# A Novel Kinematic Model for Wearable Gait Analysis

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***Abstract*— The work studies the acceleration of a single tri-axial accelerometer fixed at the sacrum position on the back of subject next to the COM (Center of Mass). The correlation between the cycle of the COM and the cycle of the walking is analyzed by using a harmonic oscillator as a model for human locomotion. The COM position is calculated without any integration of its acceleration. A double integration of raw accelerometer data can result in an accumulation of drift error resulting in wrong position and distance or step length evaluation. The acceleration of a harmonic oscillator is directly proportional to the position. We evaluate the COM position and translation into a sinusoidal pattern. For every step cycle, the maximum of COM amplitude is used to give the relative S step length. This kinematic model generates the properly attended values; all steps are detected and the absolute accuracy error in the measurement of the step length, ranges from 0.32% to 3.33% with a mean value 2.17%. In the model, many output parameters are processed to study the subject movement analysis, but all that parameters should be compared with the gold standard values using appropriate protocols. We use data of Swedish adult people to obtain coefficients C to evaluate $S_{mean}$ (mean anthropometric step length). $S_{mean}$ is used only as a reference, but is not the S value measured by the model. New protocols and data verification are carried out. The expectation is to develop a dedicated tool to support diagnosis and rehabilitation.**

***Keywords-wearable device; wearable sensors; gait analysis; human kinematics; clinical application; algorithms.***

## I. INTRODUCTION

Gait analysis is a complex and expensive technology. The setup of a limited working area in laboratory, the use of markers on the subject and the data analysis are not simple to approach. Outside the laboratory, the use of the system is not artless. Healthcare requires a novel ecologic approach to the movement analysis in order to make it friendly and designed under ergonomic constraints as well as the performance assessment in agonistic sports and/or clinical follow up in home monitoring. The wearable sensors are a possible solution to this problem. They are easy to use and not intrusive, so that it is possible to monitor subjects everywhere [1][2]. The introduction of smart fabrics and wearable sensors improves and simplifies the development of these sensors, the evaluation of the movements and also the rehabilitation of patients in their clinical pathway. In fact, the possibility to embed sensors directly into the user's garments becomes real [3][4]. In this way, their use allows for a natural walk while monitoring is in progress. Clinical tests are often conducted with manual counting of times, steps and distances; their confirmation is carried out through the support of concurrent video analysis. The use of wearable devices for the gait analysis without optoelectronic analysis is possible. This matter is still under investigation. The need to monitor health status of patients drives an improvement and an evolution to use remote control systems for analyzing data through a trained medical center. Transmission and storage of clinical data is driven by attention to the security and privacy [1][5]. On the research side, more accurate biomechanical models are being implemented, the improvement of signal processing and advanced analysis algorithms are a focus of development so to enhance the interpretation of the output data of wearable sensors for decision making [2][6][7]. Every new wearable analysis tool for gait analysis is a strong target; better results need to be introduced into the clinical practice in order to exploit for example a wearable 6MWT (Six Minutes Walking Test), or a wearable TUG (Timed Up and Go), or other trials, without the particular limitations of a laboratory. Our work goes into this direction: we developed a novel kinematic model and related method to process 3D inertial accelerometer data and to compute parameters of gait analysis. The structure of the paper is as follows: section 2.1 the Biomechanical model; section 2.2 the experimental setup; sections 2.3 and 2.4 the Methods A and B to analyze raw data; sections 3.1 and 3.2 the Data Analysis with Method A and B; section 4 the conclusions.

## II. MATERIALS AND METHODS

This section reports the description of the methodology used for the kinematic model. We divided the section in four subsections in which we explain the Biomechanical model, the experimental setup, the processing methods for bot the parts.

### *2.1 Biomechanical Model*

Describing the movement of a subject, we have to consider the external forces and, therefore, the accelerations acting on his/her body. In absence of other forces, we always have the action of the gravity force, and then the subject has to produce a counterbalanced force to remain in

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equilibrium even when stationary. The COM (Center of Mass) is a single point where we can think that the whole mass of the body is concentrated, so that it is equivalent to the entire considered object, where the external forces act according to the Newton's laws of motion. In a standing posture, its position is typically about 10 cm lower than the navel, in the sagittal plane and in correspondence of the anterior superior iliac crests (the top of the hip bones). To know as the COM moves, it means to know how the object moves. For this reason, we studied human walking by evaluating the acceleration signals of a single tri-axial accelerometer fixed on the pelvis of subjects next to the COM, i.e. in correspondence of the second sacral vertebra.
In the FIGURE 1 a subject walk and the positions in time of the COM are presented with circles and a trajectory. L is the leg length, θ is the hip angle in the sagittal plane and S is the step length. While the subject moves a step S, the COM moves vertically along an oscillating path; the maximum oscillation amplitude is hCOM. When the subject moves the next steps, the cycle repeats. While walking, the COM oscillation pattern is considered sinusoidal in the vertical and mediolateral directions [8]. In this work we describe the COM vertical oscillation, but the same method is applicable to the COM mediolateral oscillation. To describe human locomotion, it is analyzed the correlation between the cycle of COM and the cycle of walking, by using a harmonic oscillator model. The legs are considered rigid bodies. The swinging movement is described by a pendulum model [9].

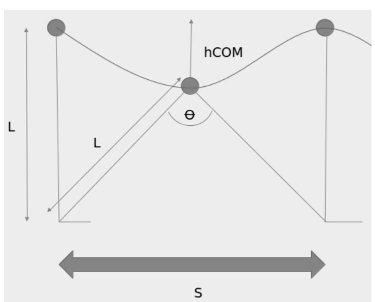


Figure 1. Representation of the Center of Mass oscillation path during walk

According to this model, we can define:

- S = the length of the step;
- L = the length of a lower limb;
- hCOM = the maximum amplitude of the vertical variation of COM trajectory (distance between the maximum and the minimum height of the COM);
- θ = the hip angle in the sagittal plane;

$$S = 2 * \sqrt{2 * L * hCOM - hCOM^2} \cdot$$

The COM position must be calculated. A direct double integration of raw accelerometer data gives the position which can result in an accumulation of drift error giving wrong position and wrong distance or wrong step length. Other studies chose to double integrate and report solution to solve the drift error [6]. We do not carry out a double integration. Some information may be obtained using the property that the acceleration of a harmonic oscillator is directly proportional to the position. For every step cycle, the maximum of COM amplitude is used to give the relative S step length and distance.

*2.2 Experimental setup*

We used a wearable 3D accelerometer (Protheo SXT s.r.l., Lecco, Italy) [10]. Protheo technical details are: 85 (l) x 53 (w) x 16 (h) mm of size, 70 g of weight, 4 digits LCD, on board ARM7 microprocessor and raw acceleration sampling frequency of 128 Hz. This device is able to log tri-axial accelerometer signals into the internal memory (up to three days of continuous monitoring) which can be downloaded at the end of the acquisition by Bluetooth® data transmission. The data storage allows for recording different tests in sequence. When we carry out different tests in sequence, we have no way to monitor over the data recording until the recording is downloaded. We calibrated the output raw signals when the data processing is carry out. Protheo could record also an electrocardiogram of the subject while he is making the test, but this option is not used for this study. The system is worn at sacrum position on the back of the subject, by means of an elastic band with a pocket for fixing the device firmly to the body. We recorded six independent walking tests of a single subject. In the test protocol the subject walked with shoes at self-selected speed over a linear path of 31.2 m; the walking time is not a constant. To control the test, the step length was kept fixed at 60 cm, so that 52 steps were necessary to complete the path. These values are the true imposed values used for accuracy assessment. For this purpose, a linear set of 60 cm interspaced lines was drawn to drive the position of the tip of the foot at each step. This step value is very close to the subject natural one that has the anthropometric estimated value of $S_{mean\ =}$ 61.9 cm (mean anthropometric step length). We evaluate $S_{mean}$ using data of Swedish adult people but different values are possible compared with different cultural background [11][12]. From this data we obtain the C coefficients depending on gender, speed and H (height) of the subject as in TABLE I.

$$S_{mean} = C * H$$

$S_{mean}$ is a reference and S the measured step length. C permits the evaluation of the mean anthropometric step length depending on gender, heigth and speed. Gender, age, weight and following anthropometric measurements of the subject under analysis were taken to complete the biomechanical model: a) lower limb (ground-greater trochanter); b) ground-malleolus; c) lateral condyle-greater trochanter; d) malleolus-lateral condyle; e) fifth metatarsal-malleolus; f) width of the foot; g) length of the foot to the ground; h) outer distance between the feet. The subject is a healthy male (w = 80 kg, h = 181 cm, age = 51 years) with normal BMI (Body Mass Index) 24.4. TABLE II reports his anthropometric measures. In the model, we use the Tanaka formula [13] to compute the HRmax (maximum Heart Rate) in bpm unit and the HRR (Heart Rate Reserve); we need the heartbeat as an input value at the beginning of the test with the resting subject and at the end of the test to compute the cardiac effort during exercise. The energy expenditure is evaluated in [METS]. Raw accelerometer data processing was implemented in Matlab© software suite. Two different processing methods were compared (Method A and B).

TABLE I. THE C COEFFICIENT CALCULATED USING SWEDISH REFERENCE DATA FOR NORMAL SUBJECTS;

| Speed | Low < 0.90 m/s | Normal [0.90-1.40] m/s | Fast >1.40 m/s |
|---|---|---|---|
| Man | 0.2928 | 0.3422 | 0.3956 |
| Woman | 0.3102 | 0.3539 | 0.3994 |

TABLE II. ANTHROPOMETRIC MEASURES OF THE SUBJECT UNDER ANALYSIS.

| Body Segment | Value [cm] | Body Segment | Value [cm] |
|---|---|---|---|
| a | 96 | e | 14.5 |
| b | 12 | f | 9 |
| c | 41 | g | 30 |
| d | 43 | h | 32 |

### *2.3. The Processing Method A*

The peaks in raw data were detected by peakdet.m function [14], using a 5th order pass-band Butterworth filter (band: 0.5 - 4 Hz). To identify S, we used the peak positions detected by the previous peakdet routine, applying the harmonic oscillator model to the original raw signals filtered with a low-pass 19th order Butterworth filter, with these cut-off frequencies for each acceleration:

- Antero Posterior acceleration: 6 Hz;
- Vertical acceleration: 7 Hz;
- Medio Lateral acceleration: 8 Hz.

The evaluation of hCOM was carried out by applying a cut-off threshold of 6 cm to the double of maxima amplitude of COM sinusoidal pattern. The amplitude to be considered in the model is the length between the maximum and the minimum for every COM oscillation. The time between two consecutive vertical peaks is the single step time. In order to identify the starting step (left or right), the analysis of mediolateral acceleration was made. The same analysis can be applied in order to extract the asymmetry of right and left steps [15]. The kinematic model can extract the following parameters:

- BMI [kg/m^2];
- cadence [step/min] and frequency [step/s];
- stepping time [s];
- stride's periods [s];
- right and left step's periods [s];
- displacement of COM [m];
- speed [m/s];
- speed for the right and left steps [m/s];
- right and left step lengths [m];
- length of steps [m];
- incremental distance traveled at each step [m];
- pace distances [m];
- pace angles [degrees];
- pace coefficients of walking efficiency;
- number of steps and strides;
- sagittal hip angle [degrees];
- the first right or left leg support;
- base of support both with aids that without [m^2];
- width of steps [m];
- acceleration peaks at ground support phase [g];
- step and stride indices of regularity and symmetry;
- total and incremental energy expenditure [METS];
- the power spectrum of the accelerometer signal;
- maximum heart rate and heart rate reserve;
- 6MWT predicted normal distance value [m];
- report with office format.

### *2.4. The Processing Method B*

Method A is consistent if subject walks with a constant step, but this is not a normal constrain; if there is a velocity variation for external causes or other voluntary choices, a new set of processing filters has to be applied to the acquired data for the research of peak acceleration and the relative values. To verify this hypothesis, we carried out a set of experiments by asking the subject to walk on a treadmill and progressively increasing its speed (from 0.5 to 1.7 m/s). The same concept could be applied to pathological patients, walking slower and asymmetrically, such as stroke patients (walking speed < 0.5 m/s).

This issue was faced in the second approach here proposed (Method B), as an evolution of Method A.

The peaks of raw signals that identify steps, were detected by peakdet.m function; the low-pass filter used is different according to the walking speed:

- at high velocity, a 5th order low-pass Butterworth filter is applied with the following cut-off frequencies:
  - Antero Posterior acceleration: 1.8 Hz;
  - Vertical acceleration: 1.8 Hz;
  - Medio Lateral acceleration: 0.9 Hz;
- at low velocity, a 4th order low-pass Butterworth filter:
  - Antero Posterior acceleration: 35 Hz;
  - Vertical acceleration: 5 Hz;
  - Medio Lateral acceleration: 3 Hz.

To identify S, we used the previous peak identification, applying the harmonic oscillator model to the original raw signals filtered by a 19th order Butterworth filter with the following cut-off frequencies:

  - Antero Posterior acceleration: 6 Hz;
  - Vertical acceleration: 35 Hz;
  - Medio Lateral acceleration: 8 Hz.

We used six different approaches to define hCOM threshold:

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Mode 0: if we do not know S, the length of the expected step, hCOM threshold is estimated by the median of twice the absolute value of the COM amplitude trend; this value is multiplied by weights that depend on the average walking speed;

Mode 1: if we know the expected step S, we impose the hCOM threshold equal to the expected hCOM

$$hCOM_{\text{expected}} = L - \sqrt{L^2 - (S/2)^2}$$

Mode 2: as in mode 1, but the expected hCOM threshold is increased of the 20%;

Mode 3: the hCOM threshold is the product of hCOM value evaluated by linear interpolation and a set of weight correction factors depending on speeds, so to matching the value of COM displacement measured by Orendurff [8], with the expected COM displacement of the model;

Mode 4: the hCOM threshold is the Lulic's COM amplitude [16] with the weights of Mode 3;

Mode 5: the hCOM threshold is 0.6 m.

Through the proper choice of the mode according to subject's feature in his/her different scenarios, the model calculates the correct length values and the number of strides and steps.

## III. RESULTS

The step counting obtained from the device was compared with the true reference value i.e. that one taken manually by the observing operator.

### *3.1 The Data Analysis with Method A*

The Method A (TABLE III) detected all 52 steps in the test (1, 2, 3, 5). In the tests (4, 6), 51 steps are detected. The absolute accuracy error in the measurement of the step length, ranges from 0.37% to 10.78% with a mean value 4.94%; the absolute accuracy error in the measurement of the walking total path, ranges from 0.37% to 10.78% with a mean value 5.22%. The subject covers the path from a standing start; the first steps are not regular because the subject is still not in a steady state. If we replace the initial steps with the next ones, we have a steady state for all the path. If the high frequencies contribute in the oscillator modeling, the removal of the low-pass filter during the step evaluation, should give a better result. The analysis with the Modified Method A is presented in TABLE IV. The absolute accuracy error in the measurement of the step length, ranges from 0.32% to 3.33% with a mean value 2.17%. The error is less that in the original Method A. The anthropometric distance is evaluated with C coefficient of TABLE I. With the Method A (TABLE V), the absolute accuracy error in the walking speed ranges from 0.37% to 10.77% with a mean value 5.55%. The stride and step frequencies measured are presented in the same table. With the modified method, A (TABLE VI), the absolute accuracy error in the walking speed ranges from 0.52% to 3.31% with a mean value 1.70%.

TABLE III. METHOD A: REAL AND CALCULATED STEPS AND DISTANCES VALUES FOR THE SIX TESTS. § ONE STEP IS MISSED.

| | Steps | Step Length [cm] | Step Length Accuracy Error % | | Distance [m] | Distance Length Accuracy Error % | |
|---|---|---|---|---|---|---|---|
| Test | Real = 52 | Real = 60 | Relative | Absolute | Real = 31.20 | Rel. | Abs. |
| 1 | 52 | 59.8 ± 7.6 | -0.37% | 0.37% | 31.08 | -0.37 | 0.37 |
| 2 | 52 | 56.3 ± 8.4 | -6.15% | 6.15% | 29.28 | -6.15 | 6.15 |
| 3 | 52 | 57.8 ± 8.2 | -3.64% | 3.64% | 30.07 | -3.64 | 3.64 |
| 4 | 51 | 57.6 ± 6.9 | -4.06% | 4.06% | 29.36 § | -5.91 | 5.91 |
| 5 | 52 | 53.5 ± 8.3 | -10.78% | 10.78% | 27.84 | -10.8 | 10.8 |
| 6 | 51 | 57.2 ± 8.1 | -4.65% | 4.65% | 29.18 § | -6.48 | 6.48 |
| mean | 52 | 57.0 | -4.94% | 4.94% | 29,50 | -5,22 | 5,22 |
| min | 51 | 53.5 | -10.78% | 0.37% | 27,84 | -10,78 | 0,37 |
| max | 52 | 59.8 | -0.37% | 10.78% | 31,08 | -0,37 | 10,78 |

TABLE IV. MODIFIED METHOD A. THE DATA ARE REAL AND CALCULATED STEPS AND DISTANCES VALUES FOR THE SIX TESTS. FOR * THE DISTANCE BY MODEL IS CORRECTED FOR THE ONE STEP MISSED.

| | Step Length [cm] | | Distance [m] a) Real 31.20 b) Anthropometric 32.21 | Step Length Accuracy Error % | |
|---|---|---|---|---|---|
| Test | Real | Model | c) Model | Relative | Absolute |
| 1 | 60.00 | 61.49 | 31.98 | 2.49 | 2.49 |
| 2 | 60.00 | 58.00 | 30.16 | -3.33 | 3.33 |
| 3 | 60.00 | 59.81 | 31.10 | -0.32 | 0.32 |
| 4 | 60.00 | 58.08 | 30.78 * | -3.21 | 3.21 |
| 5 | 60.00 | 58.95 | 30.66 | -1.75 | 1.75 |
| 6 | 60.00 | 58.84 | 31.17 * | -1.94 | 1.94 |
| mean | 60.00 | 59.20 | 30.78 | -1.34 | 2.17 |
| min | 60.00 | 58.00 | 30.16 | -3.33 | 0.32 |
| max | 60.00 | 61.49 | 31.98 | 2.49 | 3.33 |

The stride and step frequencies measured are presented in the same table. The error is less that in the original Method A. In the Method A (TABLE VII), the absolute error in the COM amplitude versus expected ranges from 1.04% to 18.92% with a mean value 11.05%. In the Modified Method A (TABLE VIII), the absolute error in the COM amplitude versus expected ranges from 0.18% to 5.92% with a mean value 3.88%. The Modified Method A is good: all steps are detected, except in some tests where one is missed, and the accuracy error is acceptable. We have the need for a development of this method for a better detection of all steps and for a more flexible tool to analyze data: the Method B.

TABLE V. METHOD A. TIMES, VELOCITIES, STRIDES, STEPS AND ACCURACY VELOCITY ERROR ARE PRESENTED FOR THE SIX TESTS. § FAILURE RECOGNITION OF ONE STEP.

| | Time [s] | Speed [m/s] | | Accuracy Speed Error % | | Frequency | |
|---|---|---|---|---|---|---|---|
| Test | | Real | Model | Rel. | Abs. | [Stride/s] | [Step/s] |
| 1 | 32.68 | 0.9547 | 0.9512 | -0.37 | 0.37 | 0.80 | 1.59 |
| 2 | 31.49 | 0.9907 | 0.9298 | -6.15 | 6.15 | 0.83 | 1.65 |
| 3 | 30.56 | 1.0209 | 0.9837 | -3.64 | 3.64 | 0.85 | 1.70 |
| 4 | 31.06 | 1.0044 | 0.9451 § | -5.90 | 5.90 | 0.80 | 1.64 |
| 5 | 31.95 | 0.9764 | 0.8712 | -10.77 | 10.77 | 0.81 | 1.63 |
| 6 | 29.84 | 1.0455 | 0.9777 § | -6.48 | 6.48 | 0.84 | 1.71 |
| mean | 31.26 | 0.9988 | 0.9431 | -5.55 | 5.55 | 0.82 | 1.65 |
| min | 29.84 | 0.9547 | 0.8712 | -10.77 | 0.37 | 0.80 | 1.59 |
| max | 32.68 | 1.0455 | 0.9837 | -0.37 | 10.77 | 0.85 | 1.71 |

TABLE VI. MODIFIED METHOD A. TIMES, VELOCITIES, STRIDES, STEPS AND ACCURACY VELOCITY ERROR ARE PRESENTED FOR THE SIX TESTS.

| | Time [s] | Speed [m/s] | | Accuracy Speed Error % | | Frequency | |
|---|---|---|---|---|---|---|---|
| Test | | Real | Model | Rel. | Abs. | [Stride/s] | [Step/s] |
| 1 | 33.77 | 0.9240 | 0.9546 | 3.31 | 3.31 | 0.77 | 1.54 |
| 2 | 32.78 | 0.9518 | 0.9331 | -1.96 | 1.96 | 0.79 | 1.59 |
| 3 | 32.35 | 0.9644 | 0.9712 | 0.71 | 0.71 | 0.80 | 1.61 |
| 4 | 32.09 | 0.9724 | 0.9475 | -2.56 | 2.56 | 0.81 | 1.62 |
| 5 | 33.05 | 0.9439 | 0.9390 | -0.52 | 0.52 | 0.79 | 1.57 |
| 6 | 31.69 | 0.9846 | 0.9732 | -1.16 | 1.16 | 0.82 | 1.64 |
| mean | 32.62 | 0.9568 | 0.9531 | -0.36 | 1.70 | 0.80 | 1.60 |
| min | 31.69 | 0.9240 | 0.9331 | -2.56 | 0.52 | 0.77 | 1.54 |
| max | 33.77 | 0.9846 | 0.9732 | 3.31 | 3.31 | 0.82 | 1.64 |

TABLE VII. METHOD A. EVALUATION OF THE COM AMPLITUDE DURING THE SIX TESTS.

| | COM Amplitude [cm] | | | | Mean Versus Expected Error % | |
|---|---|---|---|---|---|---|
| Test | Min | Max | Mean | Expected | Relative | Absolute |
| 1 | 1.50 | 6.00 | 4.86 | 4.81 | 1.04 | 1.04 |
| 2 | 0.82 | 6.00 | 4.32 | 4.81 | -10.19 | 10.19 |
| 3 | 1.25 | 6.00 | 4.55 | 4.81 | -5.41 | 5.41 |
| 4 | 0.91 | 6.00 | 4.49 | 4.81 | -6.65 | 6.65 |
| 5 | 0.39 | 6.00 | 3.90 | 4.81 | -18.92 | 18.92 |
| 6 | 0.61 | 6.00 | 4.46 | 4.81 | -7.28 | 7.28 |
| mean | 0.91 | 6.00 | 4.30 | 4.81 | -10.71 | 11.05 |
| min | 0.39 | 6.00 | 3.09 | 4.81 | -18.92 | 1.04 |
| max | 1.50 | 6.00 | 4.86 | 4.81 | 1.04 | 18.92 |

TABLE VIII. MODIFIED METHOD A. EVALUATION OF THE COM AMPLITUDE DURING THE SIX TESTS.

| | COM Amplitude [cm] | | | | Mean Versus Expected Error % | |
|---|---|---|---|---|---|---|
| Test | Min | Max | Mean | Expected | Relative | Absolute |
| 1 | 3.35 | 6.00 | 5.09 | 4.81 | 5.92 | 5.92 |
| 2 | 2.04 | 6.00 | 4.55 | 4.81 | -5.39 | 5.39 |
| 3 | 2.50 | 6.00 | 4.82 | 4.81 | 0.18 | 0.18 |
| 4 | 2.57 | 6.00 | 4.54 | 4.81 | -5.63 | 5.63 |
| 5 | 3.67 | 6.00 | 4.66 | 4.81 | -3.10 | 3.10 |
| 6 | 2.84 | 6.00 | 4.66 | 4.81 | -3.03 | 3.03 |
| mean | 2.83 | 6.00 | 4.72 | 4.81 | -1.84 | 3.88 |
| min | 2.04 | 6.00 | 4.54 | 4.81 | -5.63 | 0.18 |
| max | 3.67 | 6.00 | 5.09 | 4.81 | 5.92 | 5.92 |

### *3.2 The Data Analysis with Method B*

The Method B (TABLE XI) detected all 52 steps in every test. We do not yet used the correction for steady state but if we did, the results will be better. By using Mode 5, the absolute accuracy error in the measurement of the walking total path, as well as the step length (error is the same), ranges from 0.27% to 6.31% with a mean value 3.00%. With this modality, the relative values are both positives and negatives.

TABLE IX. METHOD B. MODE 5. REAL AND CALCULATED STEPS AND DISTANCES VALUES FOR THE SIX TESTS. THE METHOD B WITH MODE 5 IS USED FOR DATA PROCESSING.

| | Steps | Step Length [cm] | Step (and Distance) Length Accuracy Error % | | Distance [m] |
|---|---|---|---|---|---|
| Test | Real = 52 | Real = 60 | Relative | Absolute | Real = 31.20 |
| 1 | 52 | 63.8 ± 5.2 | 6.31 | 6.31 | 33.17 |
| 2 | 52 | 60.2 ± 7.5 | 0.27 | 0.27 | 31.28 |
| 3 | 52 | 60.9 ± 7.8 | 1.52 | 1.52 | 31.67 |
| 4 | 52 | 60.4 ± 6.2 | 0.71 | 0.71 | 31.42 |
| 5 | 52 | 57.0 ± 7.5 | -4.98 | 4.98 | 29.65 |
| 6 | 52 | 62.5 ± 7.5 | 4.22 | 4.22 | 32.52 |
| mean | 52 | 60,8 | 1,34 | 3,00 | 31.62 |
| min | 52 | 57,0 | -4,98 | 0,27 | 29.65 |
| max | 52 | 63,8 | 6,31 | 6,31 | 33.17 |

By using Mode 1 (TABLE X), the absolute accuracy error in the measurement of the walking total path, as well as the step length (error is the same), ranges from 1.58% to 6.80% with a mean value 3.72%. The Mode 1 underestimates the step length in every test. We can think of using a special correction factor to correct this underestimation to have more accuracy. This allows the definition of the proper filtering to be adopted by the two processing methods.

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TABLE X. METHOD B. MODE 1. REAL AND CALCULATED STEPS AND DISTANCES VALUES FOR THE SIX TESTS.

| | Steps | Step Lenght [cm] | Step (and Distance) Length Accuracy Error % | | Distance [m] |
|---|---|---|---|---|---|
| Test | Real= 52 | Real= 60 | Relative | Absolute | Real= 31.20 |
| 1 | 52 | 59.1 ± 3.4 | -1.58 | 1.58% | 30.71 |
| 2 | 52 | 57.1 ± 5.5 | -4.84 | 4.84% | 29.69 |
| 3 | 52 | 58.0 ± 6.4 | -3.32 | 3.32% | 30.17 |
| 4 | 52 | 58.1 ± 4.7 | -3.21 | 3.21% | 30.20 |
| 5 | 52 | 55.9 ± 6.5 | -6.80 | 6.80% | 29.08 |
| 6 | 52 | 58.5 ± 6.1 | -2.55 | 2.55% | 30.40 |
| mean | 52 | 57.8 | -3.72 | 3.72% | 30.04 |
| min | 52 | 55.9 | -6.80 | 1.58% | 29.08 |
| max | 52 | 59.1 | -1.58 | 6.80% | 30.71 |

In the two conditions 1) and 2) the BMI of the subject was very different. In the trial repetitions of the first protocol (six tests of a single healthy subject) and their data processing, is demonstrated that the model produced the expected values with very good accuracy. The use of coefficients C for evaluating $S_{mean}$ (mean anthropometric step length) carry out a good reference, but is not the measured value S by the model. In the Method A, non-all the steps are detected (one step is lost in two tests), but if the filter is changed as in the Method B, the 100% of the 52 steps are detected. In the Modified Method A, the absolute accuracy error in the measurement of the step length, ranges from a minimum 0.32% to a maximum 3.33% with a mean value 2.17%; in the Method B, it ranges from a minimum 0.27% to a maximum 6.31% with a mean value 3.00%. These values do not yet use the correction for the steady state (used by Modified Method A), but if we did, the results should be better, so the equivalent values of the Method A to be compared are the minimum 0.37%, the maximum 10.77% and the mean value 5.55%. The Method B is better and more flexible. The parameters extracted by the model are very complete, but should be compared with the gold standard values using appropriate protocols. Preliminary reliability of both methods is more than good. This first validation is now followed by a protocol application on a wider population of healthy subjects and post stroke patients, to validate also its clinical application. The tests with an electronically controlled treadmill are carried out on healthy subjects following the dedicated protocol. We are also investigating the clinical applicability of the approach on a group of stroke patients undergoing rehabilitation programs. The tests performed are the 10m test at low and high walking speed, the 6MWT and the TUG. A control group of healthy subjects is considered. The expectation is to develop a dedicated tool for supporting diagnosis and rehabilitation. This will also allow for investigation of model sensitivity in detecting the gait parameters improvements.

## REFERENCES


[1] W. Tao, T. Liu, R. Zheng and H. Feng, "Gait Analysis Using Wearable Sensors", Sensors (2012) 12: pp. 2255-2283;

[2] A. Muro-de-la-Herran, B. Garcia-Zapirain and A.Mendez-Zorrilla, "Gait Analysis Methods: An Overview of Wearable and Non-Wearable Systems, Highlighting Clinical Applications", Sensors (2014) 14: pp. 3362-3394;

[3] S. Patel, H. Park, P. Bonato, L. Chan and M. Rodgers, "A review of wearable sensors and systems with application in rehabilitation", Journal of NeuroEngineering and Rehabilitation (2012) 9:21;

[4] I. Spulber, E. Papi, Y.-M. Chen, S. Anastasova-Ivanova, J. Bergmann, P. Georgiou and A.H. McGregor, "Development of a wireless multi-functional body sensing platform for smart garment integration", Biomedical Circuits and Systems Conference (BioCAS), 22-24 Oct. 2014 IEEE;

[5] A.Pantelopoulos, N. G. Bourbakis, "A Survey on Wearable Sensor-Based Systems for Health Monitoring and Prognosis", IEEE Transactions on Systems, Man and Cybernetics, Part C (Applications and Reviews), (2010) 40 (1): pp. 1-12;

[6] W. Zijlstra and A. L. Hof, "Assessment of spatio-temporal gait parameters from trunk accelerations during human walking", Gait and Posture (2003) 18: pp. 1-10;

[7] N. A. Capela , E. D. Lemaire and N. Baddour, "Novel algorithm for a smartphone-based 6-minute walk test application: algorithm, application development, and evaluation", Journal of NeuroEngineering and Rehabilitation (2015) 12:19;

[8] M. S. Orendurff, A. D. Segal, G. K. Klute, J. S. Berge, E. S. Rohr and N.J. Kadel, "The effect of walking speed on center of mass displacement", J Rehabil Res Dev, Nov-Dec (2004) 41 (6A): pp. 829-834;

[9] K. G. Holt, J. Hamill and R. O. Andres, "The force-driven harmonic oscillator as a model for human locomotion", Human Movement Science (1990) 9: pp. 55-68;

[10] http://www.sxt-telemed.it/Products/Protheo/Protheo.aspx, last accessed 22 february 2017;

[11] T. Oberg, A. Karsznia and K. Oberg, "Basic gait parameters: reference data for normal subjects, 10-79 years of age", Journal of rehabilitation research and development (1993) 30 (2): pp. 210-223;

[12] S. Al-Obaidi, J. C. Wall, A. Al-Yaqoub, M. Al-Ghanim, "Basic gait parameters: A comparison of reference data for normal subjects 20 to 29 years of age from Kuwait and Scandinavia." Journal of rehabilitation research and development (2003) 40 (4): pp.361-366;

[13] H. Tanaka, K. D. Monahan, D. R. Seals, "Age-predicted maximal heart rate revisited", J Am Coll Cardiol. (2001) 37 (1): pp. 153-156;

[14] E. Billauer, Peak detect using Matlab, http://www.billauer.co.il/peakdet.html, last accessed 22 february 2017;

[15] R. Moe-Nilssen and J. L. Helbostad, "Estimation of gait cycle characteristics by trunk accelerometry", Journal of Biomechanics (2004) 37: pp. 121-126;

[16] T. Jurcevic Lulic and O. Muftic, "Trajectory of the human body mass centre during walking at different speed", DS 30: Proceedings of DESIGN 2002, the 7th International Design Conference, Dubrovnik (2002).


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