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***Abstract*—This study aimed at estimating the base of support (BoS) during gait with an unobtrusive wearable system. The BoS is the area identified by the points of contact of the body with the ground. During the double-support phase of gait both feet contribute to the BoS. The BoS is a measure of the subject's balance. The estimation of the BoS outdoors is still an open issue. The innovative hardware presented here is a combination of a magneto-inertial measurement unit (MIMU) and two infrared time of flight (IR-ToF) sensors. The methods implemented to estimate the BoS were tested on a healthy subject and validated with a stereo-photogrammetric system. The results suggest that the solution proposed may be an effective and low-cost tool for the estimate of the BoS during gait in outdoor conditions.**

***Keywords*—out-of-lab gait analysis, MIMU, infrared time-of-flight sensors, base of support.**

I. Introduction

THERE is ample evidence that the base of support (BoS) and the step width (SW) are important parameters for the assessment of dynamic stability, balance control and risk of falling in healthy and pathological gait [1], [2]. In general, widening the BoS leads to an increase of stability as often observed in parkinsonian patients and in the elderly [3].

The BoS during walking can be defined as the area comprised by the outer edges of the feet in contact with the ground [4]. It is closely related to the SW, which is defined as the mediolateral distance between the feet [1].

In most studies assessing stability, BoS and SW were estimated from force-platforms and a stereo-photogrammetric (SP) system [1], [4]-[7]. Although these technologies provide accurate estimates, they restrict gait analysis to laboratory applications. Miniaturized magneto-inertial measurement units (MIMUs) have been proposed to analyse gait in real life, mostly in terms of body segments kinematics and of gait temporal and spatial parameters, such as stride time and stride length [8]-[11]. However, it is not possible to directly determine the relative position of two body segments in a common reference coordinate system using two MIMUs attached to them, as the physical quantities recorded by each MIMU are self-referenced [12]. This limitation makes MIMUs alone not suitable for estimating the BoS in real-world conditions. To overcome this limitation, the most straightforward solution is to integrate the MIMU system with additional sensors capable to provide distance measurements (i.e. inter-foot distance, IFD). There are few studies in the literature estimating IFD in out-of-lab conditions using ultrasound or infrared (IR) sensors. Weenk *et al.* [13] proposed a method to calculate the IFD using ultrasound sensors, requiring the instrumentation of both feet, using both an emitter and a receiver. Hung *et al.* [14] proposed to attach an IR LED to a foot and a camera to the other, a solution too bulky to be suitable for prolonged use. Methods based on light-intensity IR sensors as proposed by Trojaniello *et al.* [1] overcome the limit of the hardware distinction between emitter and receiver, however their performance is not stable as the environmental conditions change [16], [17]. Recently, Bertuletti *et al.* [17] proposed to integrate IR time-of-flight (ToF) sensors and a MIMU in a single wearable system (SWING). This technology, compared to the previous ones, allows to embed the transmitter and the receiver in the same chip, to reduce the size of the sensor, to increase the output data rate up to 50 Hz, lower power consumption (~2-5 mA) and to guarantee stable performance changing the environmental conditions [17]. The system was used for IFD estimation and step detection [18].

The goal of this work was to devise and preliminarily validate innovative methods based on the combined use of IR-ToF sensors and magneto-inertial technologies for the estimation of SW and BoS. Data acquired during straight walking of a healthy subject were compared to concurrent stereo-photogrammetric data, used as a gold standard.

II. Material and methods

A. Experimental setup

The SWING system [19] was attached to the medial side of the right foot, identified as the instrumented foot (InFoot).

The MIMU (3D accelerometer range ±16 g, 3D gyroscope range ±2000 dps, 3D magnetometer range ±50 Gauss, output data rate 100 Hz) and the IR-ToF sensors (VL6180X, STMicroelectronics [20], measured distance range 0−200 mm, sampling frequency 50 Hz), were positioned as described by Bertuletti *et al.* [18].

The position of twenty retro-reflective markers attached to both feet (Fig. 1) were recorded by a 12-camera SP system (mod. Vero, Vicon, Oxford, UK).

Data were recorded while a healthy subject walked at a self-selected speed along a 5 m-walkway for 10 times with and without a flat screen applied to the medial side of the non-instrumented foot (Non-InFoot). To synchronize SWING data and SP data, the subject was asked to hit the InFoot on a force platform (electronically synchronized with the SP system) at the beginning and at the end of each acquisition.

Experiments conformed to the standards set in the Declaration of Helsinki.

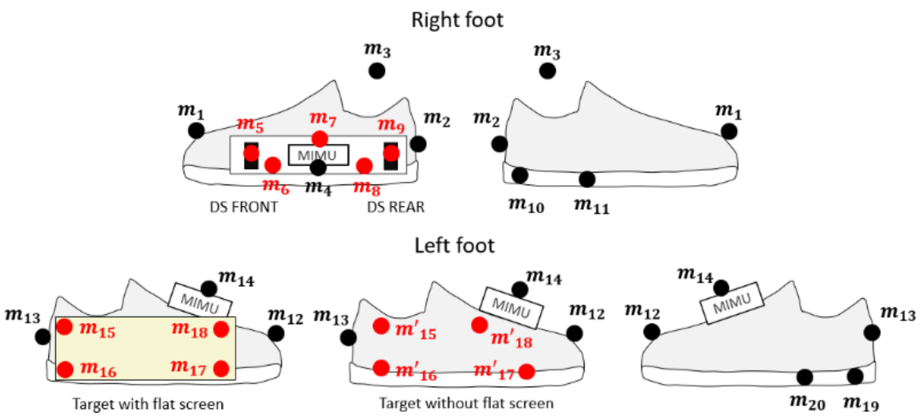

Fig. 1: Experimental setup. A rigid support with a MIMU and two IR-ToF sensors (DS REAR and DS FRONT) is attached to the medial side of the right foot. The MIMU attached to the left foot was used only to ensure that the Non-InFoot was stationary during the considered IR-ToF recordings. The black and red dots are the markers. The red markers are used only in the static acquisition for calibration purposes and removed during dynamic acquisitions not to interfere with the acquired movement. Their trajectories are reconstructed exploiting the rigidity of the three markers in the rear foot (m_2, m_{10}, m_{11} for the right foot and m_{13}, m_{19}, m_{20} for the left foot) . The first two illustrations of the left foot show the two tested setups with and without a screen attached to the medial side of the foot.

B. *Data Analysis*

The orientation of the InFoot was obtained using a sensor fusion algorithm based on the Madgwick's filter [21]. The displacement of the InFoot was obtained by double integrating the accelerations, after filtering [22] and gravity subtraction [9]. The integration interval was defined between two consecutive foot flat instants of the InFoot. To evaluate the foot flat instant a ZUPT detector based on the gyroscopic signals was applied [23]. To reduce drift, direct and reverse integration was implemented [22].

Once both orientation and position of the InFoot were determined in each instant of time, the position of the Non-InFoot with respect to a common coordinate system was obtained by following the procedure described below.

1. During the InFoot swing phase, IR-ToF readings provided the values of the distance between the IR-ToF sensors on the InFoot and the Non-InFoot surface.

2. The observed points on the Non-InFoot medial surface were all expressed in a global coordinate system (GCS), the InFoot (MIMU) coordinate system (in orange in Fig. 2) at the beginning of the swing (toe-off t_0 in Fig. 2).

3. The detected points of the Non-InFoot expressed in the GCS are fitted with a linear model, approximating the medial side of the Non-InFoot. The actual foot length is imposed to the linear model to define a segment approximating the medial edge of the Non-InFoot footprint.

C. *SW and BoS estimation*

The SW was estimated as the average of the recorded distances between the InFoot and the reconstructed Non-InFoot during the swing of a foot.

Reference values for SW were obtained based on marker trajectories, by computing the line passing through the IR-ToF sensors and perpendicular to the rigid support. The position of the IR-ToF sensors, the rigid support, as well as the medial side of the Non-InFoot (Fig. 1) were reconstructed by calibrating them during a static acquisition [24] and during dynamic acquisitions, knowing the position and orientation of both feet with respect to GCS, the trajectories of the markers were achieved.

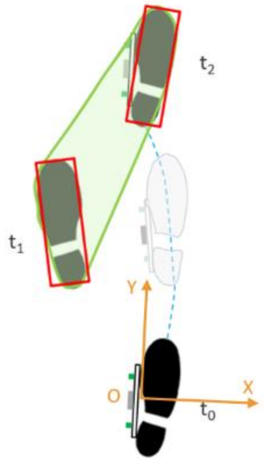

Fig. 2: Footprints from the beginning (t_0) to the end (t_2) of the swing phase of the InFoot. The BoS is considered between the Non-InFoot footprint in its stance phase (t_1) and the InFoot footprint in t_2. The red rectangles are the approximation of the footprints. The BoS is the green area. The light blue dotted line represents the right foot trajectory. The GCS is represented in orange.

The BoS was defined as the sum of the area of each footprint and the area between them during double support (Fig. 2). The footprints obtained with the SWING system were approximated with a rectangle with sides equal to the measured foot length and width, and the area between them is calculated with the Bretschneider formula for irregular quadrilaterals.

The gold standard values of the BoS were computed obtaining the feet length with the heel and toe marker, while the width of the footprint was imposed. The footprint medial vertices are linked to define the area between the feet.

Errors affecting the BoS estimation were characterized by the area error and the BoS location error (left/right shift). The latter is evaluated as the distance between the footprint centers calculated with the wearable system and with the SP system (Fig. 3). Fig. 4 shows a BoS area shifted forward and left with respect to the corresponding BoS obtained with the SP system. Thus, the footprint centers obtained with the wearable system do not coincide with those obtained with the SP system.

A descriptive statistic on both area errors and left/right shifts was computed considering the mean, standard deviation and 95% confidence interval, which is taken as descriptor of the measurement precision.

III. Results

Errors affecting the estimates of both SW and BoS were computed with and without the presence of the screen on the medial side of the Non-InFoot.

A total of 92 SW values with the screen and 69 without were analysed, whereas a total of 20 BoS values both with and without the screen were analysed. Table I-IV illustrate the results.

When the screen is used the estimated Non-InFoot footprint is on average 55 mm far from its position obtained with the SP system as opposed to 62 mm when the screen is not used (Table III and IV). The InFoot position error is on average 67 mm with the screen and 92 mm without it (Table III and IV).

IV. Discussion and Conclusions

The study proposed and validated a method for estimating SW and BoS alternative to current methods used in gait analysis laboratories. Instrumenting a foot with a MIMU and two IR-ToF sensors was shown to be adequate to obtain valuable estimates of SW and BoS, in healthy gait.

The use of a screen applied to the medial side of the non-instrumented foot was shown to be beneficial for the estimate

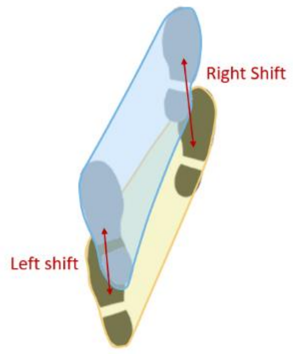

Fig. 3: BoS area obtained with wearable system (light blue) and with the SP system (yellow). If the position on the floor of the areas does not coincide, the Left and Right Shit are defined as the distance between the footprint centers.

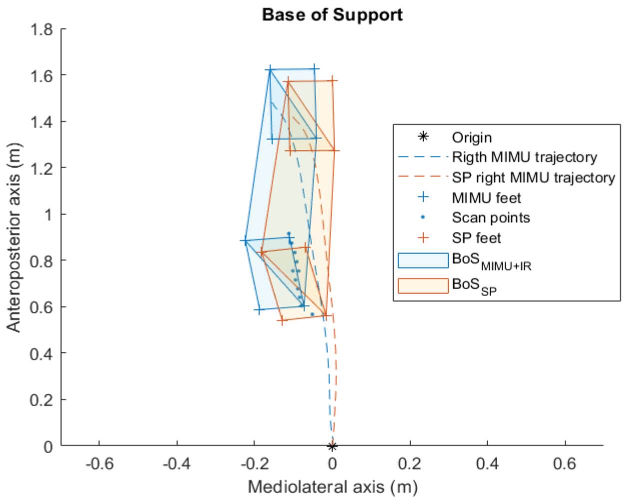

Fig. 4: A BoS obtained with the wearable system (light blue) and with the gold standard (yellow). The light blue dots (‘scan points’ in the legend) are the points of the Non-InFoot detected by the IR ToF sensors. It can be noticed that the resulting BoS is shifted forward and left..

TABLE I
SW values and errors (with screen)

Mean (mm)	*Mean Error (mm)*	*Std Error (mm)*	*Percentage Error (%)*	*95% Confidence Interval Error (mm)*
74	-23	21	23	(-27, -18)

TABLE II
SW values and errors (without screen)

Mean (mm)	*Mean Error (mm)*	*Std Error (mm)*	*Percentage Error (%)*	*95% Confidence Interval Error (mm)*
56	-27	32	32	(-34, -19)

Table I and Table II: SW obtained results in terms of estimates and errors, calculated on 92 steps with the screen and on 69 steps without it. Negative values correspond to underestimation.

TABLE III
BoS errors (with screen)

Error Type	*Mean*	*Std*	*95% Confidence interval*
Area Error (mm^2)	-4100	6300	(-6900, 1300)
Percentage Area Error (%)	3.8	2.9	(2.5, 5.1)
Left shift (mm)	55	24	(45, 65)
Right shift (mm)	67	32	(53 81)

TABLE IV
BoS errors (without screen)

Error Type	*Mean*	*Std*	*95% Confidence Interval*
Area Error (mm^2)	-12500	34500	(-27600, 2600)
Percentage Area Error (%)	10.1	19.8	(1.5, 18.8)
Left shift (mm)	62	27	(50, 74)
Right shift (mm)	94	46	(74, 114)

Table III and Table IV: BoS estimate errors calculated over 20 strides with and without the screen. Area error is the difference between the BoS area found with the wearable system and that found with the SP system. The left and right shifts are the distances between the footprint centers obtained with the wearable system and those obtained with the SP system. Negative values correspond to underestimation.

of both SW and BoS: 23% SW error with the screen vs 32% without it (Table I and II), 4% BoS area error with the screen vs 10% without it (Table III and IV). This is mainly due to the irregular surface of the Non-InFoot medial side. However, these preliminary results suggest that the experimental setup without the screen can be adopted for a greater comfort of the subject, considering that the SW errors differences between the two setups are not significant and that the 95% confidence intervals of SW and BoS errors overlap.

It was also observed that the IR-ToF sensors should not be positioned too close to the shoe sole to increase the chances of correctly recording signals even during swings characterized by a high clearance [18]. The maximum measured distance was set to 200 mm. This choice was a compromise resulting from two considerations: it had to be high enough to include cases of a significant external foot rotation, without compromising the output data rate [18].

The position estimation and the fitting of the non-instrumented foot are the most crucial sources of errors. In fact, the footprint position estimation is affected by an error both in the medio-lateral and the anteroposterior directions (Fig. 4). Besides the inaccuracy in fitting the distance data, the displacement estimation corrupts the footprint positioning due to some residual integration drift.

Future work should primarily focus on: *i)* improving the displacement estimation to enhance footprint positioning and *ii)* increasing the number of IFD values collected per stride to improve the reliability of the fitting of the distance data. This can be done either by using more than two IR-ToF sensors or by adopting a new technology that allows to estimate the IFD with an output data rate higher than 50 Hz.

Furthermore, methods' robustness should be tested on a larger number of subjects both healthy and with gait disorders.

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