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Ambulatory estimation of foot placement during walking using inertial sensors

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ABSTRACT

This study proposes a method to assess foot placement during walking using an ambulatory measurement system consisting of orthopaedic sandals equipped with force/moment sensors and inertial sensors (accelerometers and gyroscopes). Two parameters, lateral foot placement (LFP) and stride length (SL), were estimated for each foot separately during walking with eyes open (EO), and with eyes closed (EC) to analyze if the ambulatory system was able to discriminate between different walking conditions. For validation, the ambulatory measurement system was compared to a reference optical position measurement system (Optotrak). LFP and SL were obtained by integration of inertial sensor signals. To reduce the drift caused by integration, LFP and SL were defined with respect to an average walking path using a predefined number of strides. By varying this number of strides, it was shown that LFP and SL could be best estimated using three consecutive strides. LFP and SL estimated from the instrumented shoe signals and with the reference system showed good correspondence as indicated by the RMS difference between both measurement systems being 6.5 ± 1.0 mm (mean $\pm$ standard deviation) for LFP, and 34.1 ± 2.7 mm for SL. Additionally, a statistical analysis revealed that the ambulatory system was able to discriminate between the EO and EC condition, like the reference system. It is concluded that the ambulatory measurement system was able to reliably estimate foot placement during walking.

1. Introduction

Human walking is often conceived as the motion of two coupled pendula (Winter, 1995). The double support phase is viewed as a transition from one inverted pendulum to the next. An efficient means to stabilize this essentially unstable system is to adjust foot placement (Townsend, 1985). Assessment of foot placement, especially the variability of foot placement between consecutive strides, reveals important aspects of balance (Bauby and Kuo, 2000).

Traditionally, foot placement is assessed using optical position measurement systems in a gait laboratory (Winter, 1995). Although these systems are clinically accepted as 'the golden standard', there are several drawbacks. Firstly, the number of consecutive strides that can be measured is limited. This means that variability of gait, involved in balancing the body and walking during varying circumstances, cannot be investigated using the existing systems as it requires a larger number of consecutive strides to be measured. Secondly, optical measurement systems suffer from marker visibility problems, since the line of sight from camera to marker is easily blocked due to movement of the subject. Instrumented treadmills provide a solution, allowing many strides to be measured (Owings and Grabiner, 2004; Danion et al., 2003; Hof et al., 2007). However, despite the advantages associated to treadmill walking, uncertainty remains regarding the extent to which treadmill walking can be used to mimic overground walking (Dingwell et al., 2001). In addition, the narrow path offered by the treadmill hinders freedom in selection of the trajectory and does not allow measurements during everyday life. These drawbacks stimulated several research groups to start initiatives for performing these measurements outside the laboratory, in an ambulatory environment.

An alternative to the traditional measurement systems is to use inertial (accelerometers and gyroscopes) and magnetic sensors (Roetenberg et al., 2005; Luinge and Veltink, 2005; Yun and Bachmann, 2006). Although these sensors do not suffer from the drawbacks associated to optical measurement systems or instrumented treadmills, they are not ideal as well. The estimation of position and orientation requires integration of acceleration and angular velocity, respectively, which gives rise to inherent drift caused by noise and a fluctuating offset. Moreover, relative positions of sensors with respect to each other cannot be estimated using inertial sensing only. However, during walking, the integration drift can be avoided by the use of regular zero velocity updates (Sabatini et al., 2005; Schepers et al., 2007; Tong and Granat, 1999).

Up to the authors' knowledge, 3D foot placement during many strides using an ambulatory measurement system has not been

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estimated previously. Veltink et al. (Veltink et al., 2003) described a three dimensional inertial sensing system for measuring foot movements during gait in stroke patients using a drop-foot stimulator. The main interest in this study was on the effect of stimulation parameters on foot orientation during the swing phase, not on the assessment of foot position. Although several authors (Aminian et al., 2002; Dejnabadi et al., 2006; Sabatini et al., 2005; Salarian et al., 2004; Zijlstra and Hof, 2003) estimated foot movement during several strides, these studies merely considered movement in the sagittal plane, whereas adequate lateral foot placement is also essential in balancing the body. This also holds for the wireless wearable measurement system proposed by Bamberg et al. (Bamberg et al., 2008). Bauby and Kuo (Bauby and Kuo, 2000) assessed foot placement variability using magnetic measurement system mounted on a rolling cart which was pushed near the walking subject. This measurement system allows several strides to be assessed, but still the rolling cart with the measurement equipment needs to be near the subject. In our previous study (Schepers et al., 2007) foot movement was estimated under ambulatory conditions using instrumented shoes, but the analysis was limited to single stride, the variation between strides was not assessed.

This study builds upon previous studies using the instrumented shoe, where it was used to assess ankle and foot dynamics (Schepers et al., 2007), and center of mass movement (Schepers et al., 2009). The instrumented shoe consists of a pair of orthopaedic sandals with two six degrees of freedom force/moment sensors beneath the heel and the forefoot and two inertial sensors rigidly attached to the force/moment sensors (Fig. 1). The current study investigates whether it is possible to estimate foot placement of a single foot, both in forward and lateral direction, for several strides using the inertial sensors on the foot allowing gait analysis in an ambulatory environment. The performance of the proposed system is evaluated by a comparison with an optical position measurement system while varying the walking conditions to analyze if the ambulatory system is able to discriminate between them.

2. Methods

2.1. Estimation of foot position and orientation

Position and orientation can be estimated using inertial sensors by integration of the accelerometer and gyroscope signals. Position is estimated by double integration of sensor acceleration after removal of gravitational acceleration. The removal of gravitational acceleration requires the orientation to be known, which is estimated by integration of angular velocity (Bortz, 1971). However, integration of angular velocity to orientation and double integration of acceleration to position is prone to drift due to sensor noise and a fluctuating offset. Since walking is a cyclical movement, several initial and final conditions can be applied to reduce this drift (Schepers et al., 2007). For the integration of angular velocity, the inclination can be assumed equal at every stride during the time interval when the foot is flat on the floor. In addition, it is assumed that the velocity is zero, and that the vertical position is constant during these intervals. Moreover, to minimize integration time, position estimation for each stride is divided in two phases. During the stance phase, position estimation is based on the orientation and knowledge about the geometry of the shoe (Schepers et al., 2007). During the swing phase, position estimation is obtained by integrating the signals of the heel accelerometer twice. To obtain the position of the foot for several strides, the estimated position changes for the individual strides are added.

2.2. Foot placement parameters

In order to evaluate the accuracy of the ambulatory measurement system for the analysis of foot placement, two parameters were chosen: lateral foot placement (LFP) and stride length (SL). Each stride, the time instant of heel down (HD) was defined using the force transducers beneath the instrumented shoe. The evaluated parameters, LFP and SL, were estimated for each foot separately, because the inertial sensors do not provide information about the relative positions of both feet. For LFP and SL estimation, an average walking path was calculated by a first order least squares fit to a predefined number of consecutive HD positions of the same foot. For each stride, the HD position was defined in the middle of the mounting plate beneath the heel (Fig. 1). An example of the average path for three consecutive strides is shown in Fig. 2. The parameters were defined as follows:

Stride length (*SL*) was defined as the distance between heel positions of the same foot in a direction parallel to the average walking path (Fig. 2) at two consecutive HD instants.

Lateral foot placement (*LFP*) was defined as the distance between the heel position and the orthogonal projection of this heel position of the same foot on the average walking path (Fig. 2).

Since the position and orientation were estimated by integration which is prone to integration drift (Section 2.1), the predefined number of strides used to determine the average walking path was varied to analyze its influence on the accuracy and to determine the optimal number of strides. It should be noted that the method proposed, which fits an average path using a moving window, implies that multiple LFPs and SLs will be estimated for the same stride.

2.3. Experimental methods

To determine the foot placement parameters, experiments were performed in a gait laboratory using a three dimensional optoelectronic reference registration system (Optotrak), consisting of five arrays of three cameras. Ten subjects participated in this study (7 male, 3 female, age 27.5 ± 4.9 years (mean $\pm$ standard deviation), length 1.76 ± 0.07 m, and body mass 72.0 ± 8.4 kg). Two subjects were excluded for data analysis due to setup problems of the reference system. All subjects were healthy and exhibited no clinical abnormalities. Informed consent was obtained from each subject prior to the experiment. The subjects, wearing instrumented shoes, were asked to walk through the laboratory (10 m walkway) under two different conditions (eyes open (EO) and eyes closed (EC)). These conditions were chosen to show the sensitivity of the ambulatory measurement system to different experimental conditions, and it is expected that these are typical examples of situations that need to be distinguished. Foot movement was estimated using instrumented shoes (Fig. 1) consisting of orthopaedic sandals with two six-degrees-of-freedom force/moment sensors (ATI-

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. A thin flexible glass-fiber plate connects both sensors allowing regular flexion of the foot during push-off.

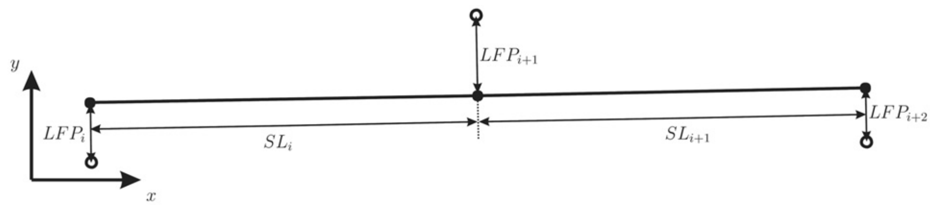

Fig. 2. Schematic view of three consecutive heel positions of the same foot (circles), and the orthogonal projections of these positions on the average walking path (dots). Stride length (SL) and lateral foot placement (LFP) are defined with respect to this average walking path.

Mini45-SI-580-20, supplier: Schunk GmbH & Co. KG) beneath the heel and forefoot. Moreover, an inertial sensor (Xsens Technologies B.V.) and an Optotrak marker were rigidly attached to each force/moment sensor.

The analog data from the force transducers and inertial sensor data were acquired at 50 Hz. Data from the reference system were acquired at 100 Hz, but resampled to 50 Hz. All data were low-pass filtered by applying a recursive second order Butterworth filter at 15 Hz. A pulse, generated by the Optotrak system, was recorded by an additional channel of the inertial sensor system. This pulse was used to synchronize Optotrak, force/moment and inertial data.

2.4. Data analysis

The data analysis is separated in three parts and was done for the ambulatory as well as the reference system. First, the number of strides used to construct the average walking path was determined by calculating the root mean square (RMS) differences between the ambulatory and the reference systems for LFP and SL. Note that this was done for the EO condition only. Second, the level of agreement between the ambulatory and the reference systems was determined by a Bland–Altman analysis (Bland and Altman, 1986). The 95% confidence intervals were determined by the mean ± 1.96 times the standard deviation of the difference (RMS difference) between both measurement systems (assuming a normal distribution). Finally, a statistical analysis was carried out using a repeated measures general linear model analysis with post-hoc comparison ($p < 0.05$) in SPSS, on the mean values of LFP and SL to determine whether the ambulatory system is able to distinguish between the EO and EC condition like the reference system. It should be noted that for LFP the mean of the absolute values was taken, since the definition of the average walking path implies a zero mean for LFP.

3. Results

3.1. Accuracy of LFP and SL estimation

The number of trials that have been analyzed were 75 for the EO condition, and 38 for the EC condition resulting in a total number of 113 trials. To determine the number of strides used to construct the average walking path, the root mean square (RMS) differences between the ambulatory and the reference systems were calculated for LFP and SL, which are shown in Fig. 3. The results indicate that the SL RMS difference is hardly influenced by the number of strides, whereas the LFP RMS difference increases as the number of strides increases. Since the smallest LFP RMS difference can be clearly seen at three consecutive strides, it is chosen to use three consecutive strides for the analysis in the remainder of this study.

3.2. Similarity of LFP and SL estimation

The LFP and SL of a representative subject for the EO condition are shown in Fig. 4. The encircled SL strides indicate the first and last stride of each trial. Fig. 5 shows the Bland–Altman plot for the EO data of all subjects by plotting the difference between both methods against the mean for LFP and SL. The 95% confidence intervals were determined to be -12.8 & 12.8 mm for LFP, and -64.0 & 76.9 for SL, based on a RMS difference of 6.5 ± 1.0 mm (mean $\pm$ standard deviation) for LFP, and 34.1 ± 2.7 mm for SL.

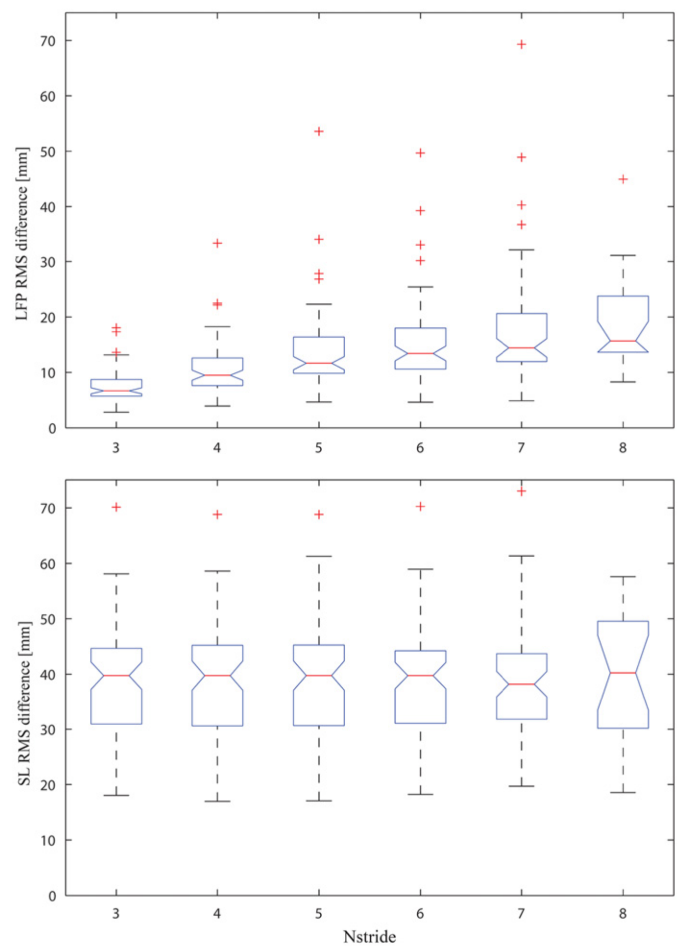

Fig. 3. Box and whisker plot of RMS differences for the LFP (top) and SL (bottom) estimated with ambulatory and reference systems for a varying number of strides used to calculate the average walking path. The RMS differences were calculated for all trials of the EO condition. The box has lines at the 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 plus signs.

3.3. Sensitivity for different experimental conditions

The results of the statistical analysis, that was used to analyze if the ambulatory measurement system can discriminate between the EO and EC condition like the reference system, are presented in Table 1. As expected, the influence of walking condition is significant for LFP and SL. The influence of walking condition is also graphically shown in Fig. 6 by an increase of LFP and a decrease of SL. Although the influence of measurement system is significant for the LFP comparison as indicated by the second column of Table 1, it is important to note that the last column of Table 1 indicates that there is no significant interaction effect.

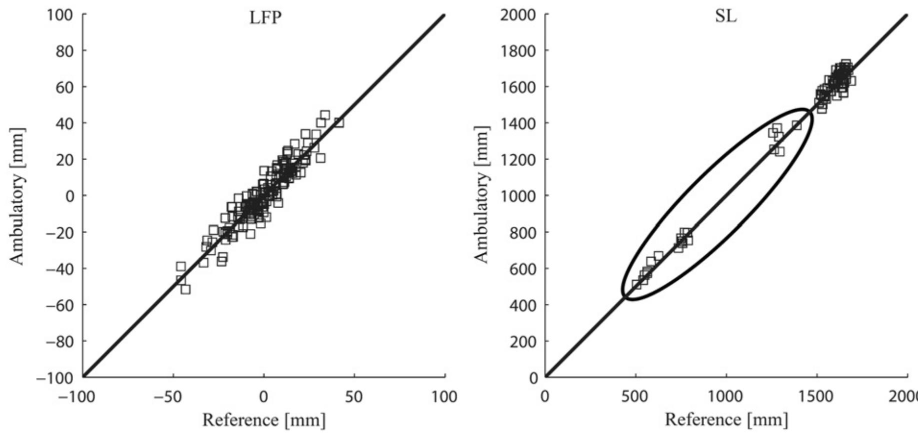

Fig. 4. LFP and SL estimated by the reference and ambulatory measurement systems of a representative subject for the EO condition. The encircled SL strides indicate the first and last stride of each trial.

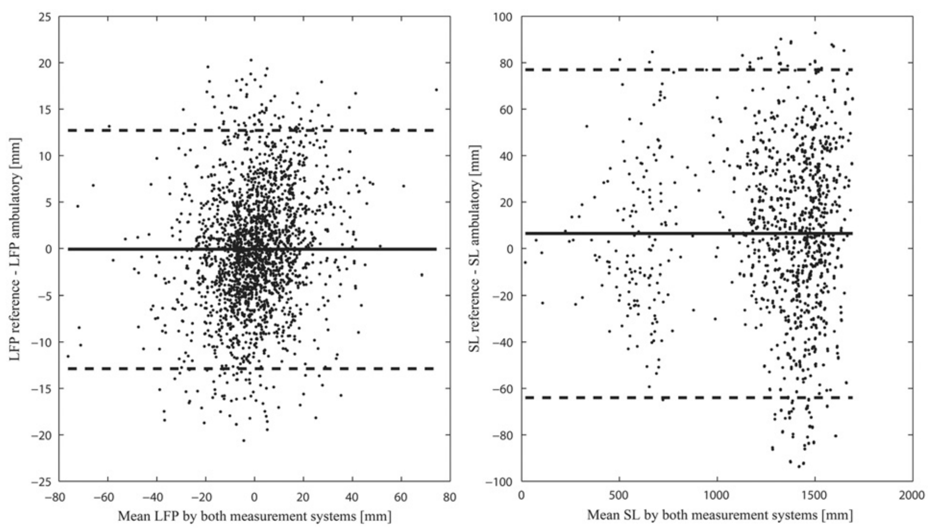

Fig. 5. Bland–Altman plot for LFP and SL estimated by the reference and ambulatory measurement systems for the EO data of all subjects. The solid line indicates the mean difference, the dashed lines indicate the 95% confidence intervals.

Table 1
Significance levels for the influence of the walking condition (EO-EC), the measurement system (ambulatory-reference), and their interaction for LFP and SL.

Parameter	System	Condition	System × condition
LFP	0.001*	0.021*	0.823
SL	0.063	0.007*	0.453

A significant difference is indicated by an asterisk (*).

The statistical analysis implies that both measurement systems register similar changes caused by different walking conditions, and are able to discriminate between the EO and EC condition.

4. Discussion

This study proposed a method to assess two spatial parameters of a single foot, lateral foot placement (LFP) and stride length (SL), during walking using an ambulatory measurement system. The measurement system consisted of a pair of instrumented shoes which were introduced previously (Schepers et al., 2007; Veltink et al., 2005). The results revealed in different ways that the ambulatory system can well be used in the assessment of LFP and SL. First, both measurement systems show good agreement, as indicated by the Bland–Altman analysis (Fig. 5). Second, the RMS differences between the ambulatory and the reference systems were lower than similar studies comparing foot placement estimation based on an ambulatory system using inertial sensing and a reference system (Aminian et al., 2002; Bamberg et al., 2008). These studies investigated stride length only, ambulatory systems have not been used before to estimate lateral foot placement. Third, the ambulatory system is able to discriminate between different walking conditions (EO and EC), like the reference system. The statistical analysis also revealed a significant difference between both measurement systems for LFP, which means that there is a systematic difference between both systems. This difference is small compared to the variance in the difference between both systems, and the Bland–Altman analysis already indicated that this variance was relatively small. It is expected that the proposed measurement system is able to distinguish different walking conditions for a patient population as well, but this should be demonstrated by a further evaluation.

The ambulatory measurement system was compared to a widely used ‘golden standard’, an optical position measurement system. Besides the drawbacks already mentioned in the

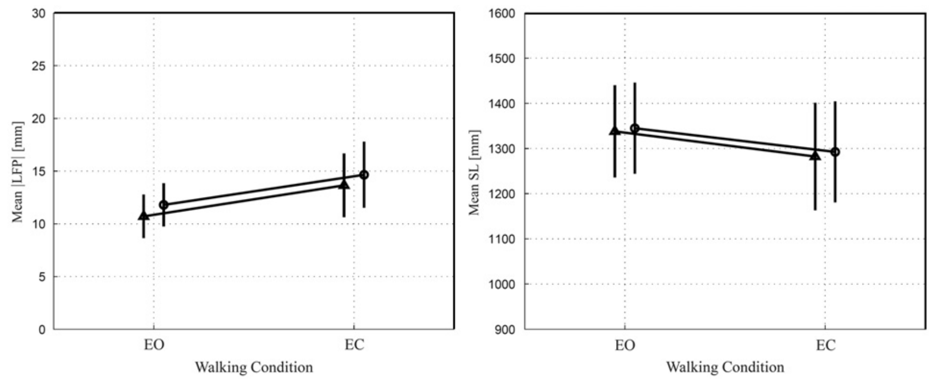

Fig. 6. Effect of measurement system and walking condition on LFP (left) and SL (right) with an average walking path determined using three consecutive strides. Absolute LFP or SL is shown on the vertical axis and walking condition (eyes open (EO) and eyes closed (EC)) on the horizontal axis. The circles and triangles indicate the mean values of the ambulatory and reference system, respectively. The vertical lines indicate the standard deviations.

introduction associated with optical position measurement systems, the relatively large measurement volume as used in the present study introduces another drawback. This caused two subjects to be excluded from the analysis as stated in Section 2.3. A solution would be to increase the number of cameras used, but this also increases the complexity of the measurement system as well as the financial costs.

The proposed method was used to estimate LFP and SL for a limited number of consecutive strides for each foot separately. If it is desired to estimate the relative positions of both feet with respect to each other, additional information needs to be included since the position is obtained by integration which means that only the change of position of the same foot can be estimated. A possible option to obtain the additional information is to use Newton's second law applied for rotational motion. Using the ground reaction force as measured by the instrumented shoe and the assumption that the product of angular acceleration and moment of inertia will be small during walking allows the lateral distance between both feet to be estimated (Schepers et al., 2009). If it is desired to relate more strides to each other or if the absolute positions in space of the feet are required, the system can be fused with another aiding system such as GPS (Tan et al., 2008) or magnetic tracking systems (Schepers et al., 2010; Roetenberg et al., 2007). However, these aiding systems will have their drawbacks as well and increase the complexity of the measurement system. Moreover, for balance assessment, most information is obtained by relating the current stride to one or a few consecutive strides.

It should be noted that the estimation of LFP and SL in an ambulatory environment do not necessarily require an instrumented shoe to be worn. In principle, an inertial sensor mounted on the foot or the shoe will suffice, while the presented methods to assess LFP and SL can be applied. In this study, the force transducers were merely used for gait phase detection, which could have been done using a gait phase detection algorithm based on inertial sensor signals (Sabatini et al., 2005). The advantage of the use of load sensors for gait phase detection, such as force transducers or footswitches, is the simple and straightforward method, since the response will only be unequal to zero during ground contact allowing accurate gait phase detection. Besides the estimation of spatial parameters, the instrumented shoe allows more gait variables to be estimated as shown in previous studies. For balance assessment for example, it is crucial to monitor center of mass trajectories, center of pressure trajectories and the relation between these variables which was shown in a previous study (Schepers et al., 2009). Another study (Schepers et al., 2007) showed the feasibility of the instrumented shoe to estimate ankle and foot dynamics accurately in an ambulatory environment. The method proposed in the current study enhances the potential of the instrumented shoe as an ambulatory measurement system allowing several spatial gait variables to be estimated in an ambulatory environment.

Besides the ability to estimate spatial parameters, the ambulatory system proposed can also be used to assess temporal parameters like stride time, (double) stance time and stride frequency. Especially a gait phase detection system based on load sensors, as used in the current study, allows these temporal parameters to be estimated easily. However, the sample frequency of the current setup was set to 50 Hz, which limits the accuracy of the estimation. For future versions of the hardware, a higher sample frequency is desired.

Although the shoe (Fig. 1) introduces extra height and weight compared to normal shoes, the influence of the shoe on the gait pattern appeared to be small (Liedtke et al., 2007). It should be emphasized that the purpose of this study was to evaluate the accuracy of the estimation of foot placement, not to analyze undisturbed gait. An optimization with respect to the design of the shoe, e.g. miniaturization of the sensors and components, will reduce the influence on gait without affecting the functionality.

The estimation of foot placement has a significant contribution to understanding human motor control. During human walking, foot placement is used to balance the body. It would be interesting to use the proposed measurement method to characterize foot placement in more challenging environments such as circular walking (Kiriyama et al., 2005) or walking on uneven terrain. In principle, the methods proposed in this study are appropriate for such applications.

Conflict of interest statement

There are no conflicts of interest.

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