

# Ambulatory Estimation of Center of Mass Displacement During Walking

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**Abstract**—The center of mass (CoM) and the center of pressure (CoP) are two variables that are crucial in assessing energy expenditure and stability of human walking. The purpose of this study is to estimate the CoM displacement continuously using an ambulatory measurement system. The measurement system consists of instrumented shoes with 6 DOF force/moment sensors beneath the heels and the forefeet. Moreover, two inertial sensors are rigidly attached to the force/moment sensors for the estimation of position and orientation. The estimation of CoM displacement is achieved by fusing low-pass filtered CoP data with high-pass filtered double integrated CoM acceleration, both estimated using the instrumented shoes. Optimal cutoff frequencies for the low-pass and high-pass filters appeared to be 0.2 Hz for the horizontal direction and 0.5 Hz for the vertical direction. The CoM estimation using this ambulatory measurement system was compared to CoM estimation using an optical reference system based on the segmental kinematics method. The rms difference of each component of the CoM displacement averaged over a hundred trials obtained from seven stroke patients was  $(0.020 \pm 0.007)$  m (mean  $\pm$  standard deviation) for the forward  $x$ -direction,  $(0.013 \pm 0.005)$  m for the lateral  $y$ -direction, and  $(0.007 \pm 0.001)$  m for the upward  $z$ -direction. Based on the results presented in this study, it is concluded that the instrumented shoe concept allows accurate and continuous estimation of CoM displacement under ambulatory conditions.

**Index Terms**—Acceleration, ambulatory, balance, center of mass (CoM), center of pressure (CoP), gait analysis, human walking.

## I. INTRODUCTION

**B**IOMECHANICAL analysis of human movement is used by many different disciplines to understand the mechanisms underlying human movement, quantify performance, or guide rehabilitation interventions. A significant contribution is provided by means of gait analysis. The easiest and most common model used to describe human walking is the inverted pendulum model [1], [2]. In this model, human walking is characterized by two variables: the center of mass (CoM) and the center

of pressure (CoP). The CoM is an imaginary point at which the total body mass can be assumed to be concentrated. The CoP is the application point of the ground reaction force (GRF), the point on the contact surface between body and ground where the moments about the horizontal axes are zero. In the inverted pendulum model, the CoM is balancing on a rod with the origin at the CoP. The double support phase is viewed as a transition from one inverted pendulum to the next.

Several methods exist for the estimation of CoM movement that differ in the underlying assumptions. The segmental kinematics method [3]–[5] is based on the definition of the CoM and models the body as a kinematic chain of rigid segments. By measuring the position and orientation of each segment and approximating the mass fractions as well as the locations of the CoM of each segment, an estimation of the overall CoM movement is obtained. Another method is the double integration of GRF method based on Newton's second law [6], [7], which states that the net external force acting upon a body is equal to its mass multiplied by its acceleration. Since the external force and body mass can be measured accurately by a force plate, an estimation of the acceleration of the CoM is obtained. The displacement of the CoM can be calculated through double integration of acceleration after subtracting gravitational acceleration, with proper consideration of initial constants of integration, i.e., initial velocity and position [8], [9]. A third method, the CoP low-pass filter method, uses the knowledge that CoP motion has higher frequency content than CoM motion [10], [11]. Moreover, the relative magnitude of the CoM movement with respect to the CoP movement can be described by a mathematical relation in the frequency domain [12], assuming that the movement is periodic and that the body can be modeled as an oscillating inverted pendulum. A fourth method is to assume that CoM movement can be approximated by movement of the pelvis [13]. This simple method assumes that the CoM is stationary within the pelvis, which may be a reasonable assumption in normal gait, but becomes inaccurate when relative movement of body segments increases, since the location of the CoM is affected by movement of individual segments. These methods have been extensively compared by many researchers [14]–[20]. Summarizing, it can be concluded that, in general, the performance of the segmental kinematics method is similar to that of the double integration of GRF method and superior to the CoP low-pass filter method and pelvic movement method.

Although widely accepted, the aforementioned methods have their limitations as well. The performance of the segmental kinematics method depends heavily on the accuracy of the approximated segmental mass fractions and segment CoM

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Fig. 1. Picture of the instrumented shoe with force/moment sensors beneath the heel and forefoot and inertial sensors rigidly attached to the force/moment sensors.

locations. Moreover, it requires a precise marker or sensor placement for accurate position and orientation determination. The performance of the double integration of GRF method is limited by the accuracy of the determined initial constants of integration. A joint drawback of all described methods is the restriction to a laboratory environment and the accompanying inability for continuous measurement of the CoM trajectory. The segmental kinematics method and pelvic movement method are limited by the measurement volume of the optical measurement system, whereas the double integration of GRF method and CoP low-pass filter method are limited by the number of force plates and the occurrence of a correct hit.

It is desirable to estimate the CoM trajectory continuously during walking, since it contains important information about dynamic balance control, and especially for pathological gait, the CoM movement can vary from step to step. Moreover, the measurement of the CoM should not be restricted to a laboratory environment, but should be possible during everyday life as well. The objective of this study is to determine the CoM trajectory continuously during walking, based on a combination of methods described earlier, using an ambulatory measurement system. The proposed method fuses CoP trajectory with double integrated CoM acceleration, such that the best characteristics of each constituent are used. The measurement system is similar to a previously proposed measurement system [21]. For validation, data from seven stroke patients were acquired and the estimated CoM trajectories were compared to the trajectories obtained by the segmental kinematics method using an optical position measurement system as a reference.

## II. METHODS

This study proposes a method to estimate the CoM trajectory continuously during walking using an ambulatory measurement system. This section first describes the methods used to obtain an estimation of the CoM displacement, followed by a description of the experimental methods.

### A. Estimation of CoM Displacement

The trajectory of the CoM is estimated by fusing CoP with double integrated CoM acceleration, derived from the GRF measured by the instrumented shoes. The instrumented shoe was introduced previously [21], [22] and is shown in Fig. 1. A

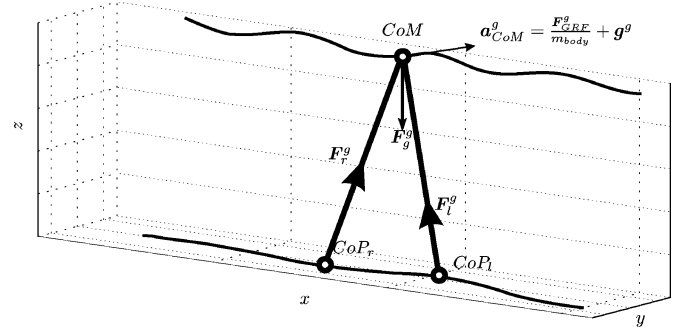


Fig. 2. Schematic drawing of a subject during walking, where the trajectories of the CoP and CoM are indicated by the solid lines. The small circles denote the current position of the CoM and the CoP of the left foot (CoPl) and right foot (CoPr). The GRFs of the left and right foot are denoted by  $F_l^g$  and  $F_r^g$ , respectively, and the gravitational force by  $F_g^g = -m_{body} \cdot g^g$ , where  $m_{body}$  denotes the body mass and  $g^g$  the gravitational acceleration. Newton's second law relates the sum of GRFs of the left and right foot  $F_{GRF}^g$  to the acceleration of the CoM  $a_{CoM}^g$  as stated in (5).

schematic drawing of a subject during walking with the most important variables is depicted in Fig. 2. The CoP position of a single step is derived from signals measured by the force/moment sensors, as described in Section II-A2. To obtain the CoP trajectory during several steps, foot position and orientation need to be estimated, which is described in Section II-A1. Fusion of CoP trajectory with double integrated CoM acceleration (Section II-A3) is based on a frequency domain-method, as described in Section II-A4.

1) *Estimation of Foot Position and Orientation:* An estimation of foot position and orientation is obtained by combining the signals of an inertial sensor consisting of three accelerometers and three gyroscopes. The accelerometers measure the sum of sensor acceleration  $a^s$  and gravitational acceleration  $g^s$  expressed in the sensor frame  $\Psi_s$ :  $s^s = a^s - g^s$ . The gyroscopes measure the angular velocity  $\omega_s^{s,g}$  of the sensor frame  $\Psi_s$  with respect to the global frame  $\Psi_g$ , expressed in the sensor frame  $\Psi_s$ . The coordinate axes of the global frame are defined by positive  $x$  in the direction of gait, positive  $z$  directed upward, and positive  $y$  perpendicular to the  $x$ - and  $z$ -direction such that the result is an orthogonal right-handed coordinate system. It is desired to express all signals in the global coordinate frame  $\Psi_g$ , which means the rotation matrix  $R_s^g$  needs to be known. This matrix is obtained by solving the differential equation [23]

$$\dot{R}_s^g = R_s^g \tilde{\omega}_s^{s,g} \quad (1)$$

where  $\tilde{\omega}_s^{s,g}$  is the skew-symmetric matrix of the angular velocity measured by the gyroscopes  $\omega_s^{s,g}$

$$\tilde{\omega} = \begin{pmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{pmatrix}. \quad (2)$$

The estimation of position is dependent on the phase of the gait cycle and is restarted each step, where the start and end of a step are determined using the magnitude of the force measured by the force/moment sensors. During the stance phase, when the foot touches the ground, the position of the foot is fixed.

During the swing phase, the position of the foot is estimated by double integration of the acceleration expressed in the global coordinate frame after subtraction of the gravitational acceleration. The estimation of position and orientation by integration is prone to integration drift caused by noise and fluctuating offsets. This drift is reduced by using zero velocity updates and the knowledge that the vertical position of the foot during midstance is equal at each step [21]. To obtain the position and orientation during several steps, the estimated end position and orientation of the previous step are used as an initial value for the current step. To remove the error caused by an unknown initial orientation, it is assumed that both feet move in the same direction on average and that the distance between the feet is equal at the start and end of a measurement [24].

2) *Estimation of CoP Trajectory*: In a previous study [21], the CoP was estimated using instrumented shoes. The CoP  $\mathbf{x}_{\text{CoP},i}^g$  below each foot  $i$  was calculated by

$$\mathbf{x}_{\text{CoP},i}^g = \begin{pmatrix} -\frac{M_{y,i}^g}{F_{z,i}^g} \\ \frac{M_{x,i}^g}{F_{z,i}^g} \\ 0 \end{pmatrix} \quad (3)$$

where  $F_{z,i}^g$  denotes the vertical component of the GRF measured by force transducers under the shoe expressed in the global coordinate frame, and  $M_{x,i}^g$  and  $M_{y,i}^g$  denote the measured moments about the horizontal axes. To obtain the CoP of a foot for several steps, the estimated foot position is used as described in Section II-A1. The contributions of both feet should be added to obtain the complete CoP trajectory. The addition is achieved by weighting the CoP trajectories of each foot by the relative magnitude of the GRF of that foot

$$\mathbf{x}_{\text{CoP}}^g = \frac{\|\mathbf{F}_l^g\|}{\|\mathbf{F}_l^g + \mathbf{F}_r^g\|} \mathbf{x}_{\text{CoP},l}^g + \frac{\|\mathbf{F}_r^g\|}{\|\mathbf{F}_l^g + \mathbf{F}_r^g\|} \mathbf{x}_{\text{CoP},r}^g \quad (4)$$

where  $\mathbf{F}_l^g$  and  $\mathbf{F}_r^g$  denote the measured GRF of the left and right foot, respectively.

3) *Estimation of Double Integrated CoM Acceleration*: In order to estimate the CoM acceleration, Newton's second law was used, which states that the net external force acting upon a body is equal to its mass multiplied by its acceleration ( $\mathbf{F} = m \cdot \mathbf{a}$ ). Since all body mass can be assumed to be lumped at the CoM and since the net external force is measured continuously by the instrumented shoes, an estimation of the acceleration expressed in the global coordinate frame is obtained. This acceleration vector consists of CoM acceleration  $\mathbf{a}_{\text{CoM}}^g$  and gravitational acceleration  $\mathbf{g}^g$ , which means the gravitational acceleration needs to be removed (Fig. 2)

$$\mathbf{a}_{\text{CoM}}^g = \frac{\mathbf{F}_{\text{GRF}}^g}{m_{\text{body}}} + \mathbf{g}^g \quad (5)$$

where  $\mathbf{F}_{\text{GRF}}^g = \mathbf{F}_l^g + \mathbf{F}_r^g$  denotes the total GRF measured by the instrumented shoes expressed in the global coordinate frame and  $m_{\text{body}}$  denotes the body mass of the subject. The estimated

acceleration is double integrated to obtain the position  $\mathbf{x}_{\text{CoM}}^g$

$$\begin{aligned} \mathbf{v}_{\text{CoM}}^g(t) &= \mathbf{v}_0 + \int_{t_0}^t \mathbf{a}_{\text{CoM}}^g(\tau) d\tau \\ \mathbf{x}_{\text{CoM}}^g(t) &= \mathbf{x}_0 + \int_{t_0}^t \mathbf{v}_{\text{CoM}}^g(\tau) d\tau \end{aligned} \quad (6)$$

where  $\mathbf{v}_0$  and  $\mathbf{x}_0$  denote the initial velocity and position, respectively.

4) *Fusion*: CoM displacement  $\mathbf{x}_{\text{CoM}}^g$  is obtained by fusing CoP trajectory [(3) and (4)] with double integrated CoM acceleration [(5) and (6)]. In essence, fusion is necessary since double integration of acceleration introduces drift, which is removed by a high-pass filter. This means, however, that the low-frequency component of the CoM trajectory is removed as well. To reinclude this component, the knowledge that the CoP varies about the vertical projection of the CoM and coincides with this projection on average is used. Fusion consists of addition of the high-pass filtered, double integrated acceleration with the low-pass filtered CoP trajectory obtained from the instrumented shoes. To assure the result is indeed the CoM displacement, the order and cutoff frequency of the low-pass and high-pass filters were identical. It should be noted that the vertical component of the CoP is always zero, since the CoP is defined on the contact surface between the body and ground. This means the low-pass filtered CoP trajectory is merely used to estimate the horizontal CoM displacement. Moreover, since the average height of the CoM is removed by the high-pass filter, the average height of the sacrum is added to the vertical component of the CoM displacement.

## B. Experimental Methods

To compare the accuracy of the ambulatory measurement system with a conventional measurement system, experiments were performed in a gait laboratory, where an optical tracking system (Vicon Oxford Metrics, Oxford, U.K.) was available. Seven stroke patients participated in this study. Their age was ( $58.3 \pm 8.6$ ) years (mean  $\pm$  standard deviation), their length ( $1.80 \pm 0.07$ ) m, and their body mass ( $86.6 \pm 12.2$ ) kg. Informed consent was obtained from each subject prior to the experiment, and the study was approved by the local ethical committee. Subjects wearing instrumented shoes were instructed to walk repeatedly through the laboratory between predefined start and end points. To assure an equal distance between the feet at the beginning of each measurement, subjects were asked to position the feet against a mold attached to the floor between the feet before each walking trial. Body movement and GRF were measured by the instrumented shoes, while the optical tracking system was used as a reference. The instrumented shoes [21] consisted of standard orthopaedic sandals equipped with two 6 DOF force/moment sensors (ATI-Mini45-SI-580-20, supplier: Schunk, Arnhem, NL) beneath the heel and forefoot, as shown in Fig. 1. Each force sensor had a miniature inertial sensor (Xsens Motion Technologies, Enschede, The Netherlands) rigidly attached to it for the estimation of position and orientation. For the reference system, markers were placed on both legs (toe,

heel, lateral malleolus, shank, knee, and thigh), both arms (upper arm, elbow, head of ulna, and styloid process of radius), left and right anterior superior iliac spine, left and right shoulder, and a three-cluster marker on the sacrum.

Data from the inertial sensors, the analogue data from the force sensors beneath the shoe, and the 3-D marker data were sampled at 50 Hz. All data were low-pass filtered by applying a second-order recursive Butterworth filter at a cutoff frequency of 15 Hz. The voltages from the force and moment sensors were converted to forces and moments by applying calibration matrices obtained from static calibration measurements. The synchronization between the inertial sensor system and Vicon was done by maximizing the correlation between pulses generated by a pulse generator that were sent to both measurement systems. Possible gaps in the Vicon data were spline interpolated prior to filtering with a maximum gap size of 15 samples.

Before fusing CoP trajectory and double integrated CoM acceleration, a low-pass filter was applied to the first component and a high-pass filter to the second component. Both filters were first-order recursive Butterworth filters.

The CoM displacement using the reference system was determined by the segmental kinematics method based on the model of Koopman *et al.* [25]. Measured marker positions were used to determine body segmental mass fractions and segment CoM locations with the regression equations of Chandler *et al.* [26]. The total body CoM is calculated as the weighted sum of the CoM of every segment [5]

$$\mathbf{x}_{\text{CoM}} = \frac{\sum_i m_i \cdot \mathbf{x}_{\text{CoM},i}}{\sum_i m_i} \quad (7)$$

where  $m_i$  is the mass of segment  $i$  and  $\mathbf{x}_{\text{CoM},i}$  its CoM location.

### III. RESULTS

Data obtained from seven stroke patients were analyzed in this study, which resulted in a total number of 100 trials. To determine the optimal cutoff frequencies used to filter CoP and double integrated acceleration data, rms differences between the CoM estimates of the ambulatory and reference systems were calculated while varying the cutoff frequencies between 0.1 and 1 Hz. The cutoff frequency at the smallest rms difference was considered to be optimal. First, the low-pass filtered CoP trajectory was evaluated separately to compare against subsequent fusion with the double integrated CoM acceleration. The results for the  $x$ - and  $y$ -direction averaged over all analyzed trials are shown in Fig. 3(a). The optimal cutoff frequencies were determined to be 0.2 Hz for the  $x$ -direction and 0.4 Hz for the  $y$ -direction. It should be noted that the CoP trajectory does not provide information about the CoM movement in  $z$ -direction. Second, the optimal cutoff frequencies for the  $x$ -,  $y$ -, and  $z$ -direction using the method combining CoP and double integrated CoM acceleration data were determined and shown in Fig. 3(b). The rms differences are smaller than in the case of CoP only, which demonstrates the need to include the high-frequency component for CoM displacement estimation. The results indicate that the optimal cutoff frequency is 0.2 Hz for the  $x$ - and  $y$ -direction and 0.5 Hz for the  $z$ -direction.

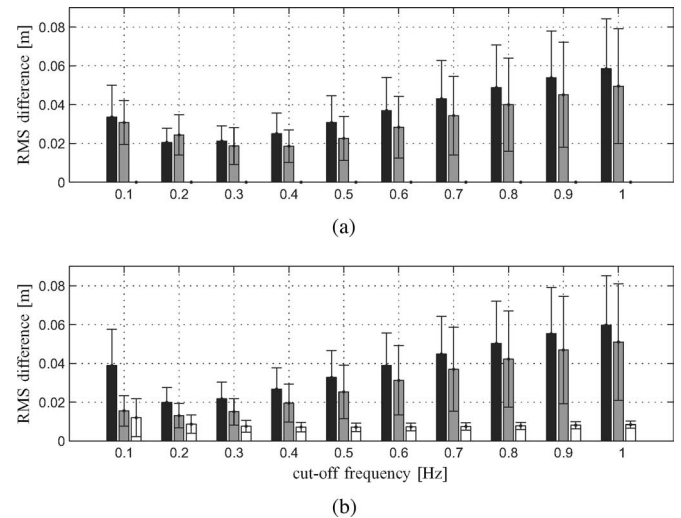


Fig. 3. RMS differences (mean and standard deviation averaged over all trials) of CoM estimates between ambulatory and reference systems with varying cutoff frequency ( $x$ : black;  $y$ : gray;  $z$ : white). (a) Only low-frequency CoP component ( $z$ -direction not applicable). (b) After fusing low-frequency (CoP) and high-frequency (double integrated CoM acceleration) components.

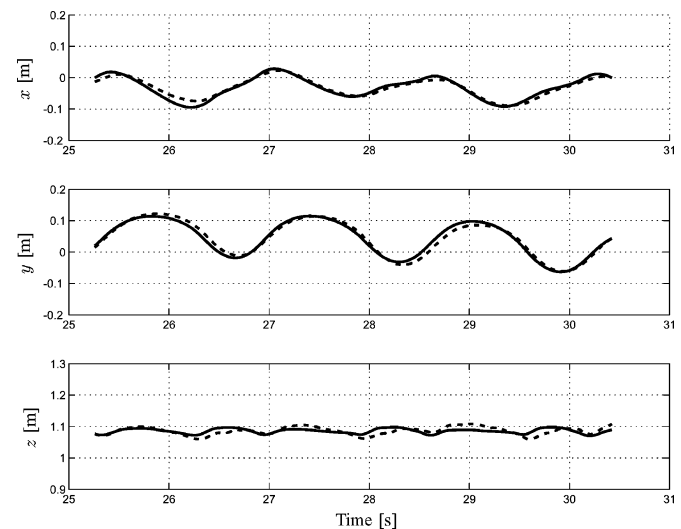


Fig. 4. CoM as a function of time for a representative trial estimated by ambulatory (solid) and reference (dashed) systems ( $x$ : upper;  $y$ : middle;  $z$ : lower). It should be noted that the integrated mean velocity has been subtracted from the  $x$  component.

The CoM displacement of a representative trial using the previously calculated cutoff frequencies is shown in Fig. 4. For the  $x$  and  $y$  components, the mean value of the CoM displacement determined by each measurement system was subtracted to remove the static position error caused by a separate origin for both measurement systems. Moreover, the integrated mean velocity has been subtracted from the  $x$  component to visualize differences between both measurement systems in that direction.

An overview of the relative rms differences between the ambulatory and reference systems for the seven analyzed stroke patients is shown in Fig. 5. The rms differences for each component of the CoM displacement averaged over all subjects were

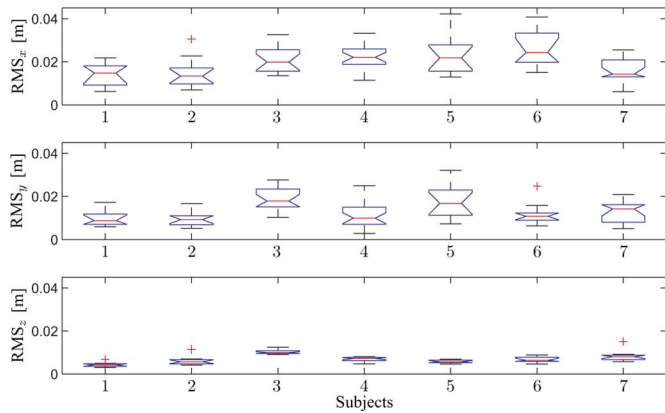


Fig. 5. Box and whisker plot of rms differences of CoM estimates using ambulatory and reference systems for each direction and all analyzed subjects ( $x$ : upper;  $y$ : middle;  $z$ : lower). The box has lines at the lower quartile, median, and upper quartile values. The whiskers are the lines showing the extent of the rest of the data. Outliers are indicated by crosses.

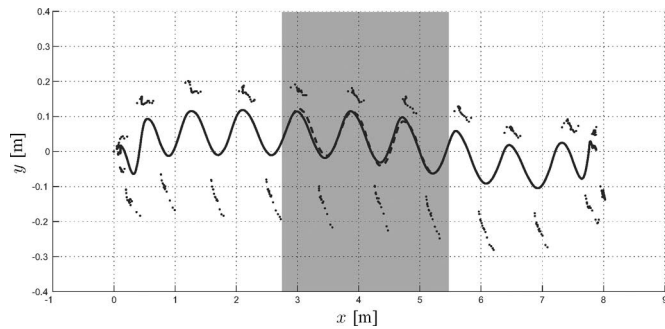


Fig. 6. Top view of the CoM of a walking stroke patient estimated by ambulatory (solid) and reference (dashed) systems. The data used to construct this figure are equal to the data used for Fig. 4. On either side of the CoM, the CoP assessed from instrumented shoe data is indicated by the dotted lines, where each dot represents a time sample. The gray area indicates the Vicon measurement volume. For display purposes, the aspect ratio of the axes has not been equalized.

( $0.020 \pm 0.007$ ) m (mean  $\pm$  standard deviation) for the forward  $x$ -direction, ( $0.013 \pm 0.005$ ) m for the lateral  $y$ -direction, and ( $0.007 \pm 0.001$ ) m for the upward  $z$ -direction. The rms distance between both CoM estimates was ( $0.025 \pm 0.007$ ) m. The rms differences as a percentage of the height of the subjects were ( $1.1 \pm 0.4$ )% for the  $x$ -direction, ( $0.7 \pm 0.3$ )% for the  $y$ -direction, and ( $0.4 \pm 0.1$ )% for the  $z$ -direction.

An indication of the potential of the ambulatory measurement system to quantify pathological gait disorders is shown in Fig. 6. The asymmetric walking pattern of a stroke subject can be clearly seen in the figure. As illustrated, the foot of the paretic leg is rotated outward, and the position of the CoM is toward the nonparetic leg most of the time. Moreover, the figure clearly shows that the CoM displacement estimation by the reference system is limited to a few steps due to a restricted measurement volume, while the ambulatory system estimates the CoM displacement continuously.

## IV. DISCUSSION

This study proposes a method to assess the CoM displacement continuously during walking using an ambulatory system. The method is based on a combination of the CoP low-pass filter method and the double integration of GRF method. The ambulatory system was compared to a reference measurement system based on the segmental kinematics method. The results were promising and comparable to those described in literature [17]. In that study, rms differences between CoM estimates as a percentage of the height of the subjects between the segmental kinematics method and the double integration of GRF method were ( $0.4 \pm 0.3$ )% for the  $x$ -direction, ( $0.4 \pm 0.2$ )% for the  $y$ -direction, and ( $0.9 \pm 0.7$ )% for the  $z$ -direction. Compared to the differences found in this study, their differences were smaller for the  $x$ -direction, comparable for the  $y$ -direction, and larger for the  $z$ -direction.

Although the segmental kinematics method has been widely used and is considered to be the standard, its quality depends heavily on the accuracy of the approximated segmental mass fractions, segment lengths, and CoM locations [20]. It is therefore questionable which of the two methods represents the true CoM displacement most accurately. Another option would have been to compare the ambulatory measurement system to the double integration of GRF method using force plates, which is widely used and accepted to be accurate as well. This is similar to a comparison of the GRF determined with both measurement systems that has been done in a previous study [21]. The signals agreed well with an rms difference of ( $0.012 \pm 0.001$ ) N/N, being ( $1.1 \pm 0.1$ )% of the maximal GRF magnitude. Every method will have its drawbacks and inaccuracies, since the CoM is an imaginary point at which the total body mass can be assumed to be concentrated, and thus cannot be measured exactly. Still, as discussed at the starting of this section, the relative rms differences acquired in this study with respect to the amplitude of the CoM movement are comparable to those described in literature [17], while the described system allows measurements in an ambulatory environment.

Continuous estimation of CoM displacement, using the ambulatory system described in this study, has several advantages compared to existing measurement systems. Compared to the reference segmental kinematics method, which requires a full-body marker configuration, the described measurement system allows estimation of CoM movement merely wearing the instrumented shoes (Fig. 1). Moreover, the measurements are not restricted to a laboratory environment or a limited measurement volume. As indicated by Fig. 6, pathological gait that can have a rather variable and asymmetrical walking pattern can still be estimated. Besides, the behavior at the start and end of a trial is incomparable to the behavior during the remainder of a trial, but interesting for evaluating motor control. Especially for pathological gait, continuous estimation of CoM displacement is important as it can be used to monitor improvement, efficiency of walking, and balance control.

Although the results presented in this study are promising, there is still room for improvements. First, estimation of position and orientation by integration introduces drift, which will

increase for longer time durations. However, walking is a cyclical movement, which means initial and final conditions can be applied every step, i.e., zero velocity when the foot is flat on the floor, with constant vertical position of the foot at mid-stance during level walking. Moreover, for the determination of gait parameters and biomechanical analysis, relative positions instead of absolute positions in space are important. The integration drift in these relative positions can be limited by the use of human body constraints. Also, to limit the integration drift in the estimation of the lateral distance between the feet and as an alternative for the mold between the feet used to estimate this distance initially, Newton's second law for rotational motion can be used. It states that the net external moment  $\mathbf{M}$  acting upon a body is equal to its moment of inertia  $\mathbf{I}$  times its angular acceleration  $\alpha$ : ( $\mathbf{M} = \mathbf{I} \cdot \alpha$ ). The net external moment is calculated by the vector product of position and force ( $\mathbf{M} = \mathbf{r} \times \mathbf{F}$ ). If the product of angular acceleration and moment of inertia is assumed to be small during walking, it follows that  $\mathbf{r} \times \mathbf{F} = \mathbf{0}$ . For the frontal plane, this results in  $F_y \cdot z - F_z \cdot y = 0$ . Since the forces  $F_y$  and  $F_z$  are measured by the force/moment sensors and the vertical position  $z$  can be estimated using the methods proposed in this study, an estimation of the lateral distance  $y$  between the feet and the projection of the CoM on the ground is obtained.

Second, the design of the instrumented shoe (Fig. 1) can be improved. The current design could raise questions about its influence on the gait pattern and the clinical applicability as it seems rather heavy and bulky. Nevertheless, the influence of the shoe on the gait pattern appeared to be small as reported by Liedtke *et al.* [27]. In that study, an evaluation of several gait parameters was performed with healthy subjects wearing the instrumented shoes, and light, normal, and heavy weight shoes. Significant differences between the shoe types were found in maximum GRF only, but these differences could not be attributed to individual shoe types. Moreover, the differences were small compared to the body weight of the subjects and were therefore not considered relevant. The small influence on gait was confirmed by the experience of patients, who were able to walk comfortably with the instrumented shoes. Irrespective of the possible gait adaptation caused by the modified shoes, the resulting gait was registered by both the ambulatory and the reference systems. The purpose of this study was to introduce the measurement method, not to optimize the design of the instrumented shoe. In principle, such an optimization will further reduce the influence on gait without affecting the accuracy of the measurement system.

The accuracy of the CoM displacement estimated by the ambulatory measurement system is dependent on the choice for the cutoff frequencies of the filters used to fuse the low-frequency and high-frequency components, as indicated by Fig. 3. Optimal values for the cutoff frequencies appeared to be 0.2 Hz for the horizontal direction and 0.5 Hz for the vertical direction. These values are, among other things, related to gait velocity and should be changed accordingly when, for example, the CoM displacement during running would be analyzed. The dependencies of the cutoff frequencies with respect to these variables need to be investigated thoroughly.

Although the rms difference between the CoM movement in vertical  $z$ -direction estimated by both measurement systems was small compared to the horizontal  $x$ - and  $y$ -direction and small compared to literature [17], the relative difference of the  $z$ -direction with respect to the CoM excursion in that direction was larger, as shown at the bottom of Fig. 4. Since the CoM movement in  $z$ -direction is estimated by high-pass filtering the double integrated CoM acceleration only, it does not contain low-frequency information. It should be noted that the horizontal and not the vertical position of the CoM with respect to the CoP is of primary interest in balance assessment [1], [28].

None of the aforementioned current drawbacks and remarks, however, jeopardize the potential value of the presented concept of continuous ambulatory CoM tracking in biomechanical analysis.

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