Project 04 – Temporal gait parameters estimation from accelerometry signals in elderly and hemiplegic subjects

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1. Introduction

In medical field, temporal gait parameters as well as spatial gait parameters and other global parameters characterizing a gait, can be used to assess the advancement degree of neuro and/or motor diseases, evaluate the effectiveness of a treatment, and in general to assess the degree of disability of a person. Efficient and correct estimation of these in an automatic manner starting from wearable magneto-inertial sensors acquired signals is an open issue, which is nowadays claiming for growing interest.

The scope of this work was the estimation of initial contact (IC) instants of the gait starting from anterior-posterior (AP) anatomical axis acceleration signals acquired from the trunk of elderly and hemiplegic subjects through the peak detection method proposed by Zijlstra et al. in [1]. A further statistical analysis of errors on IC instants estimates and on stride durations (SDs) consequently calculated has been performed for each of the two populations to assess the usability of the method. Also a new method with improved performance with respect to the peak detection method in terms of accuracy for initial contact instants estimation has been developed.

2. Analysis, study and design activities

Basically the peak detection method exploits both the a priori modelling of trunk navigation during walking motor activity as an inverted pendulum and the observation of acceleration signals acquired during walk. It is indeed simple to understand and verify that during a step cycle the forward acceleration of the trunk has an increase up to a peak after which the contact with the soil of the stepping foot, due to ground vincular reactions propagating up to the trunk, causes a sudden deceleration resulting in positive to negative zero-crossing (from here only these zero-crossings, and not those occurring passing from negative to positive, will be denoted as ZC) of the forward acceleration. Therefore, on ideally clean accelerometry signal from trunk of healthy subjects, the ICs could be detected on the forward acceleration as the last local maximum prior to a ZC.

As real-world acquisitions are affected by instrumentation noise and the gait of subjects reasonably may have some irregularity, even along a straight line path at regimen condition, Zijlstra et al. propose a 20 Hz low-pass (LP) filtering of signals in order to smooth small and sudden accelerations that may be sensed over the described pattern in order to ease the detection of the right peak. Moreover, as during the theoretically single ZC some oscillations may result in multiple ZCs, being in healthy subjects the average step-cycle period of nearly 0.5 s, a 2 Hz LP filtering is included in the method to smooth the signals from these undesired oscillations and more reliably detect only the wished step-cycle related ZC. Figure 1 shows the just described method applied on a portion of forward acceleration signal acquired from the elderly subject 6 of the provided dataset (see next paragraph). From estimated IC instants, under the assumption of right-left steps alternation, SDs are calculated as the time lapse between i-th IC and the i+2-th IC.

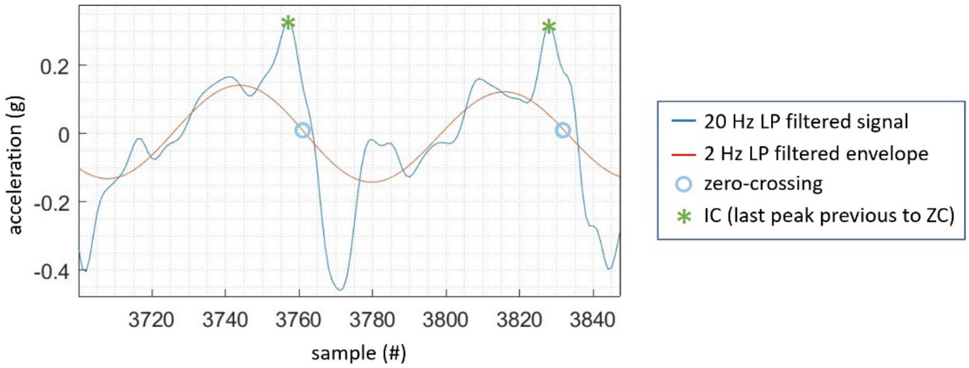

Figure 1. Peak method proposed in [1] for IC detection

The provided dataset has been acquired from 10 elderly subjects and 9 hemiplegic subjects walking for nearly one minute back and forth along a 12 m long path (for very impaired hemiplegic subjects -subjects 21 and 27- the acquisitions last longer to allow acquiring data from a statistically significant number of steps) at comfortable speed. The dataset contains for each subject both the signals acquired from a 3D accelerometer with a sampling frequency of 128 Hz placed at the pelvis level and the IC instants and SDs acquired from a 8 m long GaitRite instrumented walkway placed in the middle of the aforementioned path. The purpose of the latter was to give a gold standard (GS) to allow performance evaluation of the implemented method. Figure 2 shows the described experimental set up used during the acquisition sessions.

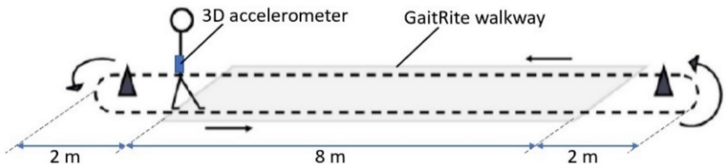

Figure 2. Data acquisition experimental set up.

As signals which we were dealing with were acquired not from healthy subjects, but rather from populations of elderly and hemiplegic subjects, moreover on walking bouts shorter than those studied in [1], therefore likely having a higher percentage of motor task's transient period, also a 0.1 Hz high-pass (HP) filtering has been inserted in the pre-processing pipeline of signals so that slow acceleration oscillations occurring during the activity could be flattened. This also permitted the reduction of residual errors committed by the acceleremoter due to eventual non-perfect calibration refinement or residual gravity acceleration. Notice that this step is not that important for the peaks detection as these remain easily detectable also on slow oscillations, but is essential to reduce the error committed in the ZC detections as the unwished slow oscillations or the offset could change, even relevantly, the point of intersection of the 2 Hz LP filtering extracted envelope with the zero level.

To assess the methods' performance in terms of accuracy, both on IC instants detection and SDs, first the matching between detected ICs and ICs given from the GaitRite GS shall be performed. To do this, as GaitRite data are collected only on 8 of the 12 meters of the path while accelerometers collect signals continuously during the trials, firstly all the estimated IC events not happening on the GaitRite walkway shall be discarded; after this the actual matching between GS detected events and remaining estimated events can be done. In this work it has been made the choice, starting from the GS IC instants, to search for each the matching estimated IC event in a neighbourhood of the fore defined by the two midpoints with its two adjacent GS ICs. If more than one estimated IC is found in a neighbourhood, only the closer is matched with the GS IC and the others are classified as extra events. On the other hand if no estimated ICs are detected in a neighbourhood, then a missed event is accounted. In this manner a discrimination on the entity and type of committed errors can be done, best understanding the critical points of the method. Yet, when a missed event occurs, all the SDs involved by this shall be discarded when calculating the errors, with particular attention to not compute stride durations as the time lapse between an IC occurring before the missed event and the IC of the contralateral foot occurring immediately after the missed event.

3. Development and prototyping activities

All the signal processing and statistical analysis described and mentioned in the previous sections have been performed in MATLAB's numeric computing environment.

The given starting point accelerometry signals were already expressed in the subjects' anatomical local reference system (actually signals re-orientation based on the alignment of forward direction with the direction of progression has not been performed, but in temporal parameters estimation context, having sensors properly orientated and fixed on the subjects during the acquisitions, this assumption can be done with few consequences on errors affecting the estimates). Starting from these, the above explained method for IC instants estimation has been implemented on the forward (AP axis) acceleration signals.

Estimated IC instants and computed SDs of each subject have been collected and matched with GaitRite detected ICs and SDs according to the procedure discussed in previous section, implemented in the provided 'IC_ON_GAITRITE' MATLAB function. After the matching, single errors on each estimated IC and SD have been calculated as $e = P_{estimated} - P_{GS}$, where P stands for the generic parameter. The committed errors have been then described both at subject level and at population level. At subject level the mean absolute error (MAE) has been calculated for both the parameters. At population level the grand mean, meant as average of the averages, of the MAE on IC instants and of percentage relative MAE on SDs has been calculated. In this context, calculating the grand mean as average of the averages instead than as the overall (sample) average on all the grouped single errors has the advantage of not incur in performance overestimation due to a higher number of events detected on less impaired subjects and a lower error on their estimates. Indeed it is likely that, in the same amount of time, less impaired subjects perform a greater number of steps than more impaired subjects.

3.1 New method proposal

The peak detection method proposed by Zijlstra et al. in [1] has been developed on gait accelerometry signals acquired from healthy subjects showing step level intra and inter-subject high repeatability. It is easily understandable that for signals acquired from impaired subjects, such as hemiplegics, the standard shape of forward acceleration signals and inter-subject step repeatability are lost. Other than that lots of additional, and usually unpatterned, trunk oscillations occur during the gait. For all these reason, while still working sufficiently well on signals from elderly subjects, the method showed poorer performance on signals from hemiplegics. In this sense the first point to fail is the detection of the ZCs, which does not properly works for two main reasons. First of all, the 2 Hz fixed cut-off frequency LP filtering aiming to extract the step level envelope of the signals fails in its purpose as for this population the step duration may vary, even of several seconds, from subject to subject. Usually being significantly higher than the average 0.5 s (see section 2) step duration of healthy subjects. Secondly, even assuming that the envelopes are extracted using a proper cut-off frequency, hemiplegic subjects mainly apport motion to their center of mass (CoM) with only one of their lower limbs while walking; the other limb, in the more optimistic of the situations, simply works as a support leg apporting only a very small amount of motion. Therefore the small envelopes related to the hemiplegic limb steps, inserted on higher envelopes related to the health limb, may not result in an actual ZC. Lastly, the sudden not standard oscillations of the trunk result in a multiplicity of forward acceleration peaks, among which the right one is not detectable as the last previous to the (erroneous) ZC. The just described situation is shown in Figure 3.

To overcome these three main problems a new method has been developed. The first problem is solved performing a spectral estimate of the acceleration signal and an harmonics analysis. Indeed, in case of hemiplegic population, the higher spectral peak usually coincides with the step related frequency if at least both the limbs apport some motion to the motor task, while it is likely that it is occurs at the stride related frequency if the subject's motion is highly asymmetric. In this case, as there is necessarily some not negligible amount of power carried by the step related frequency occurring nearly at a double frequency, the ratio between the peaks level of the higher peak and the peak level of its second harmonic is computed. If it is higher than 3.5 the frequency of the first peak is chosen as step related frequency, while if not the frequency of the second peak is chosen. The used value of 3.5 has been selected analysing the data of hemiplegic subjects in the provided dataset. The second and the third presented problems are soon bypassed, once extracted the step level envelope, segmenting the signal in step-like cycle segments (i.e. segments which actually are not steps -clearly, as this would be impossible- but that are nearly of the duration of the occurring step) individuated by the local minima of the envelope and for each segment choosing the instant of greatest contiguous deceleration pattern as IC (the instant is calculated as the midpoint between the instant at which the deceleration pattern begins and the subsequent instant in which the signal assumes a value lower than the envelope). Last but not least if the subject is impaired at such a level that practically all the amount of propulsion to the CoM is given only by one leg, while still being the IC peak well detectable on the signal for both the foot, the envelope related to the impaired limb may not produce detectable minima. Therefore a check on the step period resulting from the found ICs is performed and if it results that the period is 1.75 times greater than the average step period of the subject, calculated as the inverse of the estimated step related frequency, a correction mechanism searching for an additional greatest contiguous deceleration pattern in the segment is actuated. The fact that the method does not rely anymore on the ZCs makes it also much more robust to the choice of pre-processing stage parameters and therefore easy to use than both the methods proposed in [1]. An example of the functioning of the method is shown in Figure 4.

Both the cases in Fig. 3 and Fig. 4 refers to the same highly impaired hemiplegic subject (-ID 21 in provided dataset-) to more easily appreciate the differences between the two methos, but the very same considerations, even if with a lesser degree, can be taken as generally true.

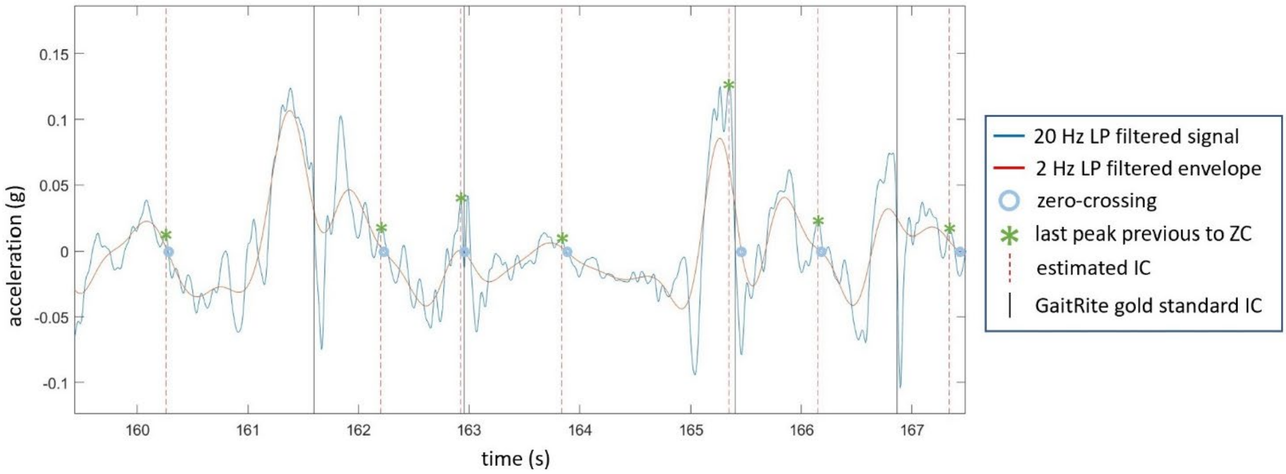

Figure 3. Estimating ICs according with the method proposed in [1] results in many extra events due to not reliable and inconsistent ZC detection. Moreover the ICs not classified as extra events generally result quite distant from the actual ones.

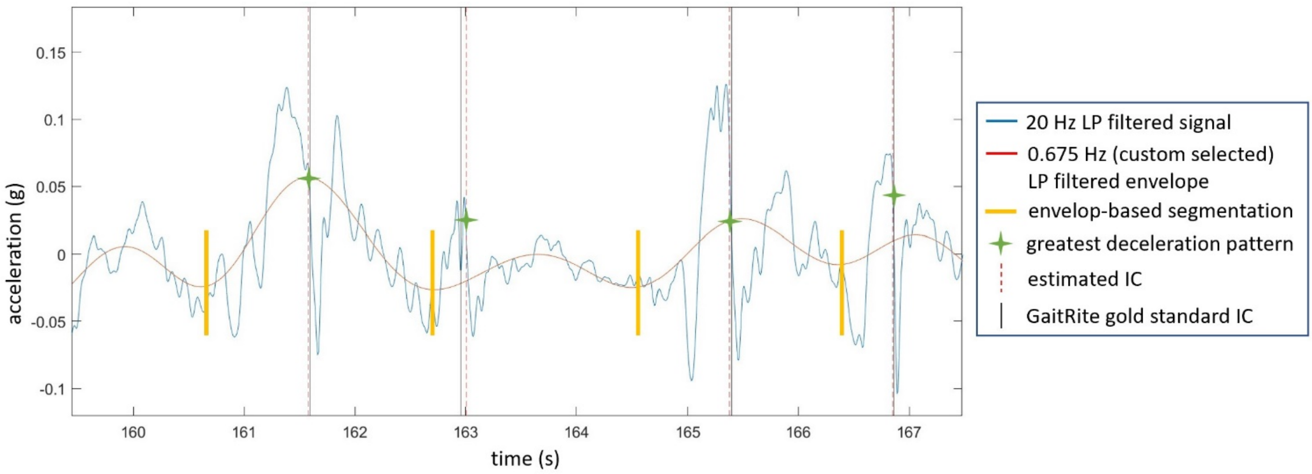

Figure 4. The same treat of signal of the one reported in Figure 3 is shown. It can be noticed that a proper signal segmentation allow to find the right number of ICs, significantly reducing the number of missed and extra events. Moreover, having to deal with these "noisy" human gait patterns, searching for the greatest deceleration pattern instead than using an ordinal criterium on the peaks results in better estimates of the correct time at which the IC occur.

4. Results and conclusions

Comparing the estimated temporal parameters with those returned by the GaitRite instrumented walkway gold standard, both the methods discussed in the previous sections have been assessed in terms of accuracy, as described in section 2, discriminating the accuracy in actual accuracy on matching events, missed events and extra events. As mentioned, at population level the grand mean has been computed as average of averages in order to avoid performance overrating.

Table 1 summarize performance both of the peak method and of the new proposed method on both the populations. The fore discretely performs on signals from elderly subjects group while shows poorer performance on signals from hemiplegics group, especially for what concerns the rate of missed events which is clearly a critical point for the application of the method on this population. In particular, as expected, the method does not work well on highly impaired subjects (21 and 27 of the available dataset). The new proposed method showed better performance, other than on the hemiplegics group, also on the group of elderly as it has been developed based on general features of the gait and not ad hoc on hemiplegic subjects' signals.

Table 1. Accuracy of the two discussed methods on Elderly and Hemiplegic groups.

Group	Peak detection method [1]				New proposed method			
	MAE on ICs (s)	% MAE on SDs	% missed events	% extra events	MAE on ICs (s)	% MAE on SDs	% missed events	% extra events
Elderly	0.0523	2.8075	0.6208	0	0.0116	0.9108	0.3217	0.4941
Hemiplegic	0.0884	3.5918	17.2076	1.7094	0.0312	1.8305	8.3730	2.7857

The new proposed method showed relevantly improved MAE on IC instants, percentage relative MAE on SDs and missed events rate at the cost of a very slight increase on extra events rate.

Moreover the new method is more robust for what concerns application to highly impaired subjects, thus being potentially usable even on these subjects. A more detailed insight on this aspect is given in Figure 5, in which the Bland-Altman plot of single errors committed on SDs estimates on all the subjects of each group is shown for both the methods.

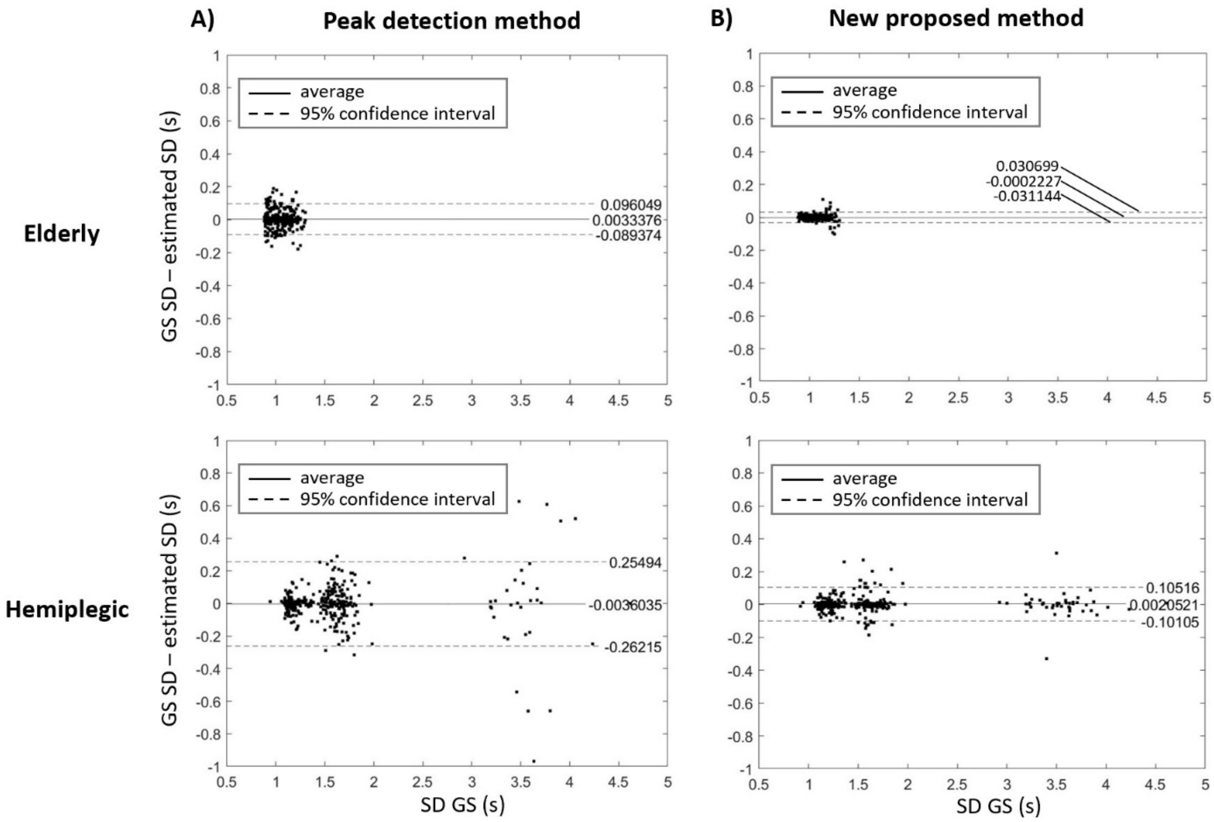

Figure 5. Bland-Altman plot of errors committed on SDs estimates from the peak detection method (A) and the new proposed method (B).

5. References

[1] Wiebren Zijlstra, At L. Hof. "Assessment of spatio-temporal gait parameters from trunk accelerations during human walking", Gait Posture 2003, 18:1-10.

6. Attachments

'Main_I_BUPA.m' contains the MATLAB code performing the estimation of IC instants and SDs calling for 'ZIJLSTRA_METHOD_BUPA.m' and 'icburc.m' functions, respectively implementing the peak detection method and the new method. 'Main_II_BUPA.m' contains the MATLAB code performing the computation and statistical description of committed errors.