 16	37 ± 15	ISW vs OSW; ISW vs OFWS
	Stride Time CV (%)	0.11±0.06	0.09±0.10	0.13±0.08	0.10±0.14	0.09±0.06	
	Step Time CV (%)	0.33±0.30	0.39±0.26	0.49±0.32	0.89±1.49	0.46±0.65	
	Stance Time CV (%)	0.47±0.52	0.44±0.45	0.36±0.35	0.54±0.62	0.52±0.36	
WAIST	**IC (ms)**	46 ± 20	49 ± 19	50 ± 17	53 ± 22	48 ± 21	
	FC (ms)	76 ± 21	79 ± 19	75 ± 16	77 ± 21	77 ± 23	
	Stride Time (ms)	6 ± 1	8 ± 1	8 ± 1	11 ± 3	11 ± 2	ISW vs OFWS; ISW vs OFWL
	Step Time (ms)	9 ± 3	10 ± 3	9 ± 2	12 ± 2	13 ± 3	ISW vs OFWS; ISW vs OFWL
	Stance Time (ms)	31 ± 12	31 ± 10	26 ± 9	30 ± 11	32 ± 10	
	Stride Time CV (%)	0.08±0.07	0.10±0.10	0.09±0.04	0.31±0.25	0.15±0.13	ISW vs OFWS
	Step Time CV (%)	0.64±0.38	0.37±0.21	0.29±0.22	0.62±0.37	0.48±0.21	
	Stance Time CV (%)	0.43±0.32	0.46±0.39	0.29±0.25	0.30±0.25	0.37±0.26	

(ISW) than those obtained in the outdoor scripted condition (OSWS). In addition, stance time absolute error during indoor scripted walking (ISW) was also significantly larger than during outdoor free walking (OSWS). There were no statistically significant differences in CV absolute errors between walking conditions for any of the temporal parameters investigated.

For the WAIST method, the Friedman test showed that the absolute errors associated with stride time and step time were significantly different between conditions ($p < 0.05$). Both parameters were significantly smaller during indoor scripted walking (ISW) than during outdoor free walking (OFWS and OFWL). In addition, step time error in the indoor scripted condition (ISW) was smaller than during indoor free walking (IFW). For gait variability measures, the |E| associated with stride time CV was found to be significantly different between the ISW and the OFWS condition.

4. Discussion

This study aimed to evaluate the accuracy of two IMU-based algorithms for the detection of gait events during free living gait, which is a necessary step towards the implementation of these methods for prolonged physical activity monitoring. Two methods were selected, named the SHANK, which was applied to data from shank-worn sensors, and the WAIST, which was applied to data form a waist-worn IMU. The SHANK method resulted more accurate than the WAIST method for both IC and FC timings. This was an expected finding since sensors that are in closer proximity to the foot-ground contact point have been already shown to be facilitated in gait events detection [8].

The results for the SHANK method across all the walking conditions provided further evidence for the robustness of this algorithm in limiting the risks of extra or missed events. In contrast to a previously published validation study in healthy subjects [6], including only straight walking conditions, the WAIST method showed some missed gait events during the free walking conditions. This confirms that attention should be paid when interpreting data collected from just one sensor on the pelvis to quantify the number of steps walked over a certain period of time [14], with an error of about 1% to be expected if using the method here investigated.

For the SHANK method, the FC timings were less accurate than the IC timings throughout all the tested conditions. This has previously been reported in literature for the indoor controlled conditions, and is likely due to the smoother movement occurring during FC making the gait event less apparent to detect [8]. For the WAIST method, IC and FC absolute errors were similar: this is likely to be due to stricter filtering applied to the signal in this algorithm.

The accuracy of the SHANK method in estimating IC timings was similar to that reported by the authors who proposed it during scripted straight walking [8], however FC timings in the present study were relatively less accurate in all the walking conditions. Accuracy estimates of the WAIST method were poorer than those reported by the original paper [9] for both IC timings and FC timings, obtained during indoor scripted walking, but similar to those reported in a subsequent validation study [15]. Possible reasons for these inconsistencies include the use of different measurement instruments, different reference methods, different path lengths between protocols, and population characteristics. Overall, stride and step time absolute errors for both methods were limited to absolute error values between 6 ms and 14 ms, while stance time error increased to up to 44 ms (SHANK) and 32 ms (WAIST). These results suggest that stride and step time were reasonably accurate, while stance time should be interpreted with more caution. For the WAIST method, stride time and step time absolute error estimates were less accurate during outdoor free walking (OFWS and OFWL). Although these differences were consistent and resulted to be statistically significant, they generated only a small increase in absolute error (6–11 ms for stride time, 9–13 ms for step time). This outcome suggests that the accuracy of the algorithm is affected by the walking conditions tested. However, it is encouraging to note that the increase in gait event timing and relevant temporal parameter errors were only moderate and should not prevent the use of this method to collect data during prolonged free living gait. Furthermore, the results suggest that in our study, the temporal parameters estimation errors were not markedly influenced by the length of the walks. Similar to the WAIST method, the stride and step time absolute errors recorded using the SHANK method were higher during outdoor free walking (OFWS and OFWL), but generated only a small increase in percentage error (6–9 ms for stride time, and

9–14 ms for step time). Surprisingly, the errors generated for FC timings and stance times were significantly higher during indoor than during outdoor straight walking. The delayed detection of FC events (as shown in Fig. 1) increased in the ISW task as a consequence of a delayed appearance of the minimum in the anterior-posterior acceleration identified as the instant of FC. If confirmed by further studies, this finding may suggest that the environment plays a role in generating different walking patterns and signals, influencing the accuracy of the FC detection.

The absolute errors generated in the computation of CV values for both methods were acceptable and similar across walking conditions, with maximum |E| of 0.13% and 0.31% in stride time CV, 0.89% and 0.64% in step time CV, and 0.54 and 0.46% in stance time CV (values are for SHANK and WAIST methods, respectively). In terms of accuracy in estimating variability of the investigated temporal parameters, generally the two methods appeared to perform similarly. Previous studies have shown that small errors in gait event detection may affect variability measures more than mean values [16]. The fact that no significant differences in accuracy were found between walking conditions for the SHANK method is encouraging and provides evidence for the appropriateness of its use in free-living studies.

The results of this study might represent a normative reference for future investigations of real life gait monitoring in healthy adults. However, if aiming at different applications, such as those involving patient populations, these results cannot be generalised and the accuracy of the algorithms should be specifically tested to account for possible additional errors.

5. Conclusions

Overall, both methods tested in the present study showed small differences in gait event timings and temporal parameter estimation, for both mean and variability measures, between different environments and different walking protocols. Consequently, this is encouraging for the application of these methods in free living gait.

During outdoor free walking, the SHANK method showed very accurate initial contact timing detection, leading to low errors for stride time and step time. Relative to the IC timing, the final contact timing was less accurate. The WAIST method performed worse than the SHANK method in both step detection and in initial and final contact detection; however, these errors only marginally affected the temporal parameter estimation during outdoor free walking.

Conflict of interest statement

None.

Acknowledgements

This study was supported by the EPSRC Frontier Engineering Awards, Grant Reference No. EP/K03877X/1 and by the MRC and Arthritis Research UK as part of the MRC—Arthritis Research UK Centre for Integrated research into Musculoskeletal Ageing (CIMA). The data used in this paper are publically available (DOI: 10.15131/shef.data.3503180).

References

[1] S. Del Din, A. Godfrey, L. Rochester, Validation of an accelerometer to quantify a comprehensive battery of gait characteristics in healthy older adults and Parkinson's disease: toward clinical and at home use, IEEE J. Biomed. Health Inform. 2194 (2015) 1–10, doi:http://dx.doi.org/10.1109/JBHI.2015.2419317.

[2] J.J. Kavanagh, H.B. Menz, Accelerometry: a technique for quantifying movement patterns during walking, Gait Posture 28 (2008) 1–15, doi: http://dx.doi.org/10.1016/j.gaitpost.2007.10.010.

[3] J.M. Hausdorff, Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking, Hum. Mov. Sci. 26 (2007) 555–589, doi:http://dx.doi.org/10.1016/j.humov.2007.05.003.

[4] S.A. Lowe, G. Ólaighin, Monitoring human health behaviour in one's living environment: a technological review, Med. Eng. Phys. 36 (2014) 147–168, doi: http://dx.doi.org/10.1016/j.medengphy.2013.11.010.

[5] W. Zijlstra, A. Hof, Assessment of spatio-temporal gait parameters from trunk accelerations during human walking, Gait Posture 18 (2003) 1–10.

[6] D. Trojaniello, A. Ravaschio, J.M. Hausdorff, A. Cereatti, Comparative assessment of different methods for the estimation of gait temporal parameters using a single inertial sensor: application to elderly, post-stroke, Parkinson's disease and Huntington's disease subjects, Gait Posture 42 (2015) 310–316, doi:http://dx.doi.org/10.1016/j.gaitpost.2015.06.008.

[7] A. Salarian, H. Russmann, F.J.G. Vingerhoets, C. Dehollain, Y. Blanc, P.R. Burkhard, et al., Gait assessment in Parkinson's disease: toward an ambulatory system for long-term monitoring, IEEE Trans. Biomed. Eng. 51 (2004) 1434–1443, doi:http://dx.doi.org/10.1109/TBME.2004.827933.

[8] D. Trojaniello, A. Cereatti, E. Pelosin, L. Avanzino, A. Mirelman, J.M. Hausdorff, et al., Estimation of step-by-step spatio-temporal parameters of normal and impaired gait using shank-mounted magneto-inertial sensors: application to elderly, hemiparetic, parkinsonian and choreic gait, J. Neuroeng. Rehabil. 11 (2014) 152, doi:http://dx.doi.org/10.1186/1743-0003-11-152.

[9] J. McCamley, M. Donati, E. Grimpampi, C. Mazzà, An enhanced estimate of initial contact and final contact instants of time using lower trunk inertial sensor data, Gait Posture 36 (2012) 316–318, doi:http://dx.doi.org/10.1016/j.gaitpost.2012.02.019.

[10] B. Najafi, J.L. Helbostad, R. Moe-Nilssen, W. Zijlstra, K. Aminian, Does walking strategy in older people change as a function of walking distance? Gait Posture 29 (2009) 261–266, doi:http://dx.doi.org/10.1016/j.gaitpost.2008.09.002.

[11] K.L. Paterson, N.D. Lythgo, K.D. Hill, Gait variability in younger and older adult women is altered by overground walking protocol, Age Ageing 38 (2009) 745–748, doi:http://dx.doi.org/10.1093/ageing/afp159.

[12] M. Brodie, M. Coppens, S.R. Lord, N.H. Lovell, Y.J. Gschwind, S.J. Redmond, et al., Wearable pendant device monitoring using new wavelet-based methods shows daily life and laboratory gaits are different, Med. Biol. Eng. Comput. (2015), doi:http://dx.doi.org/10.1007/s11517-015-1357-9.

[13] J. Mickelborough, M.L. Van Der Linden, J. Richards, A.R. Ennos, Validity and reliability of a kinematic protocol for determining foot contact events, Gait Posture 11 (2000) 32–37, doi:http://dx.doi.org/10.1016/S0966-6362(99)00050-8.

[14] F.A. Storm, B.W. Heller, C. Mazzà, Step detection and activity recognition accuracy of seven physical activity monitors, PLoS One 10 (2015) e0118723.

[15] D. Trojaniello, A. Cereatti, U. Della Croce, Accuracy, sensitivity and robustness of five different methods for the estimation of gait temporal parameters using a single inertial sensor mounted on the lower trunk, Gait Posture 40 (2014) 487–492, doi:http://dx.doi.org/10.1016/j.gaitpost.2014.07.007.

[16] T.R. Beijer, S.R. Lord, M.A.D. Brodie, Comparison of handheld video camera and GAITRite measurement of gait impairment in people with early stage Parkinson's disease: a pilot study, J. Parkinsons Dis. 3 (2013) 199–203, doi: http://dx.doi.org/10.3233/JPD-130179.