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Research Article

Method for Wearable Kinematic Gait Analysis Using a Harmonic Oscillator Applied to the Center of Mass

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A model based on a harmonic oscillator describing human walking and balance with the sinusoidal trajectory of the center of mass of a subject during gait is presented. This model allows overcoming the traditional drift due to the double integration of raw acceleration data. The protocol uses a single 3D accelerometer worn at the pelvis level. The system computes the spatiotemporal gait and balance parameters when the subject is walking with or without aids. An incremental methodological approach is proposed and followed in the implementation and accuracy assessment. Eleven healthy subjects have participated to the study performing 6 trials over a fixed linear walking path at a self-selected speed. For reference, the protocol has imposed the execution of 52 steps whose length has been fixed at 60 cm. Different processing methods have been implemented and tested. The model identifies steps, walking time, and stepping frequency with an excellent reliability (absolute percentage accuracy error < 5%). When the information about the expected step length is given to the model, the percentage error in the measure of walking distance and speed is 3.25%. Without this input, this error rises to 4.95%, while for the anthropometric method is 3.68%.

1. Introduction

Gait analysis with optoelectronic technology represents the reference gold standard for functional tests. Despite its recognized accuracy, it is complex and expensive in time and equipment, only few steps in a path can be analyzed and it requires a skilled operator to perform it. From a user's perspective and for the rehabilitation follow-up, it would be very useful to identify a way to perform gait analysis with nonintrusive technologies and to monitor for long time in natural conditions. These aspects are very important both for athletes (amateur and professionals) and for patients during their rehabilitation at hospital or at home. Wearable devices allow for monitoring subject's motor behavior everywhere and without affecting the natural and normal executions of movements and activities [1, 2]. In the field of wearable devices, we can also include smart textiles, that is, fabrics that are themselves the sensors or that can embed miniaturized and even flexible devices in their structure or layers [3, 4]. Big data are produced, and another challenge is the development of software algorithms to process data and to compute the quantitative parameters we are interested in. Fusion methods to merge data from different accelerometers and gyroscopes have been developed to obtain the kinematics of body segments both in specific districts and in total body configurations [3]. Today, gait analysis with wearable devices is possible, and research is focusing on these methods given that they have good reliability and user-friendliness. These systems also allow for long-term monitoring in ecologic settings, that is, for collecting data of patients at home, outdoor, and everywhere [1–5]. The design of integration between devices and smart textiles to produce products and services requires an interdisciplinary approach where electronic components together with different factors concerning clothes (anthropometric, aesthetical, elasticity, and washability) and the environment of use (on the ground, in the air, or in the water) must be merged [6–8].

In this frame, we have considered that wearable devices can represent an ideal platform supporting the remote execution of common clinical motor tests for continuity of care models. For instance, the 6MWT (six-minute walking test),

Table 1: The segmented validation approach to test the wearable sensor in walking.

Task number	Rationale	Protocol setup
1	Healthy subject monitoring at self-selected natural speed	Walking over a linear path, one healthy subject only, repeated tests, fixed step length at the natural anthropometric step length value, fixed total distance of a traveled path, and fixed total number of steps
2	Population of healthy subjects to be monitored at self-selected natural speed	Extension of task number 1 to a population of healthy subjects with different gender and BMI
3	Healthy subjects to be monitored at different walking speeds	Walking on a treadmill to evaluate the model when the subject is moving at different speeds
4	Monitoring of pathological subjects with asymmetrical gait at self-selected natural speed	Identification of stroke as target pathology and tests with a small population of subjects walking over a linear path
5	Validation with respect to gold standard technology	Comparison of simultaneous recordings using gold standard reference (optoelectronic gait analysis, video recording, and ground reaction forces with the Davis protocol) with healthy and poststroke patients
6	Implementation of automatic execution and reporting of functional clinical tests	Walking over a linear path, a population of healthy subjects with different gender and BMI, repeated tests, performing the 6MWT (six-minute walking test), the TUG (timed up and go), and the 10-meter and 50-meter walking test at normal or maximum self-selected speed
7	Extension of automatic execution and reporting of functional clinical tests to patients using specific aids	Extension of task number 6 to a small population of poststroke subjects with different gender and BMI and using walking aids (stick, quadripod stick, and walker)
8	Validation with respect to standard functional protocols	Extensive validation of task number 7 to a population of poststroke subjects with different gender, BMI, and walking aids
9	Automatic recognition of free gait	Walking from a linear path to a variable path
10	Analysis of running patterns (extension from walking to running)	Identification and analysis of the running pattern
11	Automatic recognition of free movements	Activity pattern recognition of categories of movement (gait, running, sit-to-stand, stairs, ...)
12	Integration of metabolic indexes	Validation of energy expenditure
13	Application to large population	Extension of the method to different clusters of subjects

the TUG (timed up and go), and the 10-meter walking test with normal or maximum self-selected speed are normally used to evaluate subject's performance. These tests can be easily implemented with a wearable system and also capable to provide quantitative automatic reporting instead of the manual measurement of distance, or time or number of steps that are the only reported data; this aspect is relevant to increase the functional data that such protocols could provide for clinical assessment and also to simplify clinicians and patients in their profession and life. This requires the development of a simple and reliable biomechanical model to have a quick and efficient data interpretation.

The development of fine biomechanical models and the improvement of signal processing through advanced analysis algorithms open new perspectives to enhance the interpretation of the output data of wearable sensors for decision making [2, 9, 10]. In this context, the approach we follow is the protocol simplicity to obtain the maximum usability, which means to use only one 3D inertial measurement unit. The corresponding kinematic model is designed to comply with every walking mode (at different speeds, asymmetrical due to pathology, and supported by assistive technologies like stick or walker) due to its simplicity but strong adaptability, thanks to the set of harmonic oscillators that can be implemented. This model is intended not to extract the joint kinematics, but it is limited to compute the standard gait parameters. The complete path for the validation of the model (and related IMU-based device) in healthy and pathological gait analysis is presented in Table 1. Tasks 1–7 have been completed. This paper focuses on the achievements from task number 2, while results of task number 1 has been presented in [11]. Processing methods are compared with various parameters in order to propose the best setup for gait kinematics assessment for single IMU-based systems.

2. Theoretical Model

2.1. Biomechanical Model. The biomechanical model introduced in [11] simplifies the human anatomical structure into a rigid body with the joints which are connected to the bars that represent the legs. Also, the legs are considered as a rigid body hinged on an axis passing through the center of mass (COM) without oscillations in the mediolateral (coronal) plane or in the anterior-posterior (transverse) plane. The swinging movement of the legs in the execution of the steps is assumed to be an oscillation of an equivalent pendulum, and the natural balance is obtained with the legs aligned along the vertical during standing. The COM is a single point where it can be assumed that the whole mass of the body is concentrated. When a subject is at rest in a standing posture,

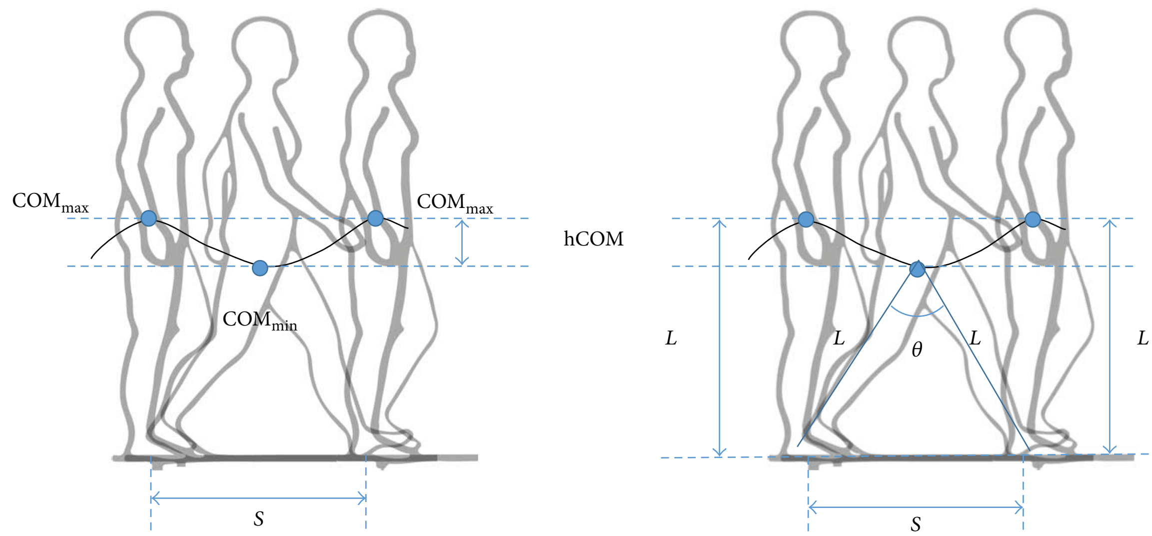

Figure 1: The walking pattern of a single gait cycle: sinusoidal oscillation of the COM in the sagittal plane.

his/her COM position is about 10 cm lower than the navel, in the sagittal plane and in correspondence of the anterior superior iliac crests (the top of the hipbones). The external forces acting on subject's body are equivalent to the same forces acting on the COM, whose trajectory describes the motion of the body (Figure 1). To better explain the basics of the proposed biomechanical model, the COM trajectory and the other related parameters are considered. In the vertical and mediolateral planes, the COM moves along an oscillating path following a quasi-sinusoidal pattern [12, 13]. In the proposed study, only the sagittal plane is considered. The trajectory of COM position during walking in the sagittal plane can be assumed to have a sinusoidal pattern as illustrated in Figure 1. Circles highlight the COM maximum/minimum positions in the trajectory; L is the leg length, θ is the hip angle in the sagittal plane, and S is the step length.

When the subject is in double stance phase, hip angle is θ_{max}, COM descends from its highest point to the lowest one; hCOM is the amplitude of this oscillation. Each step (both left and right) is carried out following this phenomenon. A simple harmonic oscillator consists of a mass m, which experiences a single force F, which pulls the mass in the vertical direction of the point and depends only on the mass's position, and is constant without being driven or damped. Its characteristic motion is the same trajectory of COM as in Figure 1. The use of a harmonic oscillator model allows for exploring the human locomotion and analyzing the correlation between the cycle of COM positions and the cycle of walking. A set of harmonic oscillators, one for each step cycle, is then adopted. The oscillation time allows defining the frequency and cadence of stepping.

In order to describe the swinging movement of the legs, according to the pendulum model [11, 14], the length of step S is given by the following equation:

$$S = 2 * \sqrt{2 * L * \mathrm{hCOM} - \mathrm{hCOM}^2}, \qquad (1)$$

where

(i) S = length of the step;

(ii) L = length of the lower limb;

(iii) hCOM = maximum amplitude of the vertical variation of COM trajectory (vertical distance between the maximum and the minimum position of the COM during the cycle of the step);

(iv) θ = hip angle in the sagittal plane.

L is known for each subject by anthropometric measure; when hCOM is evaluated from acceleration data measured by the device, (1) allows for computing S value.

If the COM of the subject moves forward with a constant speed, it has a null acceleration in the forward direction, but it has no null acceleration along vertical and mediolateral directions. A direct integration of the raw accelerometer data gives the velocities; a direct integration of the velocities gives the positions. This double direct integration can result in an accumulation of drift error giving wrong velocity, wrong position, and wrong distance and therefore wrong step length. In the literature, there are solutions to solve the drift error so that it is possible to integrate twice the raw acceleration data and to measure distance [9, 15]. By integration, the typical percentages of error over the walking distance are between 2.5% and 5.0% [16]. Other studies face the interpretation of 3D movements of the COM with more complex models [17–21]. Actually, it is possible to measure hCOM without carry out a double integration. The proposal of the harmonic oscillator wants to overcome this drift, thanks to its intrinsic properties; the acceleration of a harmonic oscillator is directly proportional to its position. For every walking cycle, in evaluating the hCOM amplitude, the step length S is obtained using (1). The incremental traveled distance is the incremental sum of these steps [11]. We evaluated the step length also through a correlation with anthropometric measures. It defines the parameter C as the proportional coefficient between subject's height and step length in function of

Table 2: Coefficients to compute the step length from subject's stature depending on gait speed and gender.

Speed	Low < 0.90 m/s	Normal (0.90–1.40 m/s)	Fast > 1.40 m/s
Male	0.2928	0.3422	0.3956
Female	0.3102	0.3539	0.3994

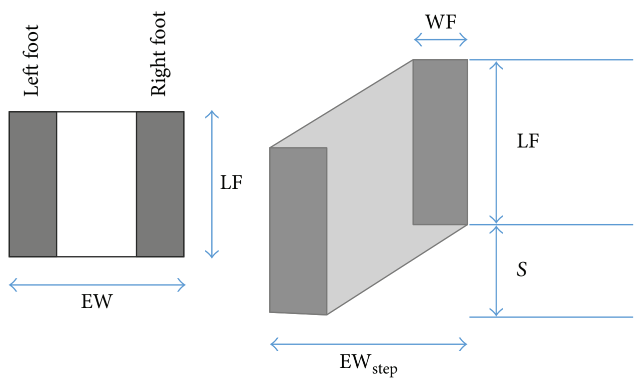

Figure 2: Evaluation of the base of support (BOS). The dark areas are the projection at ground of left and right feet. (a) The subject is standing. Equation (3) computes BOS. (b) The subject is stepping. When the subject moves the foot to step, the BOS presented in (a) changes. When the subject has complete support in double stance, the feet take the positions as in (b) and (4) computes new BOS.

gender and gait speed as shown in Table 2. We have produced the values of C from anthropometric data of Swedish adult people reported in [22, 23] assumed as the representative of the Caucasian population. The reference values of Table 1 to define low, normal, and fast speed are initially derived from the same data. Equation (2) allows computing the mean anthropometric step length S_{anth}:

$$S_{anth} = C * H. \tag{2}$$

During data processing, in some algorithms, S_{anth} is used as an average reference value for model prediction as described later in the proper section.

2.2. A Model for the Computation of the Base of Support (BOS). The base of support is "the area of ground surface (between and beneath the feet) covered by the body silhouette in an erect subject; the wider the base of support, the greater the stability of the erect body; the center of gravity is more easily maintained within the base of support" [24].

In order to compute the base of support (BOS), we define (Figure 2)

(i) S = length of the step;

(ii) LF = length of the single foot to the ground;

(iii) WF = width of the single foot to the ground;

(iv) EW = outer distance between the feet when the subject is standing at rest with natural balance;

(v) EW_{step} = outer distance between the feet when the subject is stepping.

When the subject is standing (Figure 2(a)), the BOS is

$$BOS_{rest} = EW * LF. \tag{3}$$

When the subject makes a step (Figure 2(b)), the BOS_{step} is

$$BOS_{step} = EW_{step} * LF + S * WF. \tag{4}$$

If the distance between the feet increases, then the BOS and the stability of the subject increase.

When the speed increases, EW_{step} decreases. If we label BOS_{step0}, the BOS associated with $step_0$, and BOS_{step1}, the BOS associated with the next $step_1$, the following formula can be obtained (Note: if the speed is constant, then P has a value equal to 1 and the second component is null):

$$BOS_{step1} = P * BOS_{step0} + (1 - P) * (S * WF), \tag{5}$$

where $P = EW_{step1}/EW_{step0}$.

2.3. BOS Model with Walking Aids: Stick in the Hand. The use of a walking aid requires some adjustments in the calculation of the BOS, with the request of some additional input parameters. If the subject uses a walking stick, we considered that when the subject is standing (Figure 3(a)), it is placed on the ground at a distance W_{stick} from the outer midpoint of the corresponding foot; we regard W_{stick} as constant in all the way. The stick contribution $BOS_{stickrest}$ to BOS is

$$BOS_{stickrest} = W_{stick} * \frac{LF}{2}. \tag{6}$$

Initially, at rest, the total BOS is

$$BOS_{totstickrest} = BOS_{rest} + BOS_{stickrest}. \tag{7}$$

When the subject moves a step (Figure 3(b)), one of the foot and the walking stick are in new positions so the BOS changes:

$$BOS_{totstickstep} = BOS_{step} + BOS_{stickstep}, \tag{8}$$

$$BOS_{stickstep} = BOS_{stickrest} + BOS_{stickvar}, \tag{9}$$

$$BOS_{stickvar} = \left(\frac{1}{2}\right) * \left((EW - WF) * \left(S - \left(\frac{LF}{2}\right)\right) + W_{stick} * S\right). \tag{10}$$

For each step, in the computation of $BOS_{stickvar}$ (10), the first term depends on the length and width of the steps and the second one depends on how the walking stick is put on the ground.

When the speed increases, EW_{step} decreases. If we label $BOS_{stickstep0}$, the $BOS_{stickstep}$ associated with $step_0$, and $BOS_{stickstep1}$, the $BOS_{stickstep}$ associated with the next $step_1$, the following formulas can be obtained:

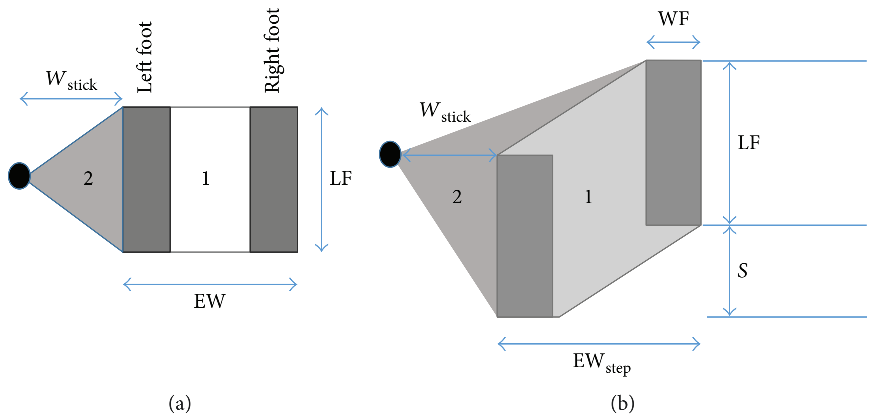

Figure 3: Evaluation of the base of support (BOS) when the subject walks using a walking stick (circle). (a) The subject is standing. (b) The subject is walking. The distance W_{stick} between the walking stick and the foot is constant in all the way. The area 1 is the BOS without an aid. Area 2 is the residual increased BOS because the walking stick is in use.

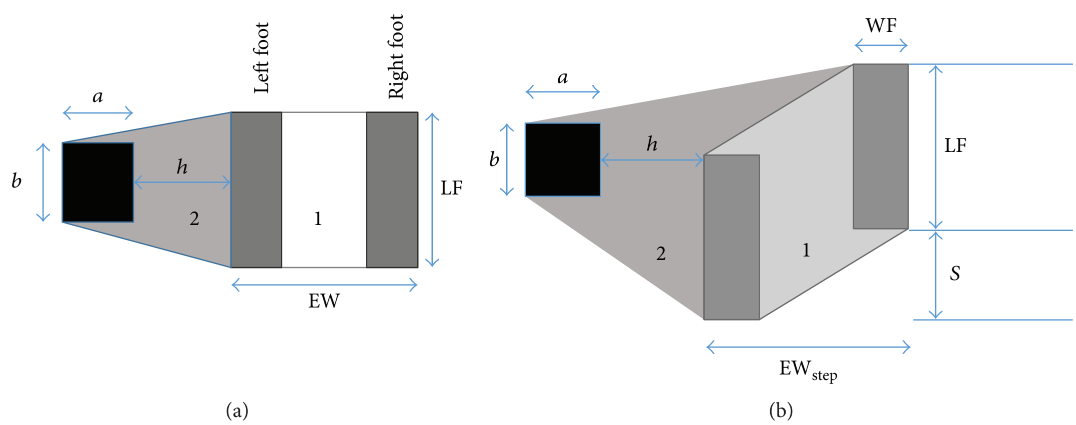

Figure 4: Evaluation of the BOS when the subject walks using a walking quadripod stick (black rectangle). (a) The subject is standing. (b) The subject is walking. The distance h between the walking quadripod stick and the foot is constant in all the way. The area 1 is the BOS without an aid. Area 2 is the residual increased BOS because the walking quadripod stick is in use.

$$\begin{gathered}\mathrm{BOS}_{\text{stickstep1}} = \mathrm{BOS}_{\text{stickstep0}} + \mathrm{BOS}_{\text{stickstepP}},\\ \mathrm{P} = \frac{\mathrm{EW}_{\text{step1}}}{\mathrm{EW}_{\text{step0}}},\\ \mathrm{BOS}_{\text{stickstepP}} = (P-1) * \left(\frac{\mathrm{EW}}{2}\right) * \left(S - \frac{\mathrm{FL}}{2}\right).\end{gathered} \tag{11}$$

If the speed is constant, then P has a value equal to 1 and the second factor is null. If the speed should unrealistically increase (because the subject who uses the walking stick is not healthy) until the feet are on the same line (running), then $\mathrm{EW}_{\text{step1}} = \mathrm{WF}$ and

$$\mathrm{BOS}_{\text{stickvar}} = \left(\frac{1}{2}\right) * \left(W_{\text{stick}} * S\right). \tag{12}$$

2.4. BOS Model with Walking Aids: The Quadripod Stick. If the subject uses a walking quadripod stick, it is considered that when the subject is standing (Figure 4(a)), it is placed on the ground at a distance h from the outer midpoint of the corresponding foot; therefore, h is assumed constant. The stick contribution $\mathrm{BOS}_{\text{stickrest}}$ to BOS is computed as follows:

$$\mathrm{BOS}_{\text{qstickrest}} = \frac{[(\mathrm{LF} + b) * (h + a)]}{2}. \tag{13}$$

Initially, at rest, the total BOS is

$$\mathrm{BOS}_{\text{totqstickrest}} = \mathrm{BOS}_{\text{rest}} + \mathrm{BOS}_{\text{qstickrest}}. \tag{14}$$

When the subject moves a step (Figure 4(b)), one of the foot and the walking quadripod stick are in new positions so the BOS changes:

$$\begin{gathered}\mathrm{BOS}_{\text{totqstickstep}} = \mathrm{BOS}_{\text{step}} + \mathrm{BOS}_{\text{qstickstep}},\\ \mathrm{BOS}_{\text{qstickstep}} = \mathrm{BOS}_{\text{qstickrest}} + \mathrm{BOS}_{\text{qstickvar}},\\ \mathrm{BOS}_{\text{qstickvar}} = \left(\frac{1}{2}\right) * \left((\mathrm{EW} - \mathrm{WF}) * \left(S - \frac{\mathrm{LF}}{2} + \frac{b}{2}\right) + (h + a) * S\right).\end{gathered} \tag{15}$$

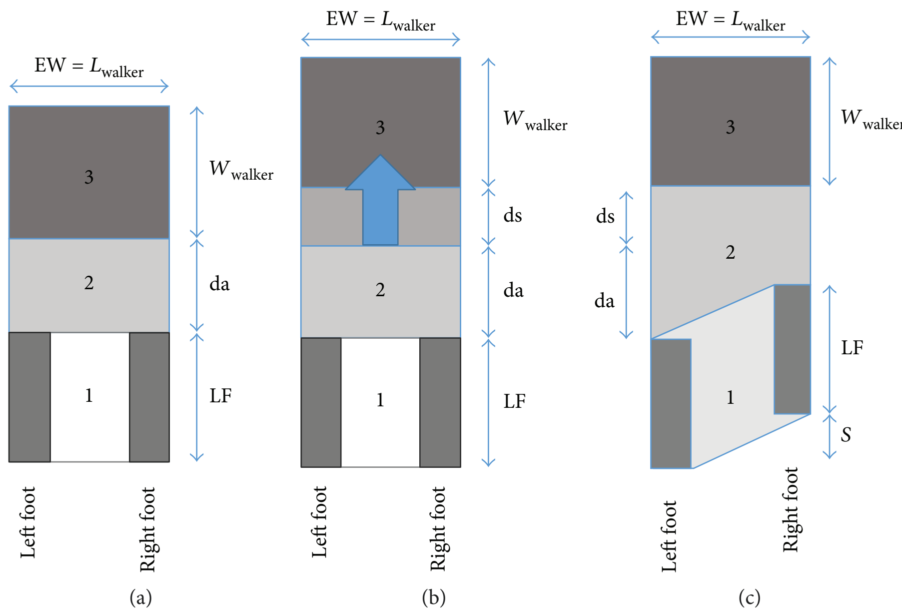

Figure 5: Evaluation of the base of support (BOS) when the subject walks using a walker. The area 3 is the walker with dimension W_{walker} (w) and L_{walker} (l); the area 1 is the BOS without the aid; the area 2 is the BOS between the subject and the walker. (a) The subject is standing. (b) When the subject moves the walker forward a distance ds, the areas 1b and 3b are constant and the area 2b increases. (c) The subject is walking. Areas 1a and 1b are BOS_{rest}. Area 1c is BOS_{step}. Area 2a is $\text{BOS}_{\text{walker0}}$. Area 2b is $\text{BOS}_{\text{walker1}}$. Area 2c is $\text{BOS}_{\text{walker2}}$. Areas 3a, 3b, and 3c are $\text{BOS}_{\text{walker}}$.

For each step, in the computation of $\text{BOS}_{\text{stickvar}}$ (10), the first term depends on the length and width of the steps and the second one depends on how the walking stick is put on the ground.

When the speed increases, EW_{step} decreases. If we label $\text{BOS}_{\text{stickstep0}}$, the $\text{BOS}_{\text{stickstep}}$ associated with step_0 and BO $\text{S}_{\text{stickstep1}}$, the $\text{BOS}_{\text{stickstep}}$ associated with the next step_1, the following formulas can be obtained:

$$\text{BOS}_{\text{stickstep1}} = \text{BOS}_{\text{stickstep0}} + \text{BOS}_{\text{stickstepP}},$$
$$P = \frac{\text{EW}_{\text{step1}}}{\text{EW}_{\text{step0}}}, \qquad (16)$$
$$\text{BOS}_{\text{qstickstepP}} = (\text{P} - 1) * \left(\frac{\text{EW}}{2}\right) * \left(S - \frac{\text{LF}}{2} + \frac{b}{2}\right).$$

If the speed is constant, then P has a value equal to 1 and the second factor is null. If the speed should unrealistically increase (because the subject who uses the walking quadripod stick is not healthy) until the feet are on the same line (running), then $\text{EW}_{\text{step1}} = \text{WF}$ and

$$\text{BOS}_{\text{qstickvar}} = \left(\frac{1}{2}\right) * (h + a) * S. \qquad (17)$$

2.5. BOS Model with Walking Aids: The Walker. The evaluation of the BOS when the subject walks using a walker is as reported in Figure 5. The walker has the width W_{walker} and length L_{walker}. We assumed that the walker only moves forward. EW is constant (the movement is low and constant) and $W_{\text{walker}} = \text{EW}$. The area of the walker (Figure 5(a)) is described by the following formulas:

$$\text{BOS}_{\text{walker}} = W_{\text{walker}} * L_{\text{walker}}. \qquad (18)$$

If the distance at rest between the foot line and the walker is denoted da, we considered an area.

$$\text{BOS}_{\text{walker0}} = W_{\text{walker}} * \text{da}. \qquad (19)$$

When the subject moves the walker for a certain distance ds (Figure 5(b)), BOS increases as follows:

$$\text{BOS}_{\text{walker1}} = \text{BOS}_{\text{walker0}} + W_{\text{walker}} * \text{ds}. \qquad (20)$$

If the subject moves a step (Figure 4(c)), this area becomes

$$\text{BOS}_{\text{walker2}} = \text{BOS}_{\text{walker1}} - (W_{\text{walker}} + \text{LF}) * \left(\frac{S}{2}\right). \qquad (21)$$

Initially, in standing, the total BOS is

$$\text{BOS}_{\text{totwalkerRest0}} = \text{BOS}_{\text{walker0}} + \text{BOS}_{\text{walker}} + \text{BOS}_{\text{rest}}. \qquad (22)$$

When the subject moves only the walker standing still with his feet, the area is

$$\text{BOS}_{\text{totwalkerRest1}} = \text{BOS}_{\text{walker1}} + \text{BOS}_{\text{walker}} + \text{BOS}_{\text{rest}}. \qquad (23)$$

When the subject moves a step, the total BOS becomes

$$\text{BOS}_{\text{totwalkerStep}} = \text{BOS}_{\text{walker2}} + \text{BOS}_{\text{walker}} + \text{BOS}_{\text{step}}. \qquad (24)$$

The width of footstep is measured while the subject is standing at the starting point of the test; this initial value is then multiplied with the coefficients obtained by the linear interpolation of the factors depending on speeds provided

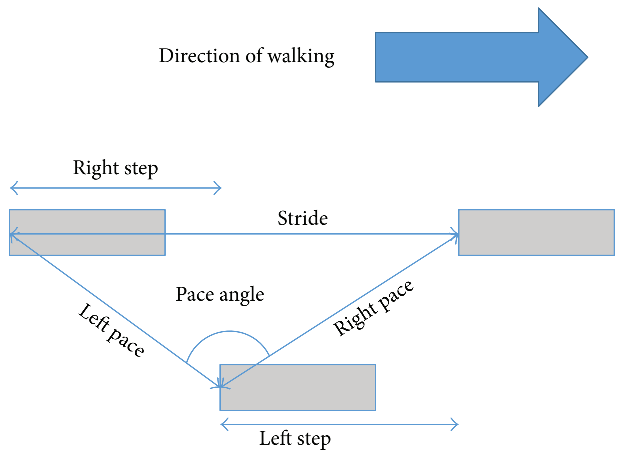

Figure 6: Pace parameters when the subject walks.

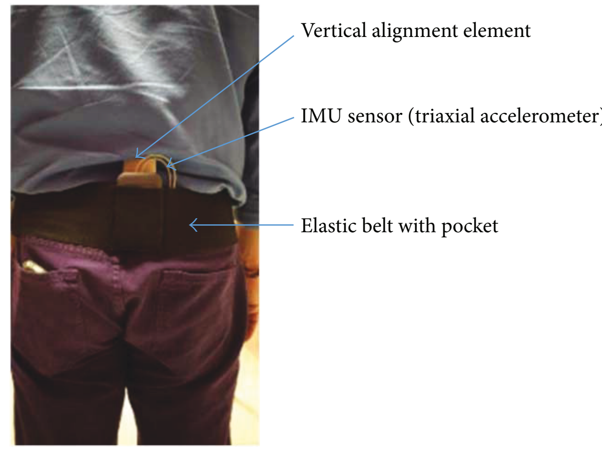

Figure 7: The system setup on the back of the subject.

by Orendurff et al. [12], with proper weighting coefficients according to subject anthropometry.

2.6. Pace Evaluation. The pace parameters (Figure 6) are computed according to the following formulas:

$$\begin{aligned} \text{Pace}_{\text{i}} &= \sqrt{S_{\text{i}}^2 + (\text{EW} - \text{WF})^2}, \\ \varphi_{\text{i}} &= \arctan\left(\frac{S_{\text{i}}}{(\text{EW} - \text{WF})}\right), \\ \varphi &= \left(\varphi_{\text{left}} + \varphi_{\text{right}}\right), \\ \text{CWE} &= \frac{\varphi}{180^\circ}, \end{aligned} \tag{25}$$

where i is the side of the body (left or right), φ is the pace angle, and CWE is the coefficient of walking efficiency.

3. Materials and Methods

3.1. Experimental Setup. In order to measure the acceleration of COM, the IMU must be placed near the COM itself. For this reason, the experimental protocol uses a single IMU—for the maximum simplification of the procedure and equipment—with a triaxial accelerometer on the pelvis of the subjects next to the second sacral vertebra (Figure 7). An elastic belt with a pocket firmly fixes the device to the body. This is done to avoid artifacts in raw data originating from movements of the device with respect to the COM position. A properly sized element is used to align the device along the vertical direction as gravity force acts.

The wearable 3D accelerometer [25] has size of 85 (l) × 53 (w) × 16 (h) mm, weight 70 g, 4 digits LCD, on board ARM7 microprocessor, and 128 Hz sampling frequency of raw accelerations. This device logs 3D accelerations 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. Therefore, data storage allows also for the recording of consecutive tests. IMU calibration, data analysis, and data processing are performed offline.

Eleven subjects (six males and five females whose anthropometric data are reported in Table 3, aged between 24 and 55 years with mean value of 38 years) have been examined. The weight of the subjects varies from 54 to 85 kg (mean value 70 kg). Their height has ranged from 1.58 to 1.92 m (mean value 1.73 m). Their BMI has ranged from 17.6 to 29.6 (mean value 23.4). S_{anth} has been calculated with (2) and the C coefficients of Table 2; it has ranged from 55.9 to 65.7 (mean value 60.0 cm).

Before starting the acquisitions, the subject has made some practice walking along the testing path, according to its need. When the subject feels ready and practiced with the environment and the setup, the test can start. Each subject is asked to walk over a linear path of 31.2 m at a self-selected speed. 52 markers (60 cm apart) are placed on the path to drive the position of the tip of the foot at each step. This was done in accordance with the intersubject average value of the step length and to have a standardization of this parameter. Each test session is repeated six times with a rest period of about 1 minute between the end of the recording and the start of the next tests in which the subject is asked to stand up still. The device is unworn when all six tasks are completed, and data are downloaded. All subjects walk with their shoes. The walking time is not constant depending on the self-selected subject's speed. The step length S has a fixed value of 60 cm, so that 52 steps are necessary to complete the linear path. These values are the true imposed values used as reference for accuracy assessment with a controlled setup. Shoes are included for anthropometric measurement on the initial setup. The following anthropometric segments, measured while the subject is standing, are given as input to the model: lower limb (ground-greater trochanter), ground-malleolus, lateral condyle-greater trochanter, malleolus-lateral condyle, and fifth metatarsal-malleolus. The width of the foot, length of the foot, and outer distance between the feet are acquired to the ground when the subject is resting in natural balance.

3.2. Data Processing. The raw acceleration processing is implemented in MATLAB© software suite. Two methods have been implemented and analyzed: the only two differences concern

TABLE 3: Anthropometric data of the subjects (legend: w = weight, h = height, A = age, BMI = body mass index, G = gender, S_{anth} = mean anthropometric step length).

Subj. number parameter	1	2	3	4	5	6	7	8	9	10	11
w (kg)	68	56	70	85	54	67	70	74	77	65	80
h (cm)	167	162	175	192	175	173	179	158	164	172	183
A (years)	46	25	24	31	38	48	40	32	55	44	35
BMI	24.4	21.3	22.9	23.1	17.6	22.4	21.8	29.6	28.6	22.0	23.9
G (m or f)	m	f	m	m	f	f	m	f	f	m	m
S_{anth} (cm)	57.1	57.3	59.9	65.7	61.9	61.2	61.3	55.9	58.0	58.9	62.6

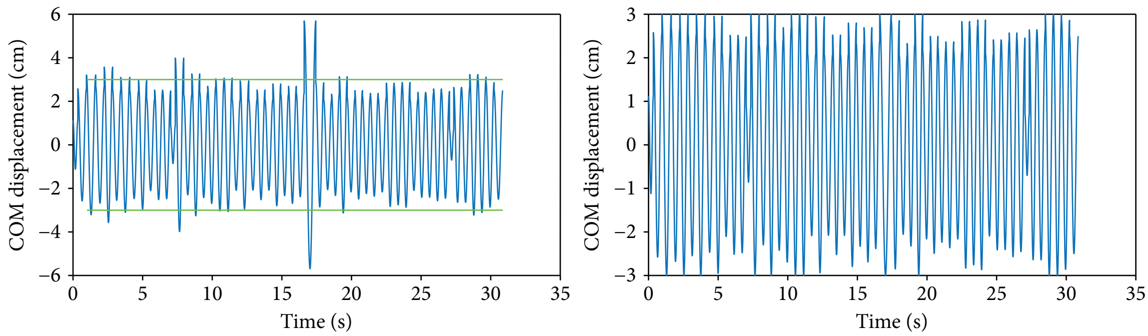

FIGURE 8: The amplitude cut-off threshold (green line) over the amplitude of the COM displacement pattern (left) and the final signal (right).

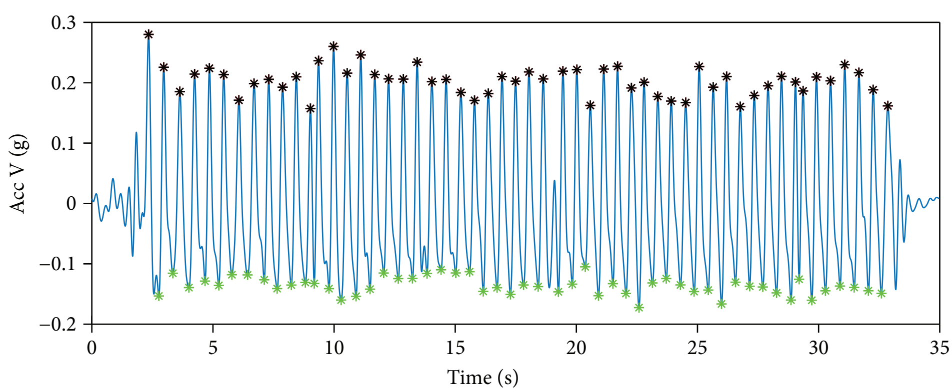

FIGURE 9: *Method A*. The black asterisks are the detected peaks. The time between two consecutive vertical peaks is the single-step time.

the cut-off frequency of the Butterworth low-pass filter applied to the raw signals and the cut-off threshold for the hCOM amplitude. Raw data have showed that without proper filtering and thresholding, the measurement of the hCOM can be overestimated. The hCOM threshold works only on the amplitude; the signal periodicity is preserved. The hCOM is a distance and so we used a relative measure unit to cut off its amplitude variation over time. Different processing modes (with small difference among them) are available and have been tested in reliability.

3.3. Processing Method A. Raw data has been filtered with a 5th-order passband Butterworth filter (band: 0.5–4 Hz) to identify the peaks. To assess S, we have applied the harmonic oscillator model to the original raw signals filtered with a low-pass 19th-order Butterworth filter. Anteroposterior, vertical, and mediolateral acceleration have, respectively, the following cut-off frequencies: 6 Hz, 7 Hz, and 8 Hz. The choice of these values depends on the power spectrum analysis of raw signals. The hCOM evaluation is carried out by applying a cut-off threshold of 6 cm to the double of the maximum amplitude of a COM pattern (Figure 8). The length distance between the maximum and the minimum for every COM oscillation is passed to the model as the identified amplitude. The time between two consecutive vertical peaks is the single-step time (Figure 9). The analysis of mediolateral acceleration allows for the identification of the first right or left leg support and the assessment of the asymmetry of right and left steps [26].

3.4. Processing Method B. The processing *method A* fits to and is applied to the analysis of data from subjects walking with a constant step and a constant velocity. These constraints imply a controlled setup, but normally subjects behave differently. To approach the natural walking setup in which the step length and speed of the subject can be variable, we

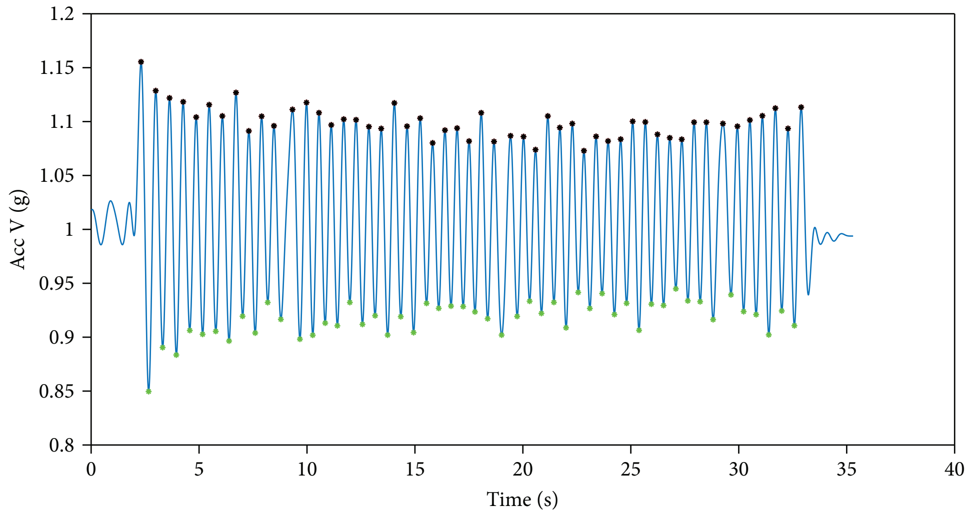

Figure 10: *Method B*. Vertical acceleration (g) versus time (s). The upper black asterisks are the detected peaks. The time between two consecutive vertical peaks is the single-step time.

Table 4: The six different approaches to define the amplitude hCOM threshold that characterize the six different processing modes.

Input to the model	Mode index	Definition of the amplitude of the hCOM threshold
Anthropometric step length	0	Median of twice the absolute value of the COM amplitude trend multiplied by a coefficient depending on the average walking speed
Expected S	1	Expected hCOM according to the following formula: $\mathrm{hCOM}_{\mathrm{expected}} = L - \sqrt{L^2 - \left(\frac{S}{2}\right)^2}$
Expected S	2	The same as in *mode 1* increased by 20%
Expected S	3	The product of hCOM obtained by the linear interpolation and a set of weighting coefficients depending on speeds, matching the value of COM displacement measured by Orendurff et al. at different speeds [12], with the expected COM displacement
Anthropometric step length	4	The COM amplitude evaluated by Lulic and Muftic [13] with the weights used in *mode 3*
Anthropometric step length	5	Set to 6 cm

analyzed a single subject walking on a treadmill where progressively the speed increases from 0.5 to 1.7 m/s. In these tests (task number 3 in Table 1), the subject is free to move as he wishes and we have a variation of both the step length and velocity. The search of the peaks of acceleration has led us to define the new filters to be applied to raw data. The experiments carried out in our work [11] confirm that choice. The same concept could be applied to pathological patients, walking slower and asymmetrically, such as stroke patients with walking speed < 0.5 m/s (task number 3 in Table 1). The peaks of raw signals identify the steps (Figure 10); the low-pass filter used is different according to the walking speed (high or low). At normal and high speed, we apply a 5th-order low-pass Butterworth filter. Anteroposterior, vertical, and mediolateral acceleration have, respectively, the following cut-off frequencies: 1.8 Hz, 1.8 Hz, and 0.9 Hz. At low speed, we apply a 4th-order low-pass Butterworth filter. Anteroposterior, vertical, and mediolateral acceleration have, respectively, the following cut-off frequencies: 35 Hz, 5 Hz, and 3 Hz. Detecting the acceleration peaks allows identifying number and timing of the steps, and, applying the harmonic oscillator model to the original raw signals filtered by a 19th-order Butterworth filter, it is possible to compute the step length S. Anteroposterior, vertical, and mediolateral acceleration have, respectively, the following cut-off frequencies: 6 Hz, 35 Hz, and 8 Hz.

To compute S, as required by (1), the parameter is the cut-off threshold of hCOM. We have tested six different approaches (called *modes*) indexed from *0* to *5* as described in Table 4.

One main difference among methods is that the expected step length can be used as input to the model. If this value is not available then *method B* cannot apply *modes 1*, *2*, and *3*. *Method A* and *method B mode 5* use a fixed amplitude threshold independent from the subject. The *anthropometric method* is obviously based on anthropometric data. Through the proper choice of the *mode* according to subject's feature in his/her different scenarios, the model should calculate the correct length values and the number of strides and steps.

3.5. Trials and Reference Values. The anthropometric method for data processing has been selected for comparison of the step length. All subjects have walked six trials. Each trial was analyzed through different processing modalities:

method A and *method B*. *Method B* has six different *modes* for hCOM thresholding. In total, we have 11 subjects, 6 trials per subject, and 7 processing techniques, producing finally 462 datasets; 11 datasets computed with the anthropometric method are added, for a total of 473 datasets. The distance, the steps, and the times obtained from the device were compared with the true reference value, that is, that one taken manually by the observing operator and/or defined by the protocol setup.

4. Results

Tables 5 and 6 report the results of the data processing.

Method A detected correctly 52 steps in 3 subjects, one step more in 2 subjects, two steps more in 5 subjects, and three steps more in 1 subject. The absolute accuracy error in the measurement of the step length has a maximum of 21.17% with a mean value of 8.48%. The absolute accuracy error in the measurement of the total distance has a maximum of 17.89% with a mean value of 7.79%.

Method B always detected correctly 52 steps for all the subjects and with all the *modes*. If the total distance traveled is calculated by using the mean step length multiplied for the number of steps, then the step length and the walked path have the same percentage error in accuracy; because all the steps are correctly detected and in a relative evaluation, the number of steps is a multiplier constant simplified to 1. If the variability of the step length is considered, then the mean absolute accuracy percentage error between step length and total distance can be different.

The mean absolute accuracy percentage error of the step length using *mode 0* is 6.07% and that of the total distance is 6.07% too, then mode 0 has good reliability. With this *mode*, the relative percentage error values are positive and negative in the subject population.

Instead, using *mode 1*, the mean absolute accuracy percentage error in the measurement of the step length and total distance are lower with values of 3.25% and 3.26%, respectively. *Mode 1* underestimates the step length in each test and for all subjects.

Mode 2 and *mode 5* have similar errors (average of absolute percentage error for the step length are 4.63% and 4.85%, resp., and that of the total distance are 4.62% and 4.95%, resp.); the relative percentage error values are both positive and negative; therefore, *modes 2* and *5* have excellent reliability.

Mode 3 has bigger values (mean absolute percentage error is 6.89% for step length and 6.89% for total distance); the relative percentage error values are both positive and negative, and *mode 3* demonstrates a good reliability.

Using *mode 4*, the mean absolute percentage error in the measurements of the step length and total distance are equal, respectively, to 18.05% and 10.04%, which is only sufficient reliability according to the selected criterion.

The *anthropometric mode* has excellent reliability (mean absolute percentage accuracy error of the step length and total distance are equal, respectively, to 3.70% and 3.68%); also with this *mode*, the relative percentage error values are both positive and negative.

The mean absolute percentage accuracy error for speed measurement is 3.25% for *mode 1*, 4.62% for *mode 2*, 4.95% for *mode 5*, and 3.68% for *anthropometric mode*. For the *other modes*, the error is bigger than 5%. Concerning the walking time, the reliability of measurements is excellent (Table 6). It has to be considered that the reference value is the one taken by an operator with a stopwatch, so even if he is an expert, it can include a certain implicit error due to the not null reaction time of the operator. For this reason, we used the time measured by the model as true time. Data about step and stride frequency (Table 6) are always very reliable for all *modes* of *method B* (excellent), except for *method A* that shows errors in step counting (good).

This study has also produced a first reference set of normalized values and ratio for new parameters and indexes related to gait and balance in standing and during walking (base of support and step width). Tables 7–10 show these data. The Supplementary Material (available here) contains all recorded data processed with *method A*, *method B—mode 0/1/2/3/4/5/anthropometric* algorithms.

5. Discussion and Conclusions

This study proposes a novel system using a single-wearable IMU to compute standard gait parameters and a set of novel kinematic indexes, investigating also their accuracy in a population of healthy subjects walking over a fixed traveled path with imposed step length to control the setup. A set of harmonic oscillators is the biomechanical mode interpreting the kinematics of 3D accelerations of COM. The method includes geometrical models for the assessment of the base of support in standing and during walking even when the subject makes use of walking aids to match all walking conditions for healthy and pathological subjects. Different algorithms (*methods A* and *B*, and different computational *modes* for this last method) have been developed and tested in a population of eleven healthy subjects. Anthropometric evaluation is also carried out.

Previous studies have no homogenous assessment grid. For this reason, here, a structured set of thresholds for reliability assessment is proposed, starting from the general accepted value of 5% [16, 27–30]. A gradual scale of reliability evaluation based on quantitative criteria for each spatiotemporal parameter and step number is used as acceptability criteria to evaluate the results (Table 11). The reference threshold applies to the mean percentage error in accuracy ($\varepsilon_{\%}$) of step length and distance.

The proposed criteria have been applied only to the absolute accuracy percentage error. This implies a stricter evaluation of method reliability because the relative accuracy percentage error could introduce some compensations (having both positive and negative values).

When using the processing *method A*, not all the steps are detected (from one to three steps are lost in several tests) and even if step length S error has a mean value of 8.48% (so sufficient) in some cases, it has a critical value of 21.17%. The same situation is shown by the total distance. Then it is to be concluded that *method A* is questionable,

Table 5: Data of the analysed parameters: mean and standard deviation for the subject population. The labels of the acceptability criterion (considering only the value for the absolute percentage error) are E = excellent, G = good, S = sufficient, and NA = not acceptable.

Parameter	Reference value	Average ± σ							
		A	B0	B1	B2	B3	B4	B5	Banth
Step number	52	53 ± 1	52 ± 0	52 ± 0	52 ± 0	52 ± 0	52 ± 0	52 ± 0	52 ± 0
Step count reliability		G	E	E	E	E	E	E	E
Step length (cm)	60	56.9 ± 5.5	59.0 ± 4.1	58.0 ± 1.3	60.9 ± 3.1	56.6 ± 3.3	49.2 ± 2.9	61.0 ± 3.3	59.9 ± 2.8
Step length accuracy percentage error	0	−5.15 ± 9.23	−1.74 ± 6.87	−3.26 ± 2.22	1.45 ± 5.22	−5.68 ± 5.43	−18.05 ± 4.83	1.73 ± 5.42	−0.21 ± 4.73
Absolute step length accuracy percentage error	0	8.48 ± 5.92	6.07 ± 3.15	3.26 ± 2.22	4.64 ± 2.44	6.89 ± 3.57	18.05 ± 4.83	4.85 ± 2.62	3.70 ± 2.71
Step length reliability		G	G	E	E	G	S	E	E
Distance (m)	31.2	30.29 ± 2.93	30.65 ± 2.14	30.18 ± 0.69	31.66 ± 1.62	29.43 ± 1.69	25.57 ± 1.51	31.78 ± 1.73	31.14 ± 1.47
Distance accuracy percentage error	0	−2.93 ± 9.38	−1.75 ± 6.87	−3.25 ± 2.21	1.46 ± 5.20	−5.67 ± 5.43	−18.04 ± 4.83	1.86 ± 5.54	−0.20 ± 4.71
Absolute distance accuracy percentage error	0	7.79 ± 5.53	6.07 ± 3.17	3.25 ± 2.21	4.62 ± 2.43	6.89 ± 3.55	18.04 ± 4.83	4.95 ± 2.74	3.68 ± 2.71
Distance reliability		G	G	E	E	G	S	E	E
Speed (m/s)	1.03 ± 0.06	0.99 ± 0.08	1.01 ± 0.07	0.99 ± 0.05	1.04 ± 0.04	0.97 ± 0.10	0.84 ± 0.05	1.04 ± 0.05	1.02 ± 0.07
Speed accuracy percentage error	0	−2.93 ± 9.38	−1.75 ± 6.87	−3.25 ± 2.21	1.46 ± 5.20	−5.67 ± 5.43	−18.04 ± 4.83	1.86 ± 5.54	−0.20 ± 4.71
Absolute speed accuracy percentage error	0	7.79 ± 5.53	6.07 ± 3.17	3.25 ± 2.21	4.62 ± 2.43	6.89 ± 3.55	18.04 ± 4.83	4.95 ± 2.74	3.68 ± 2.71
Speed reliability		G	G	E	E	G	S	E	E

Table 6: Data of walking time, step frequency and stride frequency: mean and standard deviation using the processing *methods A* and *B*.

Parameter	Reference	Model	Operator	Model accuracy error % Relative	Absolute	Reliability
Walking time (s)	Self-selected	30.5 ± 1.9	30.4 ± 2.0	0.48 ± 0.64	0.64 ± 0.47	E
Method A: step frequency (step/s)	1.71	1.75 ± 0.12	1.76 ± 0.12	2.62 ± 1.97	2.62 ± 1.97	G
Method A: stride frequency (stride/s)	0.85	0.88 ± 0.06	0.88 ± 0.06	2.62 ± 1.97	2.62 ± 1.97	G
Method B: step frequency (step/s)	1.71	1.71 ± 0.11	1.72 ± 0.11	0.00 ± 0.00	0.00 ± 0.00	E
Method B: stride frequency (stride/s)	0.85	0.85 ± 0.05	0.86 ± 0.05	0.00 ± 0.00	0.00 ± 0.00	E

Table 7: Coefficients normalized to stature for the evaluation of the base of support area in standing (AS) and during walking (AW) computed by processing *modes B0–B5*.

	AS	AW B0	AW B1	AW B2	AW B3	AW B4	AW B5
Min	0.0229	0.0391	0.0395	0.0391	0.0394	0.0392	0.0391
Mean ± SD	0.0343 ± 0.0055	0.0470 ± 0.0049	0.0471 ± 0.0049	0.0470 ± 0.0049	0.0467 ± 0.0047	0.0469 ± 0.0049	0.0470 ± 0.0048
Max	0.0443	0.0546	0.0547	0.0545	0.0545	0.0558	0.0545

Table 8: Ratio AR = AW/AS computed by processing *modes B0–B5*.

	AR B0	AR B1	AR B2	AR B3	AR B4	AR B5
Min	1.1410	1.1418	1.1487	1.1425	1.1256	1.1487
Mean ± SD	1.3883 ± 0.1554	1.3928 ± 0.1587	1.3888 ± 0.1517	1.3811 ± 0.1537	1.3853 ± 0.1535	1.3879 ± 0.1522
Max	1.7037	1.7251	1.7066	1.7208	1.7080	1.7066

Table 9: Coefficients normalized to height for the evaluation of the step width in standing (LS) and during walking (LW) computed by processing *modes B0–B5*.

	LS	LW B0	LW B1	LW B2	LW B3	LW B4	LW B5
Min	0.1000	0.9683	0.1091	0.7470	0.1218	0.7739	0.0923
Mean ± SD	0.1526 ± 0.0254	1.2412 ± 0.2008	0.1624 ± 0.0381	1.1417 ± 0.2131	0.1745 ± 0.0488	1.3214 ± 0.2622	0.1413 ± 0.0399
Max	0.1883	1.5989	0.2543	1.4905	0.3017	1.6692	0.2385

Table 10: Ratio LR = LW/LS computed by processing *modes B0–B5*.

	LR M0	LR M1	LR M2	LR M3	LR M4	LR M5
Min	0.5845	0.5693	0.5507	0.5845	0.6337	0.5507
Mean ± SD	0.7165 ± 0.1108	0.7605 ± 0.1460	0.6911 ± 0.1016	0.7395 ± 0.1260	0.8532 ± 0.1470	0.6869 ± 0.1030
Max	0.9296	1.0000	0.8794	0.9811	1.1600	0.8794

Table 11: Reliability assessment grid.

Parameter	Reliability criteria Excellent	Good	Sufficient	Not acceptable
Step length and distance	$\varepsilon_{\%} < 5\%$ (<1.56 m)	$5\% \le \varepsilon_{\%} < 10\%$ ($1.56 \le \varepsilon_{\%} < 3.12$ m)	$10\% \le \varepsilon_{\%} < 20\%$ ($3.12 \le \varepsilon_{\%} < 1.8$ m)	$\varepsilon_{\%} \ge 20\%$ (≥6.24 m)
Counting steps	$\varepsilon_{\%} < 2\%$ (<1 steps)	$2\% \le \varepsilon_{\%} < 4\%$ ($1 \le \varepsilon_{\%} < 2$ steps)	$4\% \le \varepsilon_{\%} < 6\%$ ($2 \le \varepsilon_{\%} < 3$ steps)	$\varepsilon_{\%} \ge 6\%$ (≥3 steps)
Time	$\varepsilon_{\%} < 4\%$ (<1.2 s)	$4\% \le \varepsilon_{\%} < 6\%$ ($1.2 \le \varepsilon_{\%} < 1.8$ s)	$6\% \le \varepsilon_{\%} < 8\%$ ($1.8 \le \varepsilon_{\%} < 2.4$ s)	$\varepsilon_{\%} \ge 8\%$ (≥2.4 s)

good for some subjects only, but not for all. Therefore, it has been abandoned.

Processing *method B* is providing better results: 100% of the steps are always detected for all subjects and in all trials with excellent reliability. Concerning the assessment of step length S, among the different processing options, only *modes 1*, *2*, and *5* match the excellence for the reliability criterion. *Modes 0* and *3* are good, and *mode 4* is only sufficient. It must be remembered that *modes 1* and *2* are usable if the information about the expected mean step length is given to the model. In these cases, the mean absolute accuracy error of step length is 3.26% and 4.64%, respectively; similar data are shown by distance with errors of 3.25% and 4.62%, respectively. In particular, *mode 1* always underestimates the step length and the traveled distance; if this fact will be confirmed over more large population and constrains, a correction factor (weighting coefficient or offset) could be introduced to increase accuracy. This is an input for future work. Only *mode 5* is subject-independent because it uses a fixed amplitude threshold: in this case, the mean absolute accuracy percentage error for step length and total distance rises to the value of 4.85% and 4.95%, respectively. In some subjects, this value reaches to 8.83%, which is only good for reliability. The method based on anthropometric step length S_{anth} has excellent reliability with a mean absolute accuracy percentage error of 3.70% and 3.68% for step length and distance, respectively.

Generally, the proposed model demonstrates excellent results for *mode 1*, *mode 2*, *mode 5*, and *anthropometric* and good results for *mode 2* and *mode 3*, but only sufficient results for *mode 4*. Frequency of steps and stride time also demonstrates excellent results when *method B* is used as shown in Table 6. This allows us to choose the processing *method B* as the best approach while further investigation with a larger population will support the identification of the most reliable mode among the best performing ones.

Concerning the geometrical parameters of base of support (BOS), when the subject walks, generally the step width LW during walking is lower than the step width in standing LS. As the velocity increases, LW decreases. In standing, the ratio LR = LW/LS is equal to 1. When the subject walks, LR is lower than 1 and as the velocity increases, it is reduced concurrently. The results of the proposed method are coherent with this description. Again, the method is reliable in gait description for AW, AS, and their ratio: when the subject walks, generally the base of support AW during walking is bigger than the base of support AS in standing. As the velocity increases, the step width decreases as well as the AW. In standing, the ratio AR = AW/AS is equal to 1. When the subject walks, AR is generally bigger than 1 and as the velocity increases, it is reduced accordingly. These indexes (Table 7–10) will support the future study of balance also for pathological subjects and represent a first normality dataset for reference.

In conclusion, a theoretical biomechanical model to describe human walking and balance has been presented and a methodological approach to test the accuracy of the proposed model has been adopted. Results show that *method B* is more general, better performing, and more flexible than *method A*; therefore, it has been chosen for future use.

The next step of the study will include the comparison with the gold standard reference of optoelectronic gait analysis to complete the validation of the method.

In the present study, only the COM vertical oscillation is considered (sagittal plane), but the same methodology can be applied to COM mediolateral oscillation.

With the progressive use of the new miniaturized systems, this study supports the belief that wearable devices are reliable for the ambulatory, long-term, and ecologic kinematic gait analysis. The expectation is to develop a dedicated tool for supporting diagnosis and rehabilitation in the hospital and/or at home for elderly and frail subjects. Use for sport people (amateur and professionals) is also exploitable.

Conflicts of Interest

The authors declare that there is no conflict of interest regarding the publication of this paper.

Supplementary Materials

The following tables present data from the population for eleven subjects processed, respectively, with all the proposed algorithms (*method A*, *method B—mode 0/1/2/3/4/5/anthropometric*). Table S1: accuracy assessment of step length and distance. *Method A*. Table S2: accuracy assessment of step length and distance. *Method B—mode 0*. Table S3: accuracy assessment of step length and distance. *Method B—mode 1*. Table S4: accuracy assessment of step length and distance. *Method B—mode 2*. Table S5: accuracy assessment of step length and distance. *Method B—mode 3*. Table S6: accuracy assessment of step length and distance. *Method B—mode 4*. Table S7: accuracy assessment of step length and distance. *Method B—mode 5*. Table S8: accuracy assessment of step length and distance. *Method B—mode anthropometric*. Table S9: walking time accuracy assessment. Table S10: data of walking time (T) and velocity V from all the processing methods (A, B0, B1, B2, B3, B4, B5, and BAnthropometric). Table S11: data of frequency of steps Fstep and frequency of strides Fstride for *methods A* and *B* with respect to reference values. Table 12: data of relative percentage error of gait speed for all the processing methods (A, B0, B1, B2, B3, B4, B5, and BAnthropometric). Table 13: data of absolute percentage error of gait speed for all the processing methods (A, B0, B1, B2, B3, B4, B5, and BAnthropometric). Table 14: data of base of support area in standing (AS (m^2)) and base of support area while walking (AW (m^2)) obtained from data processing with *method B* for the processing methods B0, B1, B2, B3, B4, and B5. Table 15: data of AR = AW/AS obtained from data processing with *method B* for the processing methods B0, B1, B2, B3, B4, and B5. Table 16: data of step width in standing (LS (cm)) and step width while walking (LW (cm)) obtained from data processing with *method B* for the processing methods B0, B1, B2, B3, B4, and B5. Table 17: data of LR = LW/LS obtained from data processing with *method B* for the processing methods B0, B1, B2, B3, B4, and B5. *(Supplementary Materials)*

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