Validated Assessment of Gait Sub-Phase Durations in Older Adults using an Accelerometer-based Ambulatory System

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Abstract: Validated extraction of gait sub-phase durations using an ambulatory accelerometer-based system is a current unmet need to quantify subtle changes during the walking of older adults. In this paper, we describe (1) a signal processing algorithm to automatically extract not only durations of stride, stance, swing, and double support phases, but also durations of sub-phases that refine the stance and swing phases from foot-worn accelerometer signals in comfortable walking of older adults, and (2) the validation of this extraction using reference data provided by a gold standard system. The results show that we achieve a high agreement between our method and the reference method in the extraction of (1) the temporal gait events involved in the estimation of the phase/sub-phase durations, namely heel strike (HS), toe strike (TS), toe-off (TO), maximum of heel clearance (MHC), and maximum of toe clearance (MTC), with an accuracy and precision that range from –3.6 ms to 4.0 ms, and 6.5 ms to 12.0 ms, respectively, and (2) the gait phase/sub-phase durations, namely stride, stance, swing, double support phases, and HS to TS, TO to MHC, MHC to MTC, and MTC to HS sub-phases, with an accuracy and precision that range from –4 ms to 5 ms, and 9 ms to 15 ms, respectively, in comfortable walking of a thirty-eight older adults ((mean ± standard deviation) 71.0 ± 4.1 years old). This demonstrates that the developed accelerometer-based algorithm can extract validated temporal gait events and phase/sub-phase durations, in comfortable walking of older adults, with a promising degree of accuracy/precision compared to reference data, warranting further studies.

1 INTRODUCTION

Accelerometer-based systems have been used as a reliable solution for the human gait analysis (e.g., Moe-Nilssen et al., 2004; Hartmann et al., 2009; Rueterbories et al., 2010). Their hardware part has the advantage to include low-cost, small, and lightweight accelerometer units with an easy handling and generally low power consumption. The use of these accelerometer-based systems is particularly relevant for the gait analysis of older adults considering the growing interest of using the gait pattern as a marker of risk of negative clinical outcomes or as a marker of robustness (e.g., Gillain et al., 2015). However, there is a current unmet need in terms of the extraction of gait sub-phases that allow the partitioning of the gait cycle into refined parameters, such as the swing sub-phases. These refined gait parameters could have an advantage in quantifying subtle changes during the walking of older adults. Indeed, an increased step variability has been reported to be linked to a higher fall-risk or fall history (Hausdorff et al., 2001; Allali et al., 2017).

In this context, we developed a signal processing algorithm to automatically extract validated gait events, namely *heel strike* (HS), *toe strike* (TS), and *toe-off* (TO), from three-axis accelerometer signals measured at the level of the heel and toe of the right and left feet during the walking of young and healthy subjects (Boutaayamou et al., 2015). This algorithm uses a segmentation method that roughly detects relevant signal sub-regions (Boutaayamou et al., 2017a). Gait events are further extracted with high accuracy and precision in these signal sub-regions.

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In this paper, we extend and modify this algorithm to automatically extract (1) times of occurrence of HS, TS, TO, and newly considered gait events, namely *maximum of heel clearance* (MHC) and *maximum toe clearance* (MTC), and (2) durations of stride, stance, swing, and double support phases, and durations of sub-phases that refine the stance and swing phases from foot-worn accelerometer signals in comfortable walking of older adults. In addition, we consider a stride-by-stride validation of this extraction using reference gait events and gait phase/sub-phase durations provided by a reference kinematic method (used as gold standard).

2 METHODS

2.1 Participants and Gait Setting

Volunteers, who were included in a two-years prospective research for the Gait Analysis and Brain Imagery (GABI) study, participated to the walking tests considered in this paper (Gillain et al., 2017). Briefly, the goal of the GABI study is to highlight the gait parameters associated with the fall risk in the community of old people without fall history. Inclusion criteria included: being at least 65 years old, living independently at home, being able to reach the motion analysis laboratory, and being able to sign inform consent. Exclusion criteria included: fall history in the previous year, the need of walking aids, gait disorders, and/or an increased fall risk related to a neurological or osteo-articular disease (e.g., Parkinson disease, polyneuropathy, stroke, lumbar conflict, etc.), dementia, recent hip or knee prosthesis (≤ 1 year), musculoskeletal pain during walking, an acute respiratory or cardiac illness (< 6 month), a recent hospitalization (< 3 month), non-treated or insufficiently treated co-morbidities (e.g. HTA, diabetes, etc.), and a cardiac pacing. The local ethic committee of the University hospital of Liège (CHU Liège, Belgium) approved the protocol and all participants signed informed consent.

In the context of the present study, gait signals were recorded during comfortable walking speed of thirty-eight older adults (21 women and 17 men), with (mean ± standard deviation) age = 71.0 ± 4.1 years; height = 166 ± 11 cm; weight = 71 ± 15 kg, body mass index = 25,6 ± 3.8 kg/m².

All participants were equipped with four small three-axis accelerometer units (2 cm x 1 cm x 0.5 cm; range ±12 g). These four accelerometer units were directly attached to the heel and toe of each shoe. Our accelerometer ambulatory system synchronously recorded gait data at 200 Hz from these four accelerometer units. A detailed description of the ambulatory accelerometer system is given in (Boutaayamou et al., 2017a). The participants wore their own regular shoes and were also equipped with four active markers. Each marker was attached on each accelerometer unit, i.e., the four markers were also attached to the shoes at the level of the heel and toe. A four-camera Codamotion system (Charnwood Dynamics; UK) recorded gait data from these active markers at 200 Hz, during 60 seconds for each gait test. The participants were asked to walk along a track in a wide, clear, and straight hallway, at their preferred, self-selected usual speed and by looking forward to the walking direction. Each participant walked a total distance of 99 m following the trajectories shown in Figure 1. We consider here only gait data that were recorded according to straight walking lines, i.e., during non-turning walking episodes. All the walking tests were performed at the Laboratory of Human Motion Analysis of the University of Liège, Belgium.

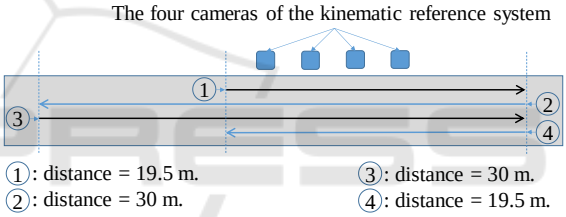


Figure 1: Experimental setting illustrating the walking trajectories with a total distance of 99 m. This total distance must be covered by each participant during a gait test in a comfortable walking speed.

2.2 Algorithm Development and Validation Method

In order to accurately and precisely quantify the durations of the stance and swing phases and their associated sub-phases, during a gait cycle (i.e., the duration of a stride phase), it is important to extract, during the same gait cycle, accurate and precise moments of gait events involved in the estimation of these phase/sub-phase durations.

The proposed extraction algorithm uses distinctive and remarkable features on both longitudinal and antero-posterior accelerations of the heel and toe for each foot. Depending on the nature of these features, a suitable method is employed to accurately and precisely extract gait events of interest. For clarity, we consider only one foot for the description of the algorithm. The algorithm would be applied in the same way for the left and right foot. We

consider, hereafter, sagittal (heel/toe) accelerations of each foot to identify the times of occurrence of the gait events, namely HS_{accel}, TS_{accel}, TO_{accel}, MHC_{accel}, and MTC_{accel}. The subscript *accel* refers to our method. All data were analyzed using Matlab R2009b (MathWorks, USA).

2.2.1 Extraction of HS, TS, TO, MHC, and MTC from Accelerometer Data

In the present study, we adapt the method described in (Boutaayamou et al., 2015) to extract TO from the vertical heel acceleration (Figure (1-bottom)). HS and TS were extracted from the vertical heel acceleration (Figure (1-bottom)) and vertical toe acceleration (Figure (2-bottom)), respectively; the detailed description of their extraction in the walking of older adults is beyond the scope of the present paper and will be considered in a future paper. Rather, we describe the newly developed method for the extraction of the gait events for refining the swing phase, namely MHC and MTC.

The algorithm extracts first (in this order) HS, TS, TO, and MTC before it extracts MHC:

- The time of the maximum of the toe clearance event: MTC_{accel}.

MTC_{accel} is defined as the moment when the toe accelerometer reaches its maximum position during the swing phase. We consider distinctive vertical toe acceleration features that indicate where MTC can be found in the time domain.

As MTC_{accel} occurs after TO and before the heel strike of the next stride, denoted by HS_{2accel}, we seek MTC_{accel} in the segment $[TO_{accel} + 0.4*(HS_{2accel} - TO_{accel}), HS_{2accel}]$. MTC_{accel} is automatically extracted in the vertical toe acceleration restricted to this segment. The resulting local signal is then filtered with a 4th-order zero-lag Butterworth low-pass filter (cutoff frequency = 7 Hz). We then detect the local minimum, t_{min}, in this filtered signal.

We consider a second local segment defined from the restriction of the vertical toe acceleration to the interval $[TO_{accel}, t_{min} + 0.4*(HS_{2accel} - TO_{accel})]$. This local segment is filtered with a 4th-order zero-lag Butterworth low-pass filter (cutoff frequency = 11 Hz). Based on the resulting local filtered signal, we define a remarkable point, t_{cz}, that corresponds to the time when a zero crossing of this resulting local filtered signal occurs before t_{min}. It is then assumed that MTC_{accel} is the time $t_{cz} + 0.75*(t_{min} - t_{cz})$.

- The time of the maximum of the heel clearance event: MHC_{accel}.

MHC_{accel} is defined as the moment when the maximum clearance between the heel accelerometer and the ground is achieved during the swing phase. In contrast to (Boutaayamou et al., 2017a), where MHC_{accel} event was extracted from the vertical heel acceleration, we consider distinctive vertical toe acceleration features that indicate where MHC_{accel} can be found in the time domain. MHC_{accel} uses the previously extracted t_{cz} and TO_{accel}, and it is assumed that MHC_{accel} is the time $TO_{accel} + 0.18*(TO - t_{cz})$.

2.2.2 Extraction of HS, TS, TO, MHC, and MTC from Kinematic System Data

Reference gait events, i.e., HS_{ref}, TS_{ref}, TO_{ref}, MHC_{ref}, and MTC_{ref} were extracted from the vertical coordinates of the left/right heel and toe markers (gold standard) during consecutive strides to validate, on a stride-by-stride basis, the considered left/right gait events and phase/sub-phase durations (Figure 2). The subscript *ref* refers to the reference method. More details about the extraction of these reference data are given in (Boutaayamou et al., 2015).

2.2.3 Extraction of Gait Phase/Sub-Phase Durations

Left/right temporal gait phases, such as durations of left/right stance, swing, stride, and double support phases, are calculated on the basis of the previous gait events as follows:

- Left stride duration (time between two consecutive left HSs)

$$Left\ stride = HS_{left}(i+1) - HS_{left}(i).$$

- Right stride duration (time between two consecutive right HSs)

$$Right\ stride = HS_{right}(i+1) - HS_{right}(i).$$

- Left stance duration (time between left HS (HS_{left}) and left TO (TO_{left}) during stride i)

$$Left\ stance = TO_{left}(i) - HS_{left}(i).$$

- Right stance duration (time between right HS (HS_{right}) and right TO (TO_{right}) during stride *i*)

$$Right\ stance = TO_{right}(i) - HS_{right}(i).$$

- Left swing duration (time between HS_{left} of stride *i*+1 and TO_{left} of stride *i*)

$$Left\ swing = HS_{left}(i+1) - TO_{left}(i).$$

- Right swing duration (time between HS_{right} of stride *i*+1 and TO_{right} of stride *i*)

$$Right\ swing = HS_{right}(i+1) - TO_{right}(i).$$

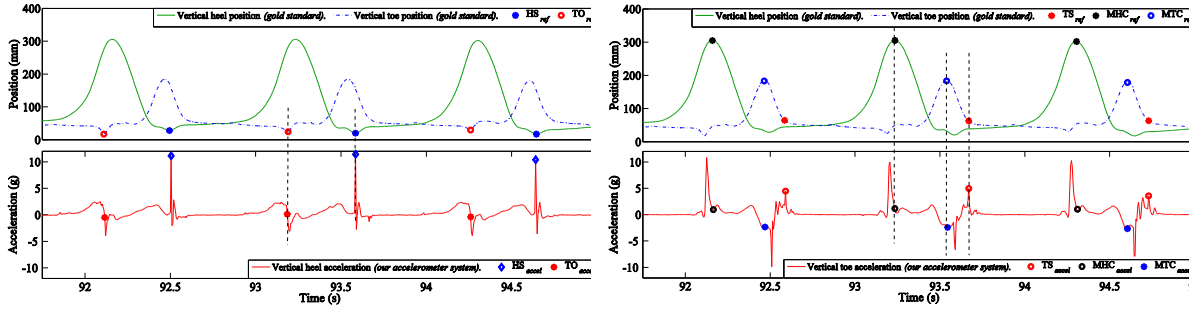


Figure 2: Left/right reference gait events, i.e., (1-top) heel strike (HS_{ref}) and toe-off (TO_{ref}), and (2-top) toe-strike (TS_{ref}), maximum of heel clearance (MHC_{ref}), and maximum of toe clearance (MTC_{ref}) were extracted from the vertical coordinates of left/right heel and toe markers (gold standard). Left/right accelerometer gait events, i.e., (1-bottom) HS_{accel} and TO_{accel}, and (2-bottom) TS_{accel}, MHC_{accel}, and MTC_{accel} were extracted from left/right vertical heel and toe accelerations (our accelerometer system). These gait events are shown on each signal to illustrate the stride-by-stride validation method.

- Left double support duration (time between TO_{left} and $HS_{leftright}$ during stride i)

  Left double support = TO_{left} (i) – HS_{right} (i).

- Right double support duration (time between TO_{right} and HS_{left} during stride i)

  Right double support = TO_{right} (i) – HS_{left} (i).

We also use the gait events TS, MHC, and MTC to calculate the left/right gait sub-phase durations as:

- Left HS2TS duration (time between left TS (TS_{left}) and HS_{left} of stride i)

  Left HS2TS = TS_{left} (i) – HS_{left} (i).

- Right HS2TS sub-phase duration (time between right TS (TS_{right}) and HS_{right} of stride i)

  Right HS2TS = TS_{right} (i) – HS_{right} (i).

- Left TO2MHC sub-phase duration (time between left MHC (MHC_{left}) and TO_{left} during stride i)

  Left TO2MHC = MHC_{left} (i) – TO_{left} (i).

- Right TO2MHC sub-phase duration (time between right MHC (MHC_{right}) and TO_{right} during stride i)

  Right TO2MHC = MHC_{right} (i) – TO_{right} (i).

- Left MHC2MTC sub-phase duration (time between MHC_{left} and left MTC (MTC_{left}) of stride i)

  Left MHC2MTC = MTC_{left} (i) – MHC_{left} (i).

- Right MHC2MTC sub-phase duration (time between MHC_{right} and right MTC (MTC_{right}) of stride i)

  Right MHC2MTC = MTC_{right} (i) – MHC_{right} (i).

- Left MTC2HS sub-phase duration (time between HS_{left} and MTC_{left} of stride i)

  Left MTC2HS = HS_{left} (i) – MTC_{left} (i).

- Right MTC2HS sub-phase duration (time between HS_{right} and MTC_{right} of stride i)

  Right MTC2HS = HS_{right} (i) – MTC_{right} (i).

2.2.4 Evaluation Method

We evaluated the level of agreement between our method and the reference method by quantifying, on a stride-by-stride basis, (1) the accuracy and precision in the extraction of the gait events, and (2) the mean error and absolute error in the extraction of the phase/sub-phase durations.

Accuracy and precision were computed as the mean and standard deviation (std. dev.), respectively, of the differences between the gait events for each stride, i.e., $HS_{accel} - HS_{ref}$, $TS_{accel} - TS_{ref}$, $TO_{accel} - TO_{ref}$, $MHC_{accel} - MHC_{ref}$, and $MTC_{accel} - MTC_{ref}$.

The mean error and the absolute error were calculated as the mean and std. dev. of the differences between the phase/sub-phases durations from our method and those from the gold standard, and the mean and std. dev. of absolute values of these differences, respectively. Bland-Altman plots were also created to evaluate the difference (1) between the extracted gait events, and (2) between the phase/sub-phases durations from our method and those from the reference method.

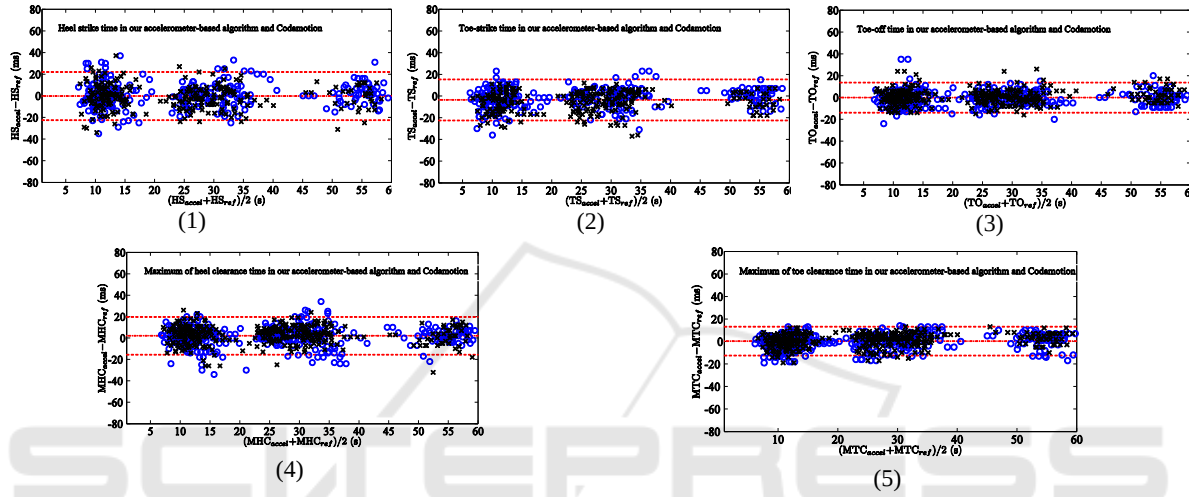
3 RESULTS

3.1 Validated Extraction of HS, TS, TO, MHC, and MTC in Comfortable Walking of Older Adults

Table 1 shows the stride-by-stride validation results of the extraction of the gait event timings, i.e., HS, TS, TO, MHC, and MTC in comfortable walking of older adults (mean walking speed = 1.324 m/s). The

Table 1: Stride-by-stride validation results of the five gait events detection in comfortable walking of older adults (mean walking speed = 1.324 m/s). These results are given as the accuracy (mean of the differences), the precision (std. dev. of the differences), limits of agreement, 95% confidence interval (CI) of the differences, and 95% CI of the lower and upper limits of agreement.

	Accuracy (ms) (precision (ms))	Limits of agreements (ms)	95% CI of the differences (ms)	95% CI of the lower limits (ms)	95% CI of the upper limits (ms)	No. of events
HS	0.8 (12.0)	[−22.7 24.3]	[−0.2 1.8]	[−24.5 − 21.0]	[22.5 26.0]	540
TS	−3.6 (9.6)	[−22.5 15.3]	[−4.4 − 2.8]	[−24.0 − 21.1]	[13.9 16.8]	517
TO	−0.1 (7.0)	[−13.9 13.6]	[−0.7 0.4]	[−14.8 − 12.9]	[12.7 14.6]	636
MHC	4.0 (8.8)	[−13.2 21.1]	[3.3 4.6]	[−14.3 − 12.1]	[20.0 22.3]	705
MTC	0.3 (6.5)	[−12.5 13.1]	[−0.2 0.8]	[−13.4 − 11.7]	[12.2 13.9]	681



(1) (2) (3) (4) (5)

Figure 3: Bland–Altman plot results of the extracted gait events, i.e., (1) HS, (2) TS, (3) TO, (4) MHC, and (5) MTC, measured using our method and the reference method, with mean (dash-dotted line in the middle) of differences $HS_{accel} - HS_{ref}$, $TS_{accel} - TS_{ref}$, $TO_{accel} - TO_{ref}$, $MHC_{accel} - MHC_{ref}$, and $MTC_{accel} - MTC_{ref}$. 95% of these differences are between the lines ± 1.96 std. dev. (dashed lines). (+) and (**o**) refer to gait events measured at the left foot and those measured at the right foot, respectively.

accuracy and precision of gait events detection ranged from −3.6 ms to 4.0 ms, and 6.5 ms to 12.0 ms, respectively. Given the sampling frequency of 200 Hz of the recorded heel and toe accelerations for both feet, the accuracy and the precision of detection are less than the durations of 1 sampling period (i.e., 5 ms) and 3 sampling periods (i.e., 15 ms), respectively.

Figure 3 shows the Bland-Altman plot results for the extracted gait events. These plots show small mean differences between the accelerometer-based algorithm extraction and the reference method in accordance with the accuracy of detection provided in Table 1. In addition, the limits of agreement (i.e., mean ± 1.96 std. dev.) and their associated 95% confidence interval (CI) exhibit small variations in the times of the gait events (Table 1).

3.2 Validated Extraction of the Gait Phase/Sub-Phase Durations in Comfortable Walking of Older Adults

Table 2 shows the results of the comparison between the values of the left/right gait phase/sub-phase durations obtained by our accelerometer-based algorithm and those obtained by the reference method. These phase/sub-phase durations could be estimated with a mean absolute error less than 11 ms. Bland–Altman plots show a mean difference between our method and the reference method of 0 ms (95% CI, −28 ms to 29 ms) for stride time, of 0 ms (95% CI, −28 ms to 27 ms) for stance time, of 0 ms (95% CI, −27 ms to 28 ms) for swing time, of 0 ms (95% CI, −28 ms to 27 ms) for double support duration, of −3 ms (95% CI, −31 ms to 24 ms) for HS2TS sub-phase duration, 5 ms (95% CI, −31 ms to 24 ms) for

Table 2: Values of left (L)/right (R) gait phase/sub-phase durations extracted by our method are compared to those extracted by a reference optoelectronic method, Codamotion, in comfortable walking of older adults (n = 38, 71.0 ± 4.1 years old). This comparison is given as the mean of differences (mean error) and mean of absolute differences (abs. error) between these values.

Gait phase/sub-phase durations	Side	Accelerometers data (ms)	Codamotion data (ms)	Mean error (ms)	Abs. error (ms)	No. of parameters
Stride time	L & R	1044 ± 78	1044 ± 79	0 ± 15	11 ± 10	373
Stance time	L & R	661 ± 58	662 ± 61	0 ± 14	11 ± 9	440
Swing time	L & R	388 ± 27	388 ± 28	0 ± 14	11 ± 9	497
Double support	L & R	136 ± 24	136 ± 26	0 ± 14	11 ± 9	437
HS2TS sub-phase	L & R	79 ± 11	82 ± 16	−3 ± 14	11 ± 10	410
TO2MHC sub-phase	L & R	53 ± 5	48 ± 10	5 ± 10	8 ± 7	607
MHC2MTC sub-phase	L & R	300 ± 22	303 ± 22	−4 ± 9	8 ± 7	582
MTC2HS sub-phase	L & R	36 ± 12	37 ± 16	0 ± 13	10 ± 8	518

TO2MHC sub-phase duration, −4 ms (95% CI, −23 ms to 14 ms) for MHC2MTC sub-duration, and of 0 ms (95% CI, −26 ms to 25 ms) for MTC2HS sub-phase duration (Figure 4).

4 DISCUSSION

We have presented in this paper an *ad hoc* algorithm for the extraction of the durations of (1) the left/right stride, stance and swing phases, and (2) the left/right sub-phases refining the left/right stance and swing phases during non-turning, overground walking episodes in older adults, from left/right sagittal heel and toe accelerations.

This algorithm takes advantage of existing remarkable features in the recorded accelerometers data to detect the gait events from relevant local acceleration signals. The validation of the extraction of the gait events and associated gait phase/sub-phase durations was carried out in comfortable walking of older adults (n=38). The experimental results show a good agreement between our algorithm and the reference method provided by a kinematic system (gold standard), and demonstrate an accurate and precise detection of HS, TS, TO, MHC, and MTC. In addition, our algorithm extracts the durations of associated gait phases/sub-phases with a good accuracy and precision.

Table 3 shows an overview of related work that reported an accuracy and precision of the extraction of gait phase durations in comfortable walking of older adults. Compared to stride, stance, and swing times calculated in (Rampp et al., 2015) (i.e., 2 ms ± 68 ms, 9 ms ± 69 ms, and −8 ms ± 45 ms, respectively), the accuracy and precision are improved in our method (i.e., 0 ms ± 15 ms, 0 ms ± 14 ms, and 0 ms ± 14 ms, respectively). Better accuracy and precision in stance and swing times are also found in our method compared to (Trojaniello et al., 2014) (i.e., 10 ms ± 19 ms and 9 ms ± 19 ms, respectively). Moreover, the accuracy and precision of the stride time in (Trojaniello et al., 2014) (i.e., 0 ms ± 14 ms) are similar to our results. The absolute error in the extraction of the stride time is also improved in our method (i.e., 11 ms ± 10 ms) compared to results reported in (Micó-Amigo et al., 2016) (i.e., 21 ms ± 12 ms).

The presented algorithm has the major advantage to quantify gait sub-phase durations that have a clear significance to clinical practitioners, since the estimation of these gait sub-phase durations is based on fundamental events of walking. Moreover, this algorithm allows a stride-by-stride extraction which may be relevant for the gait analysis of some specific population such as Parkinson's disease patients who experience freezing of gait, a sudden and brief episodic alteration of strides regulation. Moreover, the high precision achieved in the extraction of the gait phase/sub-phase durations promises excellent results in case of tracking the subtle decline/changing in these durations in older adults. This algorithm could be thus relevant for characterizing, e.g., the progression of a neurological disease, and for an early prediction of, e.g., elderly falls.

This algorithm used the cutoff frequencies of 7 Hz and 11 Hz for the MTCaccel extraction from recorded data at 200 Hz (Sec. 2.2.1); these cutoff frequencies should then be adapted in the case of lower/higher sample rate. In addition, we defined empirically the intervals $[TO_{accel}, t_{min} + 0.4*(HS_{2accel} - TO_{accel})]$ and $[TO_{accel} + 0.4*(HS_{2accel} - TO_{accel}), HS_{2accel}]$, and the times $TO_{accel} + 0.18*(TO - t_{cz})$ and $t_{cz} + 0.75*(t_{min} - t_{cz})$ for the detection of MHC_{accel} and MTC_{accel} in comfortable walking older adults (Sec. 2.2.1). These intervals and times would require further investigations in the case of slow and fast

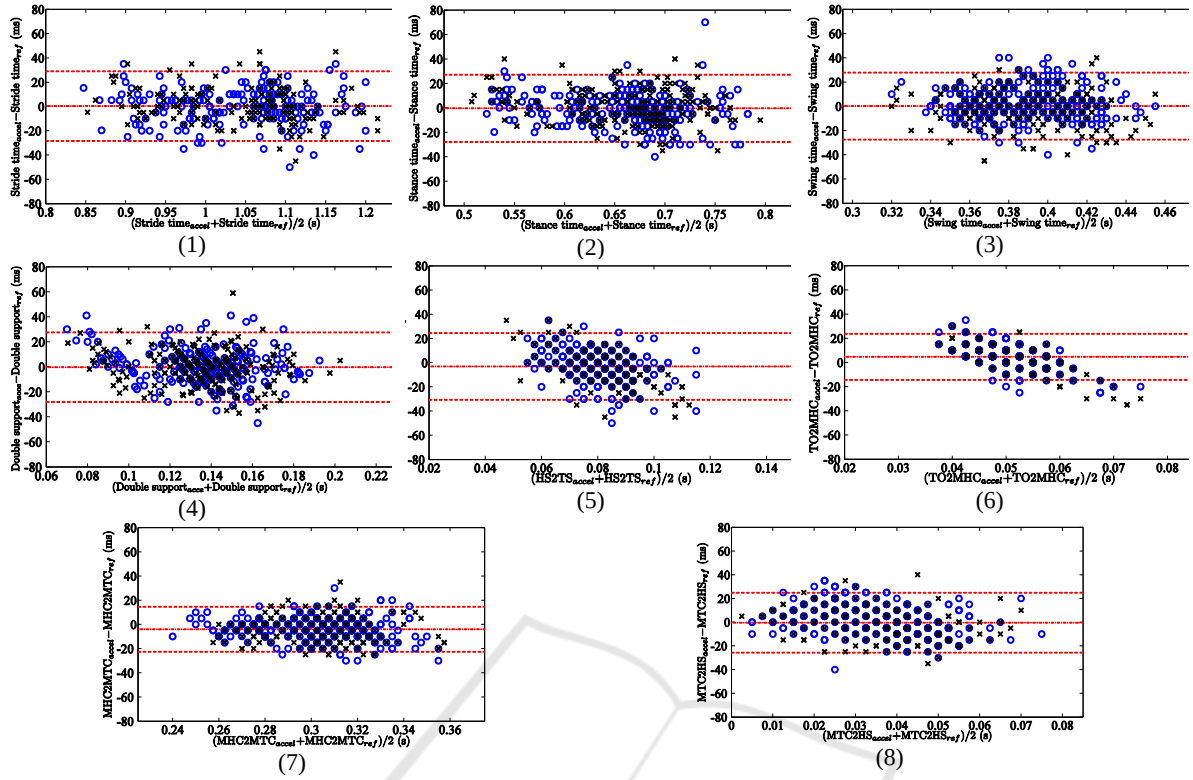


Figure 4: Bland–Altman plot results of durations of (1) stride phase, (2) stance phase, (3) stance phase, (4) double support phase, (5) HS2TS sub-phase, (6) TO2MHC sub-phase, (7) MHC2MTC sub-phase, and (8) MTC2HS sub-phase extracted during consecutive strides by our method and the gold standard method in comfortable walking of older adults. (+) and (o) refer to left and right time-related gait phases/sub-phases, respectively.

walking speeds and in the case of pathological gait patterns. Moreover, the algorithm is valid in case of a heel strike at initial contact during walking, but might be modified to be more flexible to take into account situations where the heel strike (or other events) is missing (e.g., in case of toe landing at initial contact) such as in running or in some pathological conditions.

5 CONCLUSION

We presented and validated on a stride-by-stride basis an *ad hoc* signal processing algorithm that extracts durations of (1) the left/right stride, stance, swing, and double support phases, and (2) the left/right sub-phases that refine the left/right stance and swing phases in comfortable walking of older adults (21 women and 17 men, 71.0 ± 4.1 years old), using an ambulatory foot-worn accelerometer system. The algorithm was tested against a reference kinematic system (used as gold standard) and yielded (1) an accuracy and precision that range from –3.6 ms to 4.0 ms, and 6.5 ms to 12.0 ms, respectively, for the extraction of left/right HS, TS, TO, MHC, and MTC, and (2) an accuracy and precision that range from –4 ms to 5 ms, and 9 ms to 15 ms, respectively, for the estimation of durations of left/right stride, stance, swing, and double support phases, and of left/right HS2TS, TO2MHC, MHC2MTC, and MTC2HS sub-phases.

To the best of our knowledge, this is the first study that demonstrates a good validation accuracy and precision in the extraction of sub-phase durations refining the stride phase duration during comfortable walking of older adults and using an ambulatory foot-worn accelerometer system.

In a future work, we plan to investigate (1) the effect of the walking speed on the extraction accuracy and precision of the aforementioned gait events and phase/sub-phase durations in older adults, (2) the capability of those gait phase/sub-phase durations to differentiate elderly fallers from elderly non-fallers using, e.g., classification models, (3) the application of the proposed algorithm to the study of pathological gait (e.g., gait of patients with Parkinson's disease), (4) the extension of this algorithm to deal with the

Table 3: Related work: accuracy and precision of the extraction of gait phase durations in older adults using inertial sensors.

	Subjects	Diagnose	Gait phase durations	Mean error (ms)	Abs. error (ms)
Micó-Amigo et al., 2016	20 elderly	healthy	Step/stride time	NA	21 ± 12
Rampp et al., 2015	101 elderly	geriatric	Stride time	2 ± 68	29 ± 62
			Stance time	9 ± 69	33 ± 61
			Swing time	–8 ± 45	25 ± 38
Trojaniello et al., 2014	10 elderly	healthy	Stride time	0 ± 14	10 ± 10
			Stance time	10 ± 19	22 ± 28
			Swing time	9 ± 19	22 ± 27

NA: not available.

turning walking episodes, and (5) the extraction of spatial gait parameters from the heel and toe accelerations, taking advantage from the proposed algorithm that enables splitting the gait cycle time into small time intervals and thus the drift from successive integration in these small intervals could be minimized. In this context, i.e., the extraction of spatial gait parameters from accelerometer data, we obtained promising preliminary results reported in (Boutaayamou et al., 2017b).

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