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Comparison of Step Length Estimators from Weareable Accelerometer Devices

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***Abstract*— Wearable accelerometry provides easily portable systems that supply real-time data adequate for gait analysis. When they do not provide direct measurement of a spatio-temporal parameter of interest, such as step length, it has to be estimated with a mathematical model from indirect sensor measurements.**

In this work we are concerned with the accelerometry-based estimation of the step length in straight line human walking. We compare five step length estimators. Measurements were taken from a group of four adult men, adding up a total of 800m per individual of walking data. Also modifications to these estimators are proposed, based on biomechanical considerations. Results show that this modifications lead to improvements of interest over previous methods.

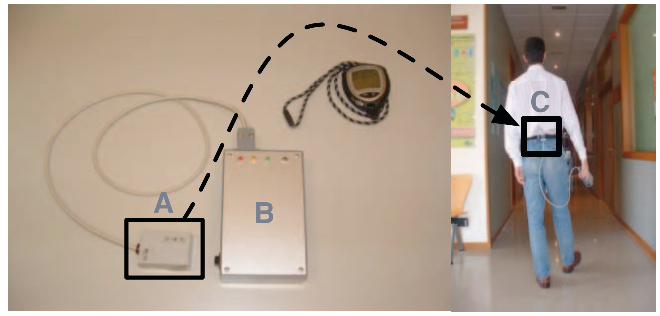

Fig. 1. Accelerometer device and data acquisition system. The triaxial accelerometer (A) is sampled at 50Hz with the data logger (B), and is fixed to the lower lumbar spine (C) with an adjustable corset (not shown in figure) to avoid movement artifacts.

I. INTRODUCTION

Gait analysis is frequently calculated by means of video-based recording of markers placed at end-points of all or a subset of body segments. However, this equipment is not easily portable, it requires off-line digitizing that can be time consuming and automated real-time systems which can be costly. This makes these systems impractical for use by the average clinician in daily practice. Other measurement devices such as force-plates, electro-goniometres or electrodes to measure EMG signals [1] are not designed for usability.

Conversely, accelerometry provides easily portable systems that supply real-time data. In addition, these systems come at a decreased cost when compared to video motion analysis systems, making them easily available to a wide range of clinics. Precision and repetibility characteristics of accelerometry make it adequate for gait analysis [2], and it has been widely used recently, mainly in ambulatory diagnostic [3], [4], [5], [6], [7].

However, such systems do not provide direct measure of several spatio-temporal parameters of interest such as step length, walking distance or walking velocity. Instead, they have to be estimated with a mathematical model from indirect sensor measurements. Specifically, in this work we are concerned with the accelerometry-based estimation of the step length in straight line human walking.

For instance, there exist empirical relations between the step length and the maximum and minimum vertical acceleration of the body's center of mass (COM) [8]. More frequently it is accepted a linear relation of the step length with the step frequency [9]. Finally, clinical studies show that an inverted pendulum model can describe the displacement of the body's COM while walking, and the model can be applied to estimate the step length from this displacement [10], [11]. All these estimators had in common that they require a previous parametric adjustment or calibration from experimental walking data for every individual. After that, they all provide comparable estimation results, as will be shown in this paper.

In the following we compare these common step length estimators reported in the literature. Also modifications to these estimators are proposed, based on biomechanical considerations. Results show that this modifications lead to improvements of interest over previous methods.

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II. METHODS

A. Subjects

Measurements were taken from a group of four adult men, ages 32 to 38. Height of subjects varied from 168 to 186 cm. None of them presented vestibular or neurological disorders that could affect the experiments. All of them gave their signed consent and the experiments procedures agreed with the Helsinki Declaration.

B. Equipment and data acquisition

The step length will be estimated from a three-axial accelerometer placed close to the L3 vertebral position, accepted as a fine approximation of the COM position during normal walking [4]. The device is fixed to the lower lumbar spine with an adjustable corset to avoid movement artifacts.

The accelerometer prototype (see Figure 1) is built over two biaxial accelerometers AXL202AE from Analog Devices [8], mounted to form a 3-axes frame, with a measuring

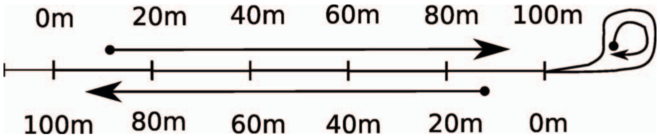

Fig. 2. Experimental setup: individuals walk a 100m distance at different paces, back and forth.

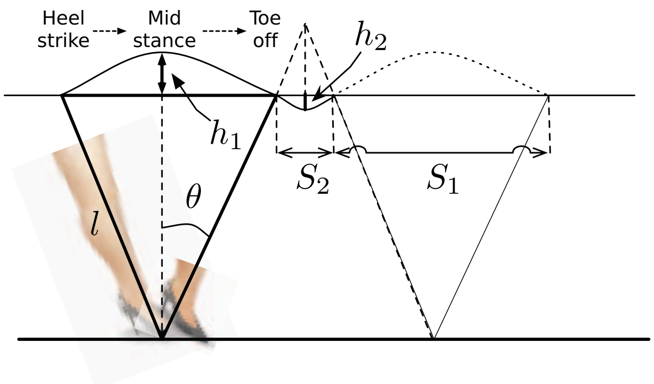

Fig. 3. Inverted pendulum model of human walking. The step length S can be estimated as the addition of two factors: the horizontal distance traversed by the COM during the mid-stance phase, S_1, and the distance traversed during the double-stance phase S_2. Both can be estimated from the vertical maximum displacements of the COM, h_1 and h_2 at each respective phase.

range of $\pm 2g$, g the gravity acceleration. A calibration procedure takes into account offset and scale factors and the orthogonality of the axes [12]. A PIC 16F877 Microchip micro-controller has been used as interface and data logger. Signals are sampled at 50Hz using a 10 bit A/D conversion. Data gathered during the experiments are stored in an internal memory and transferred to an external computer through a serial communication link for further processing. A total of 64Ksamples can be stored in memory, so we were able to extend the experiments over 20 minutes.

C. Procedures

During test procedures, subjects were asked to walk along a 100 meters long corridor following a straight path, back and forth, see Figure 2. First ten meters of walking were discarded for the analysis, as gait is not stable during initial phases of displacement. Each individual completed four independent excursions, and they were asked to maintain a constant velocity for each walk: "preferred" (first excursion), "fast" (second excursion), "low" (third excursion) and "medium" (a velocity between preferred and fast velocities, last excursion). This adds up a total of 800m per individual. Subjects were allowed to turn freely between both 100m segments, and to walk freely or to rest on a chair between different excursions for a lapse of five minutes. Time laps were taken at 20m intervals with a stopwatch. Re-test procedures were carried out similarly one month later, with the same four subjects and procedure.

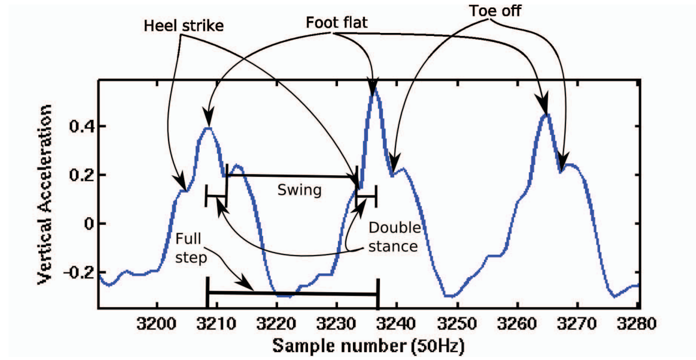

Fig. 4. The modified estimator ($M3$) breaks each step in two phases: double-stance and single-stance. Heel-strike and Toe-off events are detected in the measured accelerations. An inverted pendulum model will estimate the longitudinal displacement during the single-stance phase. The displacement during the double-stance phase will be assumed constant.

III. STEP LENGTH ESTIMATION

In its simpler form, human gait can be described by an inverted pendulum model, see [10]. From this mechanical model, necessary relations between the forward displacement S and various measurable step variables can be obtained. Here we will use the relationship between the vertical and the forward displacement, given by equation:

$$S_{M_1} = K2\sqrt{2lh - h^2} \tag{1}$$

where l stands for the leg length, and h stands for the vertical displacement of the center of mass during one step, as in Figure 3. K is a constant which has to be calibrated for each individual based on experimental data. The calibration procedure is as follows: for each individual K is the ratio between the real and the estimated walked distance for a given trial.

The vertical displacement for each step, h, is computed with a double integration of the vertical acceleration of the body center of mass. To avoid the integration drift error we need to reset the integral at any point of the step. This reset point was settled at the time of foot-flat, where the vertical velocity of the body is null. Notice that the forward acceleration can not be used because the COM forward motion has no instants of null velocity and consequently the double integration drift error grows unbounded. Mediolateral acceleration (which is responsible of the lateral movement of the hip) presents also zero velocity points, but this signal is weaker than the vertical acceleration, and therefore more sensitive to noise. In the following we will refer to expression (1) as M_1 estimator.

Estimator M_2 assumes that in the time of foot-flat, the velocity is zero and the vertical coordinate of the COM is the same that at the beginning of the step. Adding a constant offset term to the acceleration, and forcing the final values of the integrals to be the desired ones in equations (2) and (3):

$$\int_{t_0}^{t_1} (a_y + c_1)dt = 0 \tag{2}$$

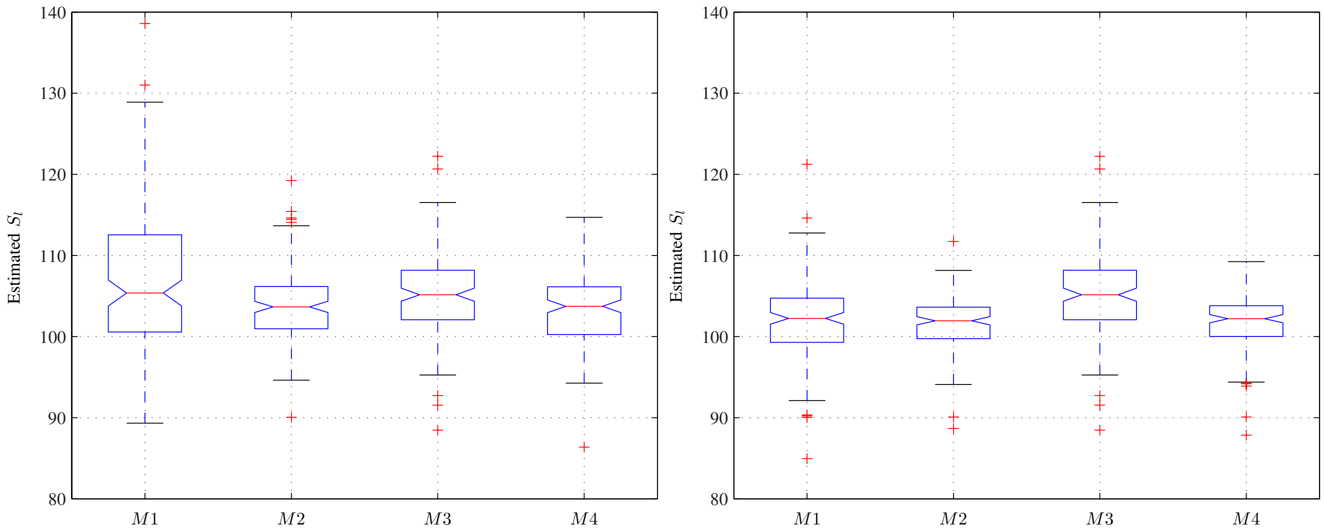

Fig. 5. Walking distance estimation, using methods $M1$ to $M4$, compared to the real walking distance. For each estimator the distribution (median, quartiles and extreme values) is represented. (Left) Calibration made using data from the experiment at preferred speed. (Right) Calibration made using data form experiments at different speeds. In both cases, the method $M3$ do not need calibration.

$$\int_{t_0}^{t_1}\left(\int_{t_0}^{t}(a_y+c_1)dt\right)+c_2dt=0 \tag{3}$$

being t_0 and t_1 two consecutive instants of foot-flat, and c_1 and c_2 two offsets. This methods reduces the drift effects, and a more exact computation of the COM excursion can be done. This method needs the same calibration as method M_1.

Estimator M_3 assumes a more complex model, as shown in Figure 3. In this model, the displacement of the COM is related with two pendulums. An inverted pendulum with length equal to the length of the leg during the swing phase, and a second pendulum (of unknown radius) during the double-stance phase.

$$S_{M_3}=S_1+S_2=2\sqrt{2lh_1-h_1^2}+S_2$$

Total displacement, S_{M_3}, is computed as the displacement during the swing, S_1, plus the displacement during the double stance, S_2. The displacement S_1 can be computed according to the same equation (1) employed by model M_1. The vertical excursion of the COM, h_1 is computed by means of the double integral of the accelerations between the time of toe-off and the time of heel-strike. Detection of toe-off and heel-strike events is done using both the vertical and the antero-posterior accelerations. The drift correction method employed in M_2 is also employed in this algorithm, so that vertical accelerations must also be double integrated between consecutive foot-flat instants in order to avoid the drift. Displacement S_2 is set as a constant equal to the foot length, from the first metatarsal head to the calcaneal tuberosity. This method does not need any experimental calibration.

Instead of the double integral, there is an empirical relation of the vertical acceleration with the stride length that depends on the step count n:

$$S_{M_4}=Kn\sqrt[4]{a_M-a_m} \tag{4}$$

being a_M and a_m maximum and minimum values of the first harmonic of the vertical accelerations at every step, and K a calibration constant that has to be computed from experimental data in the same way as in methods M_1 and M_2. The first harmonic of the acceleration is computed by means of a low pass filter set at 3Hz. This estimator, M_4, is claimed to measure distance walked to within 8 per cent across a variety of subjects of different leg lengths [13].

The fifth method, M_5, is based on the linear relation existing between the cadence and the step length. From experimental data at different speeds, a mean value of cadence and step length is calculated from the first 40m. Those data are fitted to a straight line for each individual. Using this linear model, an estimation of each single step is computed by mean of a linear regresion:

$$S_{M_5}=\frac{K_1}{T_{step}}+K_2$$

being T_{step} the period of the step, K_1 and K_2 two calibration constants.

IV. RESULTS

In Figure 5 there is a comparison among the first four estimators, M_1 to M_4. The vertical axis is the ratio between the estimated and the real walking distance. For each estimator the most extreme values in the data set (maximum and minimum values), the lower and upper quartiles, and the median are plotted. The estimation has been done for each section of 20m, and the ratio between the estimation and the real ground value (20m) has been computed. The figures show aggregated data for all trials and all individuals. Two different calibration procedures has been employed. In the first one, Figure 5-left, K was computed using the data from the first 40m walked at the preferred speed. In the second one, Figure 5-right, K was computed using the data from the

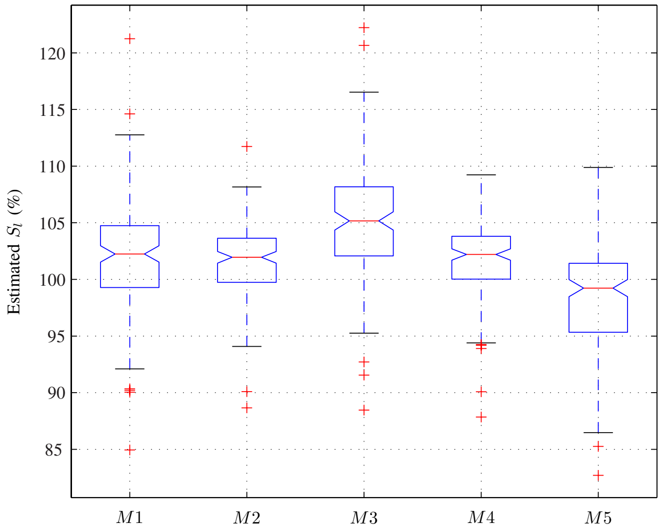

Fig. 6. Estimation based on cadence requires data gathered at different velocities for calibration. The figure represents walking distance adding the new estimator ($M5$) compared to the real walking distance. The new calibration data set is used with all estimators (except $M3$).

first 40m at each velocity. In both cases, method M_3 was not experimentally calibrated. Results were computed using the anatomical data of each individual (for methods M_1 to M_3). In Figure 6, all estimators are compared. Every method was calibrated using the data for the first 40m at each speed.

Methods M_1 to M_4 show a tendency to over-estimate the step length, as their median values vary between 100% and 105% of the real value. The worst case error was obtained with method M_1 using single speed calibration, and was almost a 40% higher than the real value. Method M_1 also has the highest variance. When data from multiple speeds are used, all the methods improve, both in median and variance. Method M_5 has the lower median of all (98%).

V. DISCUSSION

In this study five different methods of estimation of the step length were compared. Common methods reported in the literature, M_1, M_2 and M_4, present similar performance. According to the results of our study, there are no significative difference, on average, among the three methods (see Figure 5. On the other hand, method M_1 has more variability for a given prediction. There is no statistical difference between methods M_2 and M_4. This results are consistent with the fact that methods M_2 and M_4 are intended to reduce the effect of the integration drift. Method M_4 has the advantage that it is easier to implement.

Multiple velocity calibration improves the performance of methods M_1, M_2 and M_4, mainly through a significative reduction in the standar deviation (see Figure 5). Median values are closer to 100% (unbiased estimation), compared to single velocity calibration, although the difference is not statistically significative. This result is consistent with the fact that walking speed is a relevant factor of the gait.

Method M_5 produces a closer estimation to the desired value, although it also presents a high variance (see Figure 6). A possible reason for this is the low temporal resolution in the measurement of the step duration. Further experiments using a higher acquisition rate are needed to verify this hypothesis.

Method M_3, without experimental calibration, achieved similar results to methods M_2 and M_4 calibrated with a single velocity procedure (see Figure 5). This has application for wearable devices because this method avoids unfriendly and time consuming calibration procedures. Further research is required to define the optimal anatomical landmarks to use as pendulum length in the swing phase and displacement constant during the double stance period.

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