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Modified Pendulum Model for Mean Step Length Estimation

Rafael C. González, *Member, IEEE*, Diego Alvarez, *Member, IEEE*, Antonio M. López, *Member, IEEE*, and Juan C. Alvarez, *Member, IEEE*

Abstract— **Step length estimation is an important issue in areas such as gait analysis, sport training or pedestrian localization. It has been shown that the mean step length can be computed by means of a triaxial accelerometer placed near the center of gravity of the human body. Estimations based on the inverted pendulum model are prone to underestimate the step length, and must be corrected by calibration. In this paper we present a modified pendulum model in which all the parameters correspond to anthropometric data of the individual. The method has been tested with a set of volunteers, both males and females. Experimental results show that this method provides an unbiased estimation of the actual displacement with a standard deviation lower than 2.1%.**

I. Introduction

STEP length estimation is an important issue in areas such as gait analysis, sport training or pedestrian localization. In some applications, it is necessary to obtain that estimation with a wearable system. A good approach is the use of accelerometers attached to the subject body near the body center of gravity (COG). A well known method based on this approach consists in the calculation of the vertical excursion of the COG and the application of an inverted pendulum model [1-3] for the estimation of step length. This model provides a mathematical relation between the horizontal and the vertical displacements of the COG. Reported results using this method show that step length is underestimated. To solve the problem, a calibration constant for each subject is calculated [3].

According to Cavagna et al. [4], displacement during single stance, in level forward walking, can be described using an inverted pendulum model. In addition, during double stance phase forward displacement remains approximately constant. Nilsson et al. [5], describes that, in normal walking, the product of speed and double support time is approximately constant.

This paper shows the results achieved using a two phase model for step length estimation. The parameters for the model are obtained from direct anthropometric measures, namely leg and foot length. Following the description given by Cavagna et al. [4], our approach divides each step in two phases: *single-stance* and *double-stance*. In the *single-stance* phase, an inverted pendulum model computes the longitudinal displacement of the COG as a function of the vertical excursion. In the *double-stance* phase, the longitudinal displacement is computed as a percentage of foot size. This assumption is based on studies made by Han et al. [6] and Schimd et al. [7], which describes the movement of the centre of pressure during gait.

Section II of this paper describes the model and the algorithms used to produce the estimations. Experimental setup and results are shown in Section III. These results are discussed in Section IV. It will be shown that this method provides an unbiased estimation of the actual displacement with a standard deviation lower than 2.1%.

II. Methods

A. Model description.

We approximate the step length (*SL*) by the forward displacement of the hip during a step. Following the work by Cavagna et al. [4], this quantity is the sum of the antero-posterior displacement during single stance phase (L_{ss}) and the displacement during double stance (L_{ds}):

$$SL = L_{ss} + L_{ds} \tag{1}$$

1) Forward displacement during single stance

Single stance phase begins with the final-contact (FC) of the contralateral foot and finishes with the initial contact (IC) of the same foot. The body swings forward, using the reference foot as an anchor to the floor. This movement can be modeled by an inverted pendulum. Forward displacement (L_{ss}) can be computed from the vertical excursion of the hip (h), during single stance phase, and the leg length (l):

$$L_{ss} = 2\sqrt{2lh - h^2} \tag{2}$$

Leg length is measured as the distance from the lateral malleolus to the greater trochanter.

2) Forward displacement during double stance

We model the displacement of the hip during double stance (L_{ds}) as the forward displacement of the center of pressure in the foot. Following the results of Han et al. [6], and Schmid et al. [7], the displacement is proportional to the foot length (p):

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All authors are with the Multisensor Systems & Robotics Lab (SiMuR), Electrical Engineering Dept., University of Oviedo, Viesques, Ed. 2, 33204 Gijón, Spain (phone: +34 985 18 25 28, fax: +34 985 18 25 28, e-mail: simur@isa.uniovi.es).

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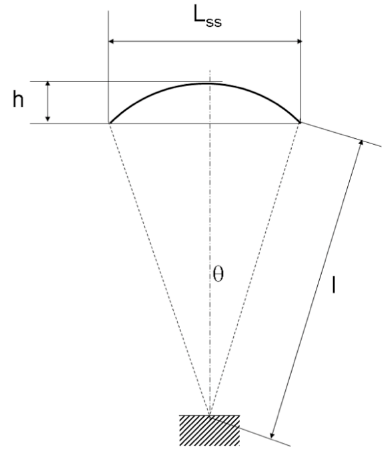

Fig. 1. The forward displacement of the hip during single stance phase (L_{ss}) is modeled as an inverted pendulum. L_{ss} can be determined from the variation in vertical position (h) and the radius of the

$$L_{ds} = K\ p \tag{3}$$

K is the proportionality constant. This value is estimated as 0.83 by Han et al. [6] and 0.67 by Schmid et al. [7]. We will estimate the value of K from a random experiment of each individual.

B. Calculation of h

Calculation of h is done by double integrating the vertical acceleration, measured by an accelerometer placed close to the L3 vertebra. Accelerometer data are prone to drift problems. We make the assumption that the speed and height of the hip in consecutive ICs is the same, provided that the subject is walking over a flat surface and at constant forward speed. As a consequence, the mean value of vertical acceleration, and vertical speed between two successive ICs must be zero. These boundary conditions are used to correct the drift effect in the double integral. The following equations describe the five steps involved in the calculation:

$$a_{zc}(t) = a_z(t) - \frac{\int_{ta}^{tc} a_z(\tau)d\tau}{tc - ta} \tag{4.a}$$

$$v_z(t) = \int_{ta}^{tc} a_{zc}(\tau)d\tau + v_z(0) \tag{4.b}$$

$$v_{zc}(t) = v_z(t) - \frac{\int_{ta}^{tc} v_z(\tau)d\tau}{tc - ta} \tag{4.c}$$

$$z(t) = \int_{ta}^{tc} v_{zc}(\tau)d\tau + z(0) \tag{4.d}$$

$$h = \max_{t\in[tb,tc]}\big(z(t)\big) - \min_{t\in[tb,tc]}\big(z(t)\big) \tag{4.e}$$

Where $a_z(t)$ represents the raw values provided by the accelerometer, $a_{zc}(t)$ is the corrected acceleration value. In the same way, $v_z(t)$ and $v_{zc}(t)$ represents the raw and corrected values of vertical speed respectively. The vertical displacement is given by $z(t)$. All these calculations are made between two successive ICs, i.e. in the interval [t_a, t_c] (see

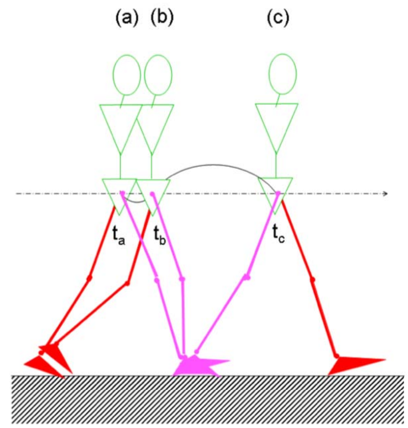

Fig. 2. A single step begins with the initial contact (IC) of the reference leg (t_a). In the time t_b, the contralateral foot leaves the ground (FC), being the beginning of the single stance phase. This phase finishes when the contralateral foot hits the ground again (t_c). Double support phase goes from t_a to t_b.

figure 2). Equation 4.e shows that h is the difference between the maximum and the minimum of $z(t)$ between the final contact of the contralateral foot and the next initial contact, i.e. in the interval [t_b, t_c].

C. Event Detection

This method requires that IC and FC can be detected for each step. Different authors have characterized the events which divides the different phases of human gait [1, 8-14].

Event detection is based on the descriptions given by Zijlstra [1] and Auvinet [9]. We extract the principal harmonic of the vertical acceleration using a zero-lag, low-pass filter at 2.5 Hz. Then IC are identified as the maxima of the anteroposterior acceleration which precedes a maximum of the principal harmonic. Final contacts are located as local

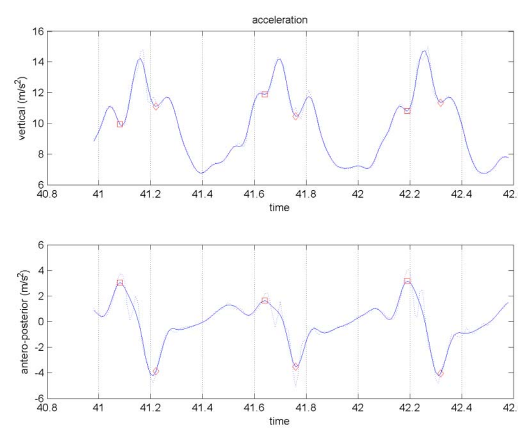

Fig. 3. Initial and Final Contact points are located using the acceleration data. Initial contacts correspond to the maximum of the antero-posterior acceleration that precedes a maximum in the principal harmonic of the vertical acceleration. The final contact is the local minimum, in vertical acceleration, after a maximum of the principal harmonic of the vertical acceleration.

minima in a small neighborhood after each maximum of the principal harmonic.

We approximate the step duration by computing the position of the first peak in the unbiased autocorrelation of the anteroposterior acceleration. Using this frequency as the first harmonic of the signal, it is low-pass filtered to remove frequencies above the fifth harmonic, by means of a fourth order, zero-lag low-pass Butterworh filter. Initial contact points are the points that are a local maximum in a neighborhood of half the step duration at both sides of the candidate point. Final contact points are located as a local minimum in the 50% of the step following the initial contact.

III. Experiments and Results

A. Experimental Setup

Tests have been done in a long corridor placed in the Electrical Engineering Department of the Oviedo University.

Measurements were taken from a group of 16 adults (9 men and 7 women). Height ranges from 1.58 to 1.86m while age ranges from 21 to 44 years. No subject reported any kind of impairment of the locomotion system that could affect the experiments. Also, there were no footwear requirements. All subjects gave their signed consent.

During test procedures, subjects were asked to walk 25 meters following a straight path along an interior corridor. The first and the last 2.5 meters of walking data were discarded for the analysis, since gait is not stable during the initial and final phases, so that 20m of valid data were recorded for each path.

Each subject completed 12 independent excursions (this adds up a total of 240m. per subject). They were asked to maintain a constant velocity for each walk: *preferred* in the first 4 excursions, *fast* in the next 4 ones and s*low* in the last 4 ones. The *preferred*, *fast* and s*low* velocities were subjectively selected by each person. Subjects were allowed to turn freely between each excursion, and to walk freely or to rest between different tests.

Two camcorders were placed at the beginning and at the end of the 20m segment. The IC event of both the first and last step was visually identified in the records, with a maximum error of 2 video frames (0.08s). Floor marks were placed each centimeter, and the actual distance was visually measured for each excursion. Total time was also measured, as the difference between these two events. Velocity for each excursion is computed, as the relation between the total length and the excursion duration.

Acceleration data were gathered by means of a triaxial accelerometer placed close to the L3 vertebral position, accepted as a fine approximation of the COG position during normal walking. The device is fixed to the lower lumbar spine by means of an adjustable corset to avoid movement artifacts.

The accelerometer is an XSens MTx sensor. Measurement range is ±2g, being g the gravity acceleration. Data were gathered at 100Hz. by using a 12bit A/D conversion. MTx sensor is wired to the XSens Xbus Master, placed in the subject's belt. All data are stored in a PC, where the Xbus Master device transfers them in real time by means of a Bluetooth communication link for further processing.

Synchronization between the camcorders and the accelerometer was done inserting a special mark in both systems. Final synchronization error was less than a video frame (0.04s).

B. Experimental Results

Total time needed for each excursion varied from 10.36s to 23.16s as a function of the selected velocities. Actual distance for each path, as measured with the camcorders, varied from 19.28 m to 20.47m. The maximum velocity was 1.96 m/s, while the minimum one was 0.86 m/s. Labeled s*low, preferred* or *fast* velocities were different for each individual, as each subject freely chose them. Even so, for every subject all experiments at *fast* velocity were done faster than at *preferred* velocity, and the same happens for the *preferred* and s*low* velocities.

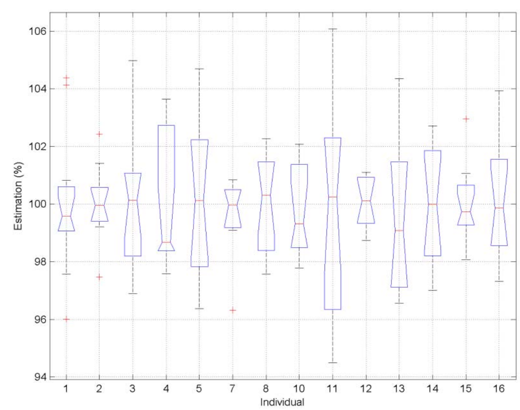

Fig. 4. The figure shows the distribution of distance estimation for each individual. Each boxplot represents the twelve results obtained for each individual.

Event detection has been evaluated by visual inspection of the results. Two individuals (numbered 6 and 9) have been discarded because their acceleration patterns were too deformed from the expected waveform that detection was too faulty. In addition, subject 7 made some small jumps at the beginning of the fast excursions, so those data were also discarded. The accuracy of the software reached a 99.6% of correct event detection. The mean value for K was 74.06% (±13.63).

In order to facilitate comparison, all estimations were refereed to the distance traveled in each excursion; therefore a score of 100% means an exact estimation.

The mean estimation was 99.99% (±2.09), considering all the experiments together. Estimations ranged from 94.50% to 106.07%. Figure 4 shows the distribution of the estimations for each individual. A two way ANOVA shows that differences among different individuals are not significant ($p=0.99$). On the other hand, the estimation depends on forward speed ($p<0.01$), it can be seen that for higher velocities, the proposed model has a trend to underestimate the mean step length.

IV. Discussion

We have presented a model to estimate the mean step length based on the inverted pendulum model. We apply this model during the single stance phase, while double stance phase is modeled by a constant value, proportional to foot length. The main idea behind this model is that the forward displacement during double stance is related to the displacement of the centre of pressure in the foot. The proportionality constant was determined for each individual from a randomly selected experiment. Calculated values were inside the limits found in the literature.

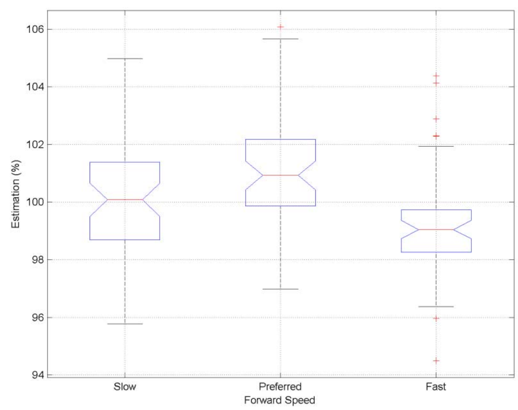

Fig. 5. Estimations grouped by velocities. It can be seen that at high speeds the method presents a tendency to underestimate the mean step length..

Event detection plays a main role in this model, as it is necessary to distinguish between single and double stance phases. The algorithm used to detect events presented a 99.6% of correct detections when acceleration signals correspond to the expected waveform. Anyway, signal variability is very large and further work should be done.

Mean estimation is unbiased and has a standard deviation lower than 2.1% of the expected value. Due to the variability in the proportion of the foot traveled by the center of pressure, it would be interesting to analyze if this quantity can be obtained from anatomical points of the foot.

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