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Experimental evaluation of indoor magnetic distortion effects on gait analysis performed with wearable inertial sensors

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Abstract
Magnetic inertial measurement unit systems (MIMU) offer the potential to perform joint kinematics evaluation as an alternative to optoelectronic systems (OS). Several studies have reported the effect of indoor magnetic field disturbances on the MIMU's heading output, even though the overall effect on the evaluation of lower limb joint kinematics is not yet fully explored. The aim of the study is to assess the influence of indoor magnetic field distortion on gait analysis trials conducted with a commercial MIMU system. A healthy adult performed gait analysis sessions both indoors and outdoors. Data collected indoors were post-processed with and without a heading correction methodology performed with OS at the start of the gait trial. The performance of the MIMU system is characterized in terms of indices, based on the mean value of lower limb joint angles and the associated ROM, quantifying the system repeatability. We find that the effects of magnetic field distortion, such as the one we experienced in our lab, were limited to the transverse plane of each joint and to the frontal plane of the ankle. Sagittal plane values, instead, showed sufficient repeatability moving from outdoors to indoors. Our findings

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provide indications to clinicians on MIMU performance in the measurement of lower limb kinematics.

Keywords: IMU/MIMU, inertial sensors, gait analysis, magnetic field distortion, joint kinematics

(Some figures may appear in colour only in the online journal)

1. Introduction

Recent advances in MEMS and micro-packaging technology have rekindled interest in both accelerometer and gyroscope sets, known as an inertial measurement unit (IMU), to conduct movement analyses (Bachmann *et al* 2003, Cappa *et al* 2007, Fong and Chan 2010, Abaid *et al* 2013). The main attractive merits of the IMU, in comparison to the optical motion capture systems, are: no workspace limits, low cost and easy operation. The main drawbacks are the random walk affecting angular rotation measurements, and their unsuitability as a positional tracking system. Commercially available IMUs combine three linear accelerometer axes and three gyroscope axes into one system.

The IMU orientation can be computed by the integration of the gyro outputs, but the procedure is affected by relevant data degradation. In fact, the noise affecting measurement signals causes a random drift, generally addressed as a 'random walk', of the output obtained from the numerical integration of the gyro signal. The gyro drift can be partially corrected by processing the accelerometer outputs using data fusion algorithms based on: linear Kalman filtering (Roetenberg *et al* 2005, 2007), extended or unscented Kalman filtering (Sabatini 2006, St-Pierre and Gingras 2004, Shin and El-Sheimy 2004), and particle filtering (Wang *et al* 2007).

Moreover, the noise affects also the position estimation, which is based on the double integration of the acceleration signal; in order to correct the position drift, additional information provided by a global positioning system signal is usually employed (Godha and Lachapelle 2008, Morrison *et al* 2012). Thus, accelerometers are generally used to estimate the tilt angle by measuring the gravitational acceleration components on its axes. Nevertheless, because gravitational acceleration is invariant with respect to rotations in the horizontal plane, the accelerometers are not suitable for correcting the heading drift of gyros, and, as a consequence, the 3-DoF evaluation of anatomical joint angles is partially compromised (Cooper *et al* 2009, Findlow 2008, Simcox *et al* 2005, Lau *et al* 2009).

In order to cope with the inaccuracy of orientation data, a three-axis magnetometer is embedded in the IMU, to obtain a magnetic IMU (MIMU) system (also called a MARG sensor or IMMS). The magnetometers introduce further error due to magnetic field disturbances induced by iron objects close to the sensor (Gebre-Egziabher *et al* 2006). Kendell and Lemaire (2009) reported that metals embedded in mobility aid devices cause errors of up to 35° in orientation measurements. De Vries *et al* observed a decrease in MIMU accuracy induced by indoor magnetic field disturbances. They reported a heading error up to 29° (de Vries *et al* 2009) with an initial period of 50–60 s to stabilize the Kalman filter. They also recommended to limit the use of MIMUs inside 'safe areas' and to map the magnetic field if the trials are carried out indoors. However, no quantitative information has been given on the actual reliability of kinematic analyses performed with MIMUs. In actual fact, an evaluation of the repeatability level obtainable indoors with MIMU-based gait analysis is crucial to assess the viability of such systems in routine clinical analyses and, in perspective, as a possible alternative to optoelectronic system (OS).

The aim of the present work, therefore, is twofold. Firstly, to assess the effect of the magnetic field distortion, present in a typical motion analysis lab, on the evaluation of lower limb joint kinematics performed with a commercial MIMU system. Secondly, to assess whether the measurement of the actual magnetic field distortion at the start of the trial could be beneficial to correct the MIMU heading error measured during gait. Thus, we seek to answer to the two following questions. (i) What is the decrease of repeatability on lower limb joint kinematics in a motion analysis lab due to a non-uniform magnetic field? (ii) Does a preliminary evaluation of the MIMU heading error, carried out at the start of the walk, increase the repeatability associated with kinematic data?

2. Theoretical approach

2.1. Gait kinematics

The solution of the lower limb kinematics consists in the evaluation of the joint rotations between the body segments and, therefore, in the calculation of the joint rotation matrices. The rotation matrix ${}^{b_i}\mathbf{R}_{b_j}$ between two consecutive body coordinate systems CS_{b_i} and CS_{b_j} can be computed as

$$ {}^{b_i}\mathbf{R}_{b_j} = {}^{b_i}\mathbf{R}_g \, {}^{g}\mathbf{R}_{b_j} \tag{1} $$

where CS_g is the ground fixed coordinate system. Introducing the sensor coordinate system CS_{s_i}, associated with the ith MIMU sensor, we obtain

$$ {}^{g}\mathbf{R}_{b_i} = {}^{g}\mathbf{R}_{s_i} \, {}^{s_i}\mathbf{R}_{b_i}. \tag{2} $$

Since soft and hard iron disturbances (Gebre-Egziabher *et al* 2006) may cause the MIMUs not to have the same ground-fixed reference frame CS_g (Picerno *et al* 2011), a different CS_{g_i} for each sensor should be used in equation (2):

$$ {}^{g}\mathbf{R}_{b_i} = {}^{g}\mathbf{R}_{g_i} \, {}^{g_i}\mathbf{R}_{s_i} \, {}^{s_i}\mathbf{R}_{b_i} \tag{3} $$

where ${}^{g_i}\mathbf{R}_{s_i}$ is computed continuously in real-time by the MIMU sensor. While ${}^{g}\mathbf{R}_{g_i}$ is dependent only on the local magnetic field deviation, ${}^{s_i}\mathbf{R}_{b_i}$ can be assumed as constant, neglecting the effect induced by soft-tissue artefacts.

The rotation matrix of the joint between the ith and jth body segment is, therefore, equal to

$$ {}^{b_i}\mathbf{R}_{b_j} = ({}^{g}\mathbf{R}_{b_i})^{\mathrm{T}} \, {}^{g}\mathbf{R}_{b_j} = ({}^{g_i}\mathbf{R}_{s_i} \, {}^{s_i}\mathbf{R}_{b_i})^{\mathrm{T}} \, {}^{g_i}\mathbf{R}_{g_j} \, {}^{g_j}\mathbf{R}_{s_j} \, {}^{s_j}\mathbf{R}_{b_j} \tag{4} $$

where ${}^{g_i}\mathbf{R}_{g_j}$, ${}^{s_i}\mathbf{R}_{b_i}$ and ${}^{s_j}\mathbf{R}_{b_j}$ should be determined by means of ad-hoc procedures. So far, only methods focused on ${}^{s_i}\mathbf{R}_{b_i}$ and ${}^{s_j}\mathbf{R}_{b_j}$ are reported in the literature. ${}^{g_i}\mathbf{R}_{g_j}$, instead, is always assumed to be equal to the identity matrix, i.e. all the sensors are considered to be referred to the same earth frame.

The methodologies for the measurement of the body-to-sensor rotation matrix fall into two categories: (i) procedures based on a functional approach (Luinge *et al* 2007, O'Donovan *et al* 2007, Cutti *et al* 2008, Favre *et al* 2009), and (ii) procedures which imply the use of an additional position/orientation measurement device such as, for example, a supplementary MIMU (Picerno *et al* 2008). Both categories, however, do not take into account the effect induced by both the environment magnetic distortion and the intrinsic inaccuracy associated with the magnetic sensor, which could affect the reliability of the joint kinematics (Picerno *et al* 2011).

2.2. Body-to-sensor calibration

The procedure adopted in this study falls into the second category above reported, and we decided to use an OS to estimate ${}^{s_i}\mathbf{R}_{b_i}$ to overcome the low repeatability associated with functional body-to-sensor calibration. Placing reflective markers on each MIMU and on each involved body segment, the absolute orientation of all of the coordinate systems, necessary for the evaluation of ${}^{s_i}\mathbf{R}_{b_i}$, can be estimated. Specifically, if ${}^{0}\mathbf{R}_{b_i}$ and ${}^{0}\mathbf{R}_{s_i}$, where CS_0 indicates the OS reference frame, are estimated via OS in a static trial, the body-to-sensor calibration matrix is

$$ {}^{s_i}\mathbf{R}_{b_i} = {}^{0}\mathbf{R}_{s_i}^{\mathrm{T}}\,{}^{0}\mathbf{R}_{b_i}. \tag{5}$$

It is important to point out that, in order to gather kinematic data coherent with anatomical conventions using MIMUs (Wu *et al* 2002), the body-to-sensor calibration is mandatory and it has to be repeated if the sensor should be repositioned.

2.3. Heading correction

Built-in sensor fusion algorithms based on the Kalman filter are able to auto-compensate magnetic distortion when the acquisition is short-lasting (less than 1 min) (de Vries *et al* 2009). When MIMUs stay in a magnetically distorted area for more than 1 min, they should no longer be considered as having the same ground frame. Moreover, the distortion effect lasts for at least 1 min if the sensors are in an area with a uniform magnetic field. Since the use of MIMUs in uniform magnetic field areas may be a demanding constraint for a clinical gait analysis, we decided to perform an initial heading correction of the magnetic field distortion and to quantify the consequent benefit.

At the beginning of each walking trial, when the subject stood still, the actual ${}^{g_i}\mathbf{R}_{g_j}$ is

$$ {}^{g_i}\mathbf{R}_{g_j} = \left({}^{0}\mathbf{R}_{s_i}\,{}^{g_i}\mathbf{R}_{s_i}^{\mathrm{T}}\right)^{\mathrm{T}}\,{}^{0}\mathbf{R}_{s_j}\,{}^{g_j}\mathbf{R}_{s_j}^{\mathrm{T}}. \tag{6}$$

We used the OS to estimate ${}^{0}\mathbf{R}_{s_i}$ and ${}^{0}\mathbf{R}_{s_j}$, whereas the MIMU outputs were collected to obtain ${}^{g_i}\mathbf{R}_{s_i}$ and ${}^{g_j}\mathbf{R}_{s_j}$. The heading correction procedure was carried out only indoors, because outdoors: (i) we cannot use the OS, and (ii) the magnetic distortion was negligible, as is reported in the following section.

3. Experimental validation

3.1. Evaluation of magnetic field map inside and outside the laboratory

In a preliminary phase, we mapped the magnetic field of the two walking areas: inside and outside the lab. The Movement Analysis and Robotics Laboratory (MARlab) of the Children's Hospital 'Bambino Gesù' consisted of a large room, with a 10 m gait path equipped with two six-component platforms, embedded in the floor, and an OS (figure 1(a)). The outdoor walkway was a levelled area, the same length as the laboratory one and free from metallic objects.

The MIMUs used in the present paper were MTx sensors (XSENS, static RMS-error $<1^\circ$, dynamic RMS-error $<2^\circ$) plugged to a XBus-Master system. Data were logged at a frequency of 50 Hz. Four MIMUs were fixed to a rigid wood plank, sized $160 \times 15 \times 5$ cm, with double-sided tape. By aligning the same edge of the sensors with one of the edges of the plank, we ensured the sensors had the same orientation. The vertical distance among the sensors was equal to 40 cm. The plank was placed perpendicularly to the ground, mapping a grid of 20×5 points spaced 30 cm. After a warm-up period for the sensors of about 20 min, in accordance

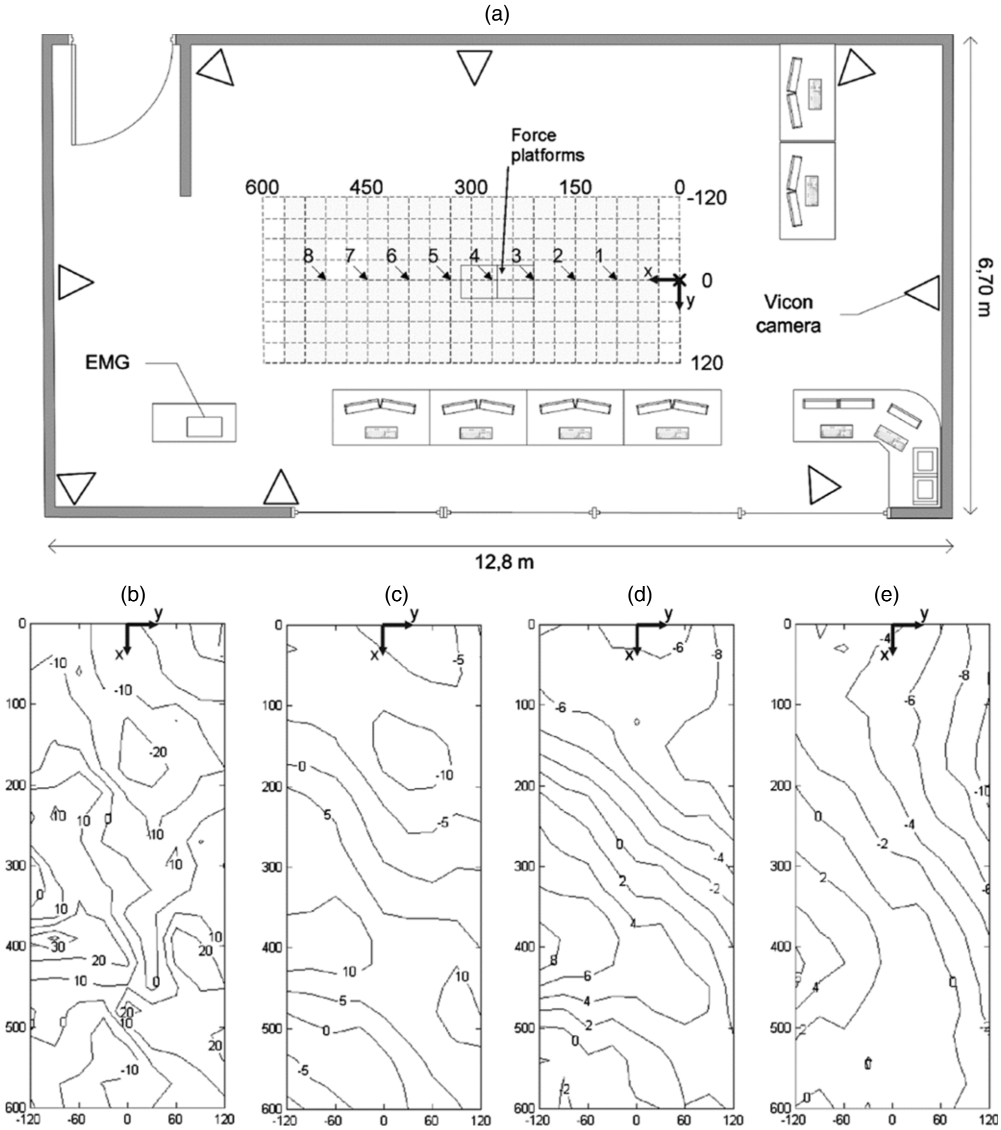

Figure 1. (a) Plan of Movement Analysis and Robotics Laboratory, Children's Hospital 'Bambino Gesù'; numbers (1–8) indicate the starting positions during gait analysis trials. Maps of magnetic field distortion at different heights: (b) 10 cm height level; (c) 50 cm height; (d) 90 cm height; and (e) 130 cm height. Contour lines express levels of magnetic field vector deviation (in degrees) from the value measured at the reference point marked in bold [0; 0; 90 cm] in (a).

with the manufacturer's specifications, the magnetometer outputs were acquired for 10 s, and the average direction of the magnetic field was computed for each grid node. All the magnetic deviations are reported, assuming as a reference the field direction measured at the point [0; 0; 90 cm]; see figure 1(a).

The maps of the magnetic deviation for the four different heights (10, 50, 90 and 130 cm) are shown in figures 1(b)–(e). The figure shows that higher distortions were measured, as expected, at ground level, both in terms of mean value and homogeneity. Moving from the lowest plane to the highest, the ranges of the magnetic field deviation decreased (50°, 20°, 16° and 16°). Spatial uniformity, instead, increased rapidly from the 10 cm level to the 50 cm level, and more slowly moving to higher levels.

From the measurements conducted outside the lab, the outdoor magnetic field result was basically homogeneous (a maximum spatial gradient of about 2° m^{-1} along the walkway) with a maximum deviation of less than 4° and, therefore, it is not reported in a figure.

3.2. Experimental procedure

A young adult subject (30 year-old male, 170 cm, 68 kg) with no previous history of orthopaedic or neurological pathologies was analysed. The protocol was approved by the Ethics and Medical Board of the 'Bambino Gesù' Children's Hospital. The subject provided informed consent to be involved in the study.

The MTx sensors were used for the kinematic analysis. The analysis was articulated into four experimental sessions repeated on four different days. Each session included eight repetitions of gait analysis trials performed both outdoors and indoors.

In order to minimize the effects of inherent inaccuracies of the MIMUs, each sensor was placed with different order on the examined body segments, so that each segment was coupled with a different sensor in each session. Finally, the time length of each trial was approximately equal to 2 min so that about 70 strides per trial were collected.

The scheme of the adopted experimental methodology is depicted in figure 2 and the detailed description is reported below.

3.2.1. Outdoor session. After a warm-up period for the sensors of about 20 min, in accordance with the recommendation of the Xsens' manufacturer, the MIMUs were aligned outdoors with each other, by placing them on the same wood plank used for magnetic field mapping and described in section 3.1. The heading offsets were reset using a specific procedure provided by the manufacturer's software. Then, the right leg of the healthy subject was equipped with MIMUs on the pelvis, thigh, shank and foot. The subject performed the trials standing still at the starting place for 2 min, in order to stabilize the sensor output in accordance with the findings of de Vries *et al* (2009) based on the same sensors used in the present work. Then the subject was asked to walk with a self-selected cadence for 2 min. The trial was repeated eight times.

3.2.2. Indoor session. After the completion of outdoor session, the subject moved into the lab, maintaining the MIMUs in place. The right leg and each MIMU were instrumented with reflective markers: a ten-marker set for the subject, according to the plug-in-gait model based on the Davis' protocol (Davis *et al* 1991), and a twelve-marker set for the MIMUs, i.e., three markers per sensor, placed as in figure 3. An eight-camera OS (Vicon MX camera-workstation, Nexus 1.7 software, 200 Hz, Oxford, UK) was used to perform the body-to-sensor calibration and the heading correction. Before starting the set of eight gait analysis trials, while the subject was still standing upright, the body-to-sensor calibration procedure was performed to estimate the matrices ${}^{s_i}\mathbf{R}_{b_i}$. Then, at the beginning of each walking trial, the subject waited for the stabilization of the MIMU filter for 2 min at the starting place, as he did outdoors, and, standing still few seconds, permitted via the OS the gathering of the relative ground matrices ${}^{g_i}\mathbf{R}_{g_j}$ for the heading correction. Each of the eight trials started at a different point in the lab indicated in figure 1(a) with $1, \ldots ,8$. We defined the start and the end of the gait path with two turn lines where the subject reversed their walking direction by 180°. The first walking

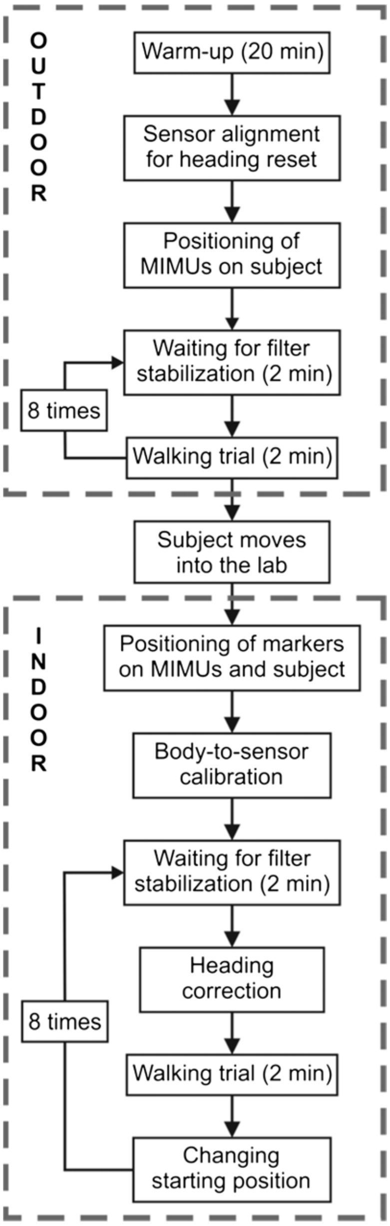

Figure 2. Workflow of each of the four sessions articulated in two sub-sessions performed outdoors and indoors.

direction for each of the eight trials was chosen in order to gather a convenient number of strides before the first direction change.

3.3. Data analysis

Custom-written software (MATLAB, Natick, MA, USA) was used to process the data, and the analysis steps described below are depicted in figure 4. Outdoor gait trials (OUT) were

Figure 3. Scheme of the subject instrumented with the MIMUs and the reflective markers. Blue, green and red arrows indicate the X, Y and Z axes of the ith MIMU sensor frame CS_{s_i}.

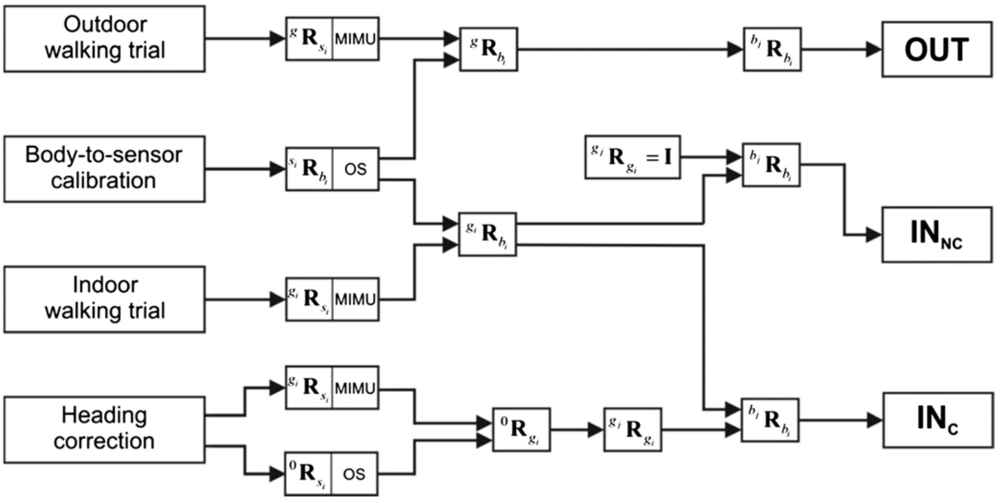

Figure 4. Data flow corresponding to the three analysed conditions: outdoor (OUT), indoor not corrected ($\mathrm{IN_{NC}}$), and indoor corrected ($\mathrm{IN_C}$).

post-processed without correcting the heading error due to the negligible distortion of the magnetic field. In order to analyse the effect of the magnetic field non-uniformity on the joint kinematics' evaluation, indoor gait trials were post-processed by following two procedures: (i) referring the sensors to the same ground CS_g, addressed as $\mathrm{IN_{NC}}$; and (ii) referring them to each specific ground, CS_{g_i}, addressed as $\mathrm{IN_C}$.

In the $\mathrm{IN_{NC}}$ condition, we assumed ${}^{g_j}\mathbf{R}_{g_i} = \mathbf{I}$ in equation (4) for all of the sensors. In the $\mathrm{IN_C}$ condition, we computed the actual ${}^{g_j}\mathbf{R}_{g_i}$ as reported in equation (6).

The CS_{b_i} and joint angles were estimated according to Wu *et al* (2002); the chosen Cardan sequence was ZXY. The trials were partitioned in gait cycles and the first 60 gaits of each trial

Table 1. Standard deviation of the offset (SD-O) of joint angles during the first 60 gait cycles of each trial, also expressed as a percentage of the mean ROM obtained outdoor, for the three conditions: outdoor (OUT), indoor non-corrected (IN_{NC}), indoor corrected (IN_C).

		OUT		IN_{NC}		IN_C	
		(°)	(%)	(°)	(%)	(°)	(%)
Hip	Flex/Ext	3	8	3	8	3	8
	Abd/Add	4	23	4	23	4	23
	Int/Ext rot	6	36	12	76	12	76
Knee	Flex/Ext	6	9	6	9	6	9
	Abd/Add	4	27	3	22	4	27
	Int/Ext rot	7	32	11	50	13	56
Ankle	Flex/Ext	5	14	4	13	5	14
	Abd/Add	6	26	5	24	4	20
	Int/Ext rot	12	47	18	70	13	52

were evaluated for the three joint angles: the mean value, addressed as offsets (O), and the range of motion (ROM).

We chose, as angular repeatability indices, the standard deviation of the above-mentioned quantities (SD-O and SD-ROM), expressed as an absolute value and as a percentage of the respective outdoor ROM. The above-defined parameters represent the repeatability of the offset and ROM due to the combined effect of subject gait variability and measurement system repeatability. The adopted experimental methodology does not allow us to separate the contributions of the subject and the MIMUs in the observed overall data spread. However, assuming that the subject keeps constant his walking variability indoors and outdoors, any change in the data spread may be related to the effect of magnetic field distortions. Therefore, the results discussed in the present paper mainly focus on the differences between indoors (with or without the correction of the heading) and outdoors, instead of examining each trial condition in absolute terms.

In addition, the SD-O and SD-ROM were also computed into a moving window of five consecutive strides (one stride shifted) of each one of the eight trials and each one of the four sessions (i.e. $4 \times 8 \times 5$ strides), from the start of the walking trial in order to evaluate the stability of the kinematic indices during the trials. Repeatability indices were also analysed as a function of the gait starting position.

4. Results

The typical outcomes of the body-to-sensor calibration procedure is reported in figure 5, showing the patient stick diagram and the reconstruction of the anatomical and MIMU reference frames obtained via OS.

The average values of the absolute heading correction occurring among all the four sessions were 3°, 5° and 15° for hip, knee and ankle joints, respectively; the heading correction reached a maximum value of 25° at the ankle level.

Table 1 shows the SD-Os in OUT, IN_{NC} and IN_C, and for hip, knee and ankle. The sagittal plane always showed a lower SD-O in comparison with the other anatomical planes. In particular, in the sagittal plane, the highest SD-Os were obtained for the ankle, with a maximum value limited to 4°–5° (13–14%). In the sagittal and frontal anatomical planes, no noticeable differences were observed between OUT and IN_{NC}. In the transverse plane, instead,

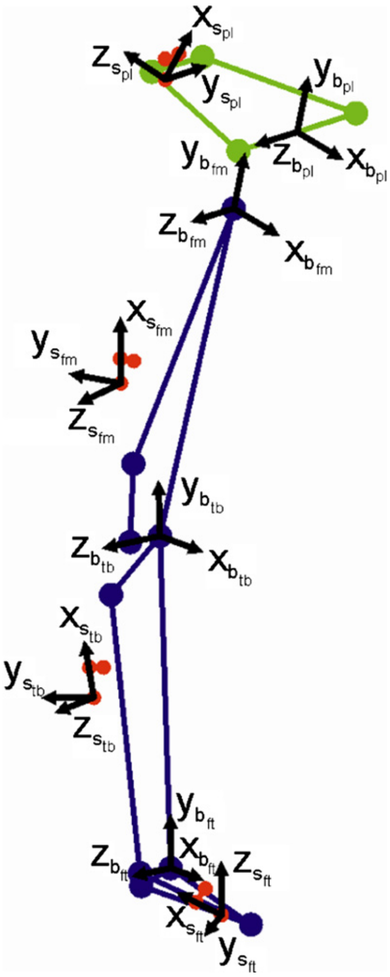

Figure 5. A typical outcome of the body-to-sensor calibration procedure performed via the OS: anatomical and MIMU reference frames. Dots represent markers (both the reflective ones and the virtual ones, reconstructed by VICON software for the joint rotations centres). Adjacent points are connected by means of lines to facilitate the interpretation of the figure. Colours indicate: green for the pelvis area; blue for the right thigh, shank and foot; red for the reflective markers placed on the MIMUs. The coordinate systems CS of each segment are depicted in black.

the SD-O increased noticeably from the first to the second of the above-mentioned conditions. The IN_C condition was able to reduce SD-O only for the ankle Int/Ext rotation from 70% to 52%.

SD-ROMs are reported in table 2. Knee and ankle SD-ROM showed the highest values in the frontal plane. For each joint, the lowest values occurred in the sagittal plane, where the highest one was for the ankle and was limited to 4° (12%). The SD-ROMs in IN_{NC} and OUT were comparable, with the exception of the ankle Abd/Add, where we obtained higher values in IN_{NC} than in OUT, i.e., 11° (49%) versus 7° (30%). The IN_C condition permitted the reduction of the SD-ROM only for the ankle in the frontal plane from 11° (49%) to 8° (34%).

Computing the moving standard deviation (figure 6), an increasing trend of the SD-O for Int/Ext rotation was observed for all joints, both IN_{NC} and IN_C. In the OUT condition, the

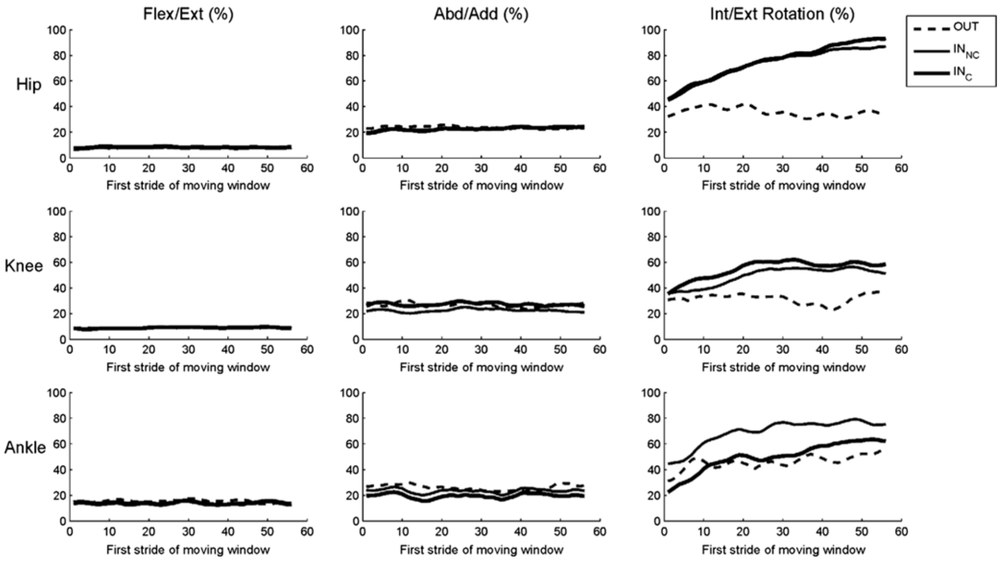

Figure 6. Standard deviation of the offset (SD-O) computed on a moving window of five consecutive strides of each trial and each session (one stride shifted).

Table 2. Standard deviation of the ROM (SD-ROM) of joint angles during the first 60 gait cycles of each trial, also expressed as a percentage of the mean ROM obtained outdoors, for the three conditions: outdoor (OUT), indoor non-corrected (IN_{NC}), indoor corrected (IN_C).

		OUT		IN_{NC}		IN_C	
		(°)	(%)	(°)	(%)	(°)	(%)
Hip	Flex/Ext	2	6	2	6	2	6
	Abd/Add	2	15	2	15	4	23
	Int/Ext rot	3	16	2	10	2	10
Knee	Flex/Ext	3	5	3	5	3	5
	Abd/Add	8	52	7	44	9	55
	Int/Ext rot	5	20	4	19	4	19
Ankle	Flex/Ext	4	12	4	12	4	12
	Abd/Add	7	30	11	49	8	34
	Int/Ext rot	6	24	7	27	6	24

increasing trend in the transverse plane was observable only for the ankle, while the SD-O of the hip and knee seemed to be not steady but also not increasing. The Flex/Ext and Abd/Add did not show a dependence on the trial duration. The SD-ROM increased with the trial duration for the ankle Abd/Add in IN_{NC} (figure 7) and the IN_C condition limited in part that tendency. In OUT, instead, the SD-ROM was not related to the trial time.

Figure 8 shows the SD-O as a function of the starting position of the trials performed inside the lab. Indoor values were greater than the outdoor ones only in the transverse plane. The highest SD-O values for the Int/Ext rotation of the ankle occurred at the points 3–5, where the magnetic field distortions were more noticeable (see figure 1). The heading correction provided a slight improvement limiting the data spread, as a tendency, with values lower than the outdoors ones. For the hip and knee joints, the influence of the starting point was not evident and the effect of the correction of the heading was not significant.

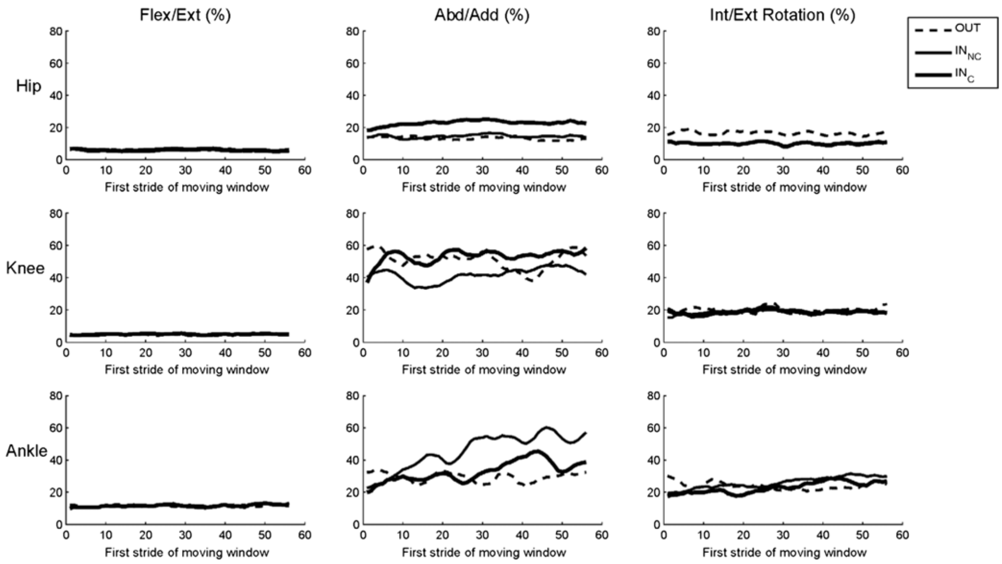

Figure 7. Standard deviation of ROM (SD-ROM) computed on a moving window of five consecutive strides of each trial and each session (one stride shifted).

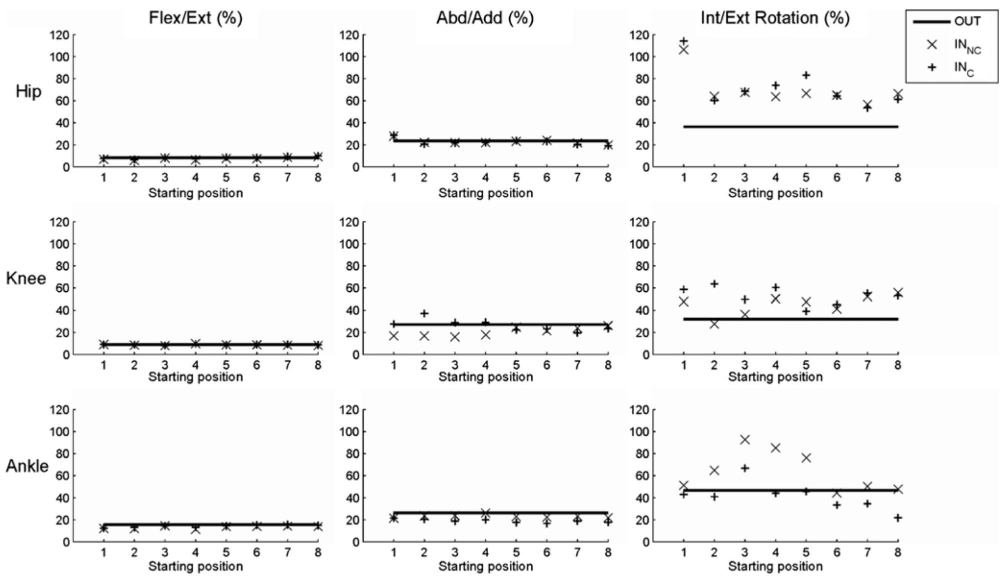

Figure 8. SD-O plotted as a function of the starting position inside the lab for the two indoors different conditions IN_{NC} and IN_C. The outdoor SD-O is drawn as a solid line.

The SD-ROM as a function of the starting position is depicted in figure 9. The sagittal plane appeared again to be immune from the measured distortion of the indoor magnetic field. Some effects were observable, instead, for the ankle rotation in the frontal plane. In this case, the distortion in points 3–5 (see figure 1) induced a noticeable increase in the ankle SD-ROM for the Abd/Add. The IN_C condition in the frontal plane limited the SD-ROM only for the ankle, and a slight increase was also observed for the hip.

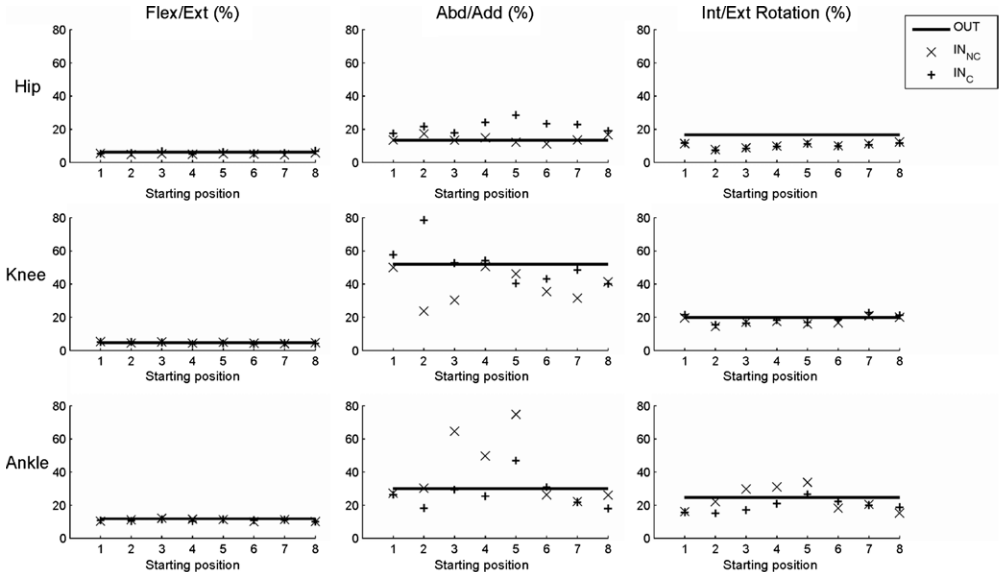

Figure 9. The SD-ROM plotted as a function of the starting position inside the lab for the two indoors different conditions IN_{NC} and IN_C. The outdoor SD-ROM is drawn as a solid line.

5. Discussion

The present work is focused on the evaluation of the effects of magnetic field distortion on the measurement of lower limb kinematics performed with a commercially available MIMU system. For the sake of clarity, we have divided the discussion in four subsections: firstly, we compare the indoor/outdoor conditions; secondly, the effect of the heading correction is examined; then, we discuss some aspects of the results not related to the influence of magnetic field distortion; and, finally, we explore the study limitations.

5.1. Evaluation of the effects of magnetic field distortions in data repeatability: IN_{NC} versus OUT

As reported in the results section, the SD-Os and SD-ROMs in the sagittal and frontal planes were comparable between outdoors and indoors, with the exception of the ankle SD-ROM, which increased in the frontal plane for the indoor condition. This increase was due to the high magnetic field distortion detected locally by the MIMU placed on the foot, and it indicates that the non-uniformity of the field noticeably affected the rotation in the frontal plane only at the floor level. Moreover, the orientation error, originated by the MIMUs, did not equally propagate to the three anatomical planes, probably due to the Cardan sequence chosen for the kinematic computations.

In the transverse plane, the SD-O showed noticeable differences between outdoors and indoors, because MIMUs use magnetic field information to evaluate the sensor heading. During walking, in fact, the Int/Ext rotation axes of body segments are almost parallel to the vertical axis and, consequently, the sensor heading affects mostly the Int/Ext rotations rather than the Flex/Ext and Abd/Add ones. However, the reliability of the Int/Ext rotation measured during walking is limited when the gait analysis is performed by means of an OS, because of the low repeatability observed among the protocols used (Ferrari *et al* 2008), especially where the actual ROM is small and comparable with the noise induced by the whole measurement procedure. On the contrary, the SD-ROM in the transverse plane did not show noticeable differences between outdoors and indoors. The increase of the SD-O, in contrast to the SD-ROM, implies that magnetic field distortion affected the offset of joint angles and not the waveform recorded. This effect was due to the characteristics of the MIMU's data fusion algorithm, which was optimized as a function of the chosen Kalman filter scenario. The selection of the scenario, (called by the MIMU manufacturer 'human-large acceleration' set), consisted of a set of weight parameters for accelerometers, gyroscopes and magnetometers capable of compensating for the magnetic field distortion occurring in a time length comparable to the human stride. As a side effect, instead, the measurement of the heading obtained with the magnetometer produced a slow drift, called random walk, in a longer time length, of the Int/Ext joint rotation. Therefore, that drift affected the mean value of each stride, i.e., the offset, more in comparison to the ROM.

The observed trend of the SD-O and SD-ROM obtained with the moving standard deviation confirmed all the previous mentioned outcomes. In fact, the Int/Ext rotation showed, indoors, an offset drift for all of the joints, which, outdoors, was slightly observable only for the ankle. The drift, however, did not affect the ROM value, which remained always comparable with the outdoor one, except for the Abd/Add of the ankle.

5.2. Evaluation of the effects of the heading correction in data repeatability: IN_C versus IN_{NC} and IN_C versus OUT

The heading correction performed at the start of the trials did not improve the repeatability associated with the offset and with the ROM. Only the ankle Int/Ext SD-O decreased noticeably, from 70% (18°) to 52% (13°), but the residual error, even if close to the value computed outdoors, indicates a poor level of repeatability in the transverse plane. Evidently, the 50% repeatability, obtained at the ankle, is intrinsic to the measurement system, and *de-facto* it makes the correction of the heading useless.

Moreover, even if the heading was corrected, the offset error associated with the Int/Ext rotation in the IN_{NC} condition was characterized by (i) a noticeable drift for all of the joints, (ii) a high value since the start of the walk. These observations confirmed the limited effectiveness of the heading correction procedure and the consequent unfeasibility of the MIMU system in the estimation of the offset in the Int/Ext rotation patterns both over short and long periods of time. The observed trend of the SD-O and SD-ROM as a function of the starting position confirms that the repeatability was only slightly dependent on the initial level of the magnetic field distortion. In fact, IN_C and IN_{NC} exhibited the same repeatability level as OUT for the hip and knee, even with the correction of the heading, since the pelvis, thigh and shank move in less-distorted field areas than the foot does. As regards the ankle, instead, the error due to the initial stabilization of the Kalman filter to the local direction of the magnetic field was more evident in correspondence to the centre of the laboratory room, where two force platforms were camouflaged into the floor. Moreover, in that area, the heading correction had a noticeable effect on the repeatability of the offset and ROM measured in the transverse plane. In addition, whereas the filter stabilization affected in a significant manner the Int/Ext rotation, especially for the offset, the ankle ROM in the Abd/Add plane was also influenced. That behaviour can be justified considering that an error occurring in the estimation of a joint angle in one plane propagated itself to the whole Cardan decomposition of the rotation. The magnetic field distortion effect, in fact, acts like a rotation of the ground frame of the involved sensor (i.e. an apparent bias rotation of the sensor). For small values of this rotation, the final effect is mainly an offset of the joint curve in the transverse plane, with a small waveform distortion that does not produce a noticeable ROM error. Conversely, for higher values of the apparent sensor rotation, the actual body segment rotation in a plane is measured by MIMUs as decomposed also in other planes, affecting significantly the ROM estimation.

5.3. Data repeatability not related to magnetic field distortions

The present work was conceived to highlight the final effects of indoor magnetic field non-uniformity on gait analysis evaluation performed via a commercial MIMU system, rather than to assess the absolute accuracy of the system. The methodology, therefore, by comparing outdoor versus indoor results, has been designed with the main aim of estimating the influence of magnetic field distortions. The data repeatability observed outdoors, however, showed that the contribution of the only MIMU system to the overall kinematic repeatability, irrespective of the uniformity of the magnetic field, was, in some conditions, not negligible.

In the frontal and transverse planes of the knee and ankle, in fact, it resulted that the SD-ROMs were similar outdoors versus indoors and were higher than the typical one for healthy adults (Steinwender *et al* 2000). Moreover, the outdoor ankle SD-O was about 47%, a high value that confirms the low repeatability of the here-adopted MIMU system in the evaluation of the rotations on that plane.

5.4. Study limitations

In the present research, only one subject was enrolled, however the obtained results can provide clinical users with an indication of the overall performance obtainable with MIMU systems in gait analysis. In fact, we studied only the variations of kinematics' repeatability between an indoor environment and an outdoor one, assuming that the subject gait repeatability remained unaltered. The differences found in terms of data spread, therefore, have only to be ascribed to the magnetic field distortion and not to the subject behaviour. Furthermore, the repeatability indices measured outdoors and indoors are not expected to change significantly with the inclusion of more subjects, in accordance with similar studies on different gait analysis protocols and/or laboratories' repeatability performed (Ferrari *et al* 2008). Finally, it is important to point out that the results are not extendable to motion analysis conducted on movements different from the gait and/or involving other joints, where the sagittal plane could move parallel to the horizontal plane.

6. Conclusion

In conclusion, the effect of magnetic field distortion, as we experienced in our gait lab, was limited to the transverse plane of each joint and to the frontal plane of the ankle. This implies that gait analysis on pathological subjects with locomotor disabilities affecting the rotations on the frontal plane for the ankle and on the transverse plane for the hip, knee and ankle, can be conducted with a device like the one we adopted, only taking into account a low repeatability of computed kinematic parameters. The here-measured kinematic variables in the sagittal plane, instead, were sufficiently immune from magnetic distortion effects, and did not show a decrease in repeatability moving from outdoors to indoors, whether or not an initial heading correction of the MIMU sensor here-adopted was performed. Moreover, the heading correction here-performed was not able to significantly improve the repeatability of the adopted MIMUs.

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