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An alignment procedure for ambulatory measurements of lower limb kinematic using magneto-inertial sensors

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Abstract— **In this work, an alignment procedure of magneto-inertial units in the Special Orthogonal Space SO(3) is presented and discussed. The procedure, designed for ambulatory measurements of lower limb kinematic, is based on simple rotation movements around anatomical axes of hip joint and its accuracy is independent of the speed as well as the range of the movements. This is particularly important for movement analysis of subjects with motor impairments.**

Despite such procedure was designed for lower limb movement analysis, it can be applied to every anatomical compartment (e.g.: upper limb).

I. INTRODUCTION

Motion tracking can count on a host of different technological solutions, operating on entirely different physical principles, with different performance characteristics and designed for different purposes. As shown in [1], there is not a single technology that can fit all needs. Each application defines the best technology to be implemented.

The objective assessment of gait is one of the field most studied. First technological approach to gait analysis are date back to 19th century when Muybridge studied the horse galloping using a set of cameras placed in line, each one triggered by a thread as the horse passed [2]. In the same period a French physiologist, Étienne-Jules Marey tried to apply a new technique called cronophotography to study human walking. It was based on multiple exposure of the same glass plate so that all the movements could be analyzed on the same print. Although several years are passed, optical systems are still very used for gait analysis. Optoelectronic Stereophotogrammetry (SP) for example, tracks a subjects movement through at least 2 cameras (usually three or more) and a set of reflective markers on the body of the subject. These systems, by means the triangulation principle can provide three-dimensional position of body segments with both high accuracy and reliability, but they suffer from severe limitations of applicability in clinical environment mainly due to high costs and complexity of the technology which needs high structured and dedicated Work Environments (WE).

A compromise between the necessity to have accurate kinematic reconstruction and ecological WE is represented by wearable Inertial/Magnetic Units (IMU). These systems integrate data from accelerometers, gyroscopes and magnetometers to estimate 3D orientation in the WE with respect to a fixed reference frame [3]. They are able to operate in little- or not-structured WEs, are small enough to be easily worn, and low-cost. On the other hand, they are generally less accurate than SP systems for position measurements [8] and cannot be used in the proximity of ferromagnetic-objects or devices emitting static magnetic field. Considering every body segment involved in the motion as a rigid link of known dimensions, it is possible to reconstruct a stick diagram of the body only knowing the orientation of each link at any time. Through this simplification, with one IMU on each segment, the motion of the entire kinematical chain with respect to a common reference frame can be tracked. To define such common frame several approaches have been used based on initialization movements [4]-[7]. The accuracy of such procedures are usually dependent by Range of Motion and rate so they are not suitable to study movement of subjects with motor disabilities. In this work we propose the use of the least-square algorithm developed by Park and Martin [9] and verify its dependency from Range of Motion.

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II. MATERIALS AND METHODS

The natural configuration space for a rigid body is the so-called Special Orthogonal group $SO(3)$, i.e. the space of 3×3 rotation matrices R such that $R^{-1} = R^T$ and $\det R = +1$. In static conditions, calibrated data from accelerometers and magnetometers are sufficient to determine the orientation R with respect to a global fixed frame defined by gravity and by the geomagnetic North [3]. In dynamic conditions, especially at higher frequencies, this estimate is much less reliable and is traditionally fused with information from the gyroscopes for robust attitude tracking by means of complementary and Kalman filters (see [10]-[13]).

When measurements coming from different IMUs are used, a key issue is to determine the initial misalignment between reference frames of each IMU used, in order to refer the all measurements to a common fixed reference frame.

Considering two sensors, one on the thigh and one on the shank. Let's A_i and B_i the sequences of rotation matrices respectively of the first and second sensor with respect to their initial orientation. To refer their measurement to a common reference frame, it is necessary defining their initial misalignment X (see Fig. 2). This issue is classically denoted as:

$$A_i X = X B_i \tag{1}$$

Since the sequences A_i and B_i are derived from noisy data, the solution should be found in a least-squares sense. Park

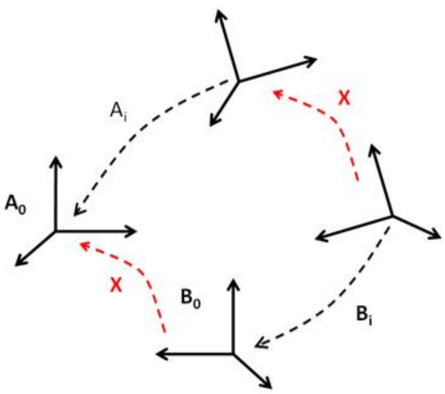

Fig. 1. $AX = XB$ problem: A_0 initial orientation of sensor A; B_0 initial orientation of sensor B; X misalignement matrix

and Martin [9] showed that

$$X = (M^T M)^{-\frac{1}{2}} M^T \tag{2}$$

where $M = \sum r_{A_i} r_{B_i}^T$, $\widehat{r}_{A_i} = \log A_i$, $\widehat{r}_{B_i} = \log B_i$ and the logarithmic map on $SO(3)$ is reported in (A.2).

To verify the possibility to use Park and Martin algorithm to this specific application we use two Xsens MTx sensors embedding 3D gyroscope, accelerometer, and magnetometer. The manufacturer reports a static accuracy of $0.5\,^\circ$ for roll and pitch, $1\,^\circ$ for yaw, and a $2\,^\circ$ RMS (Root Mean Square) dynamic accuracy. The two sensors were fasten to the thigh and shank of a subject and their misalignment measured by means of micro-metric positioning system produced by NEWPORT Ltd, with resolution of $0.1\,^\circ$. The subject knee was blocked and subject was asked to perform three kinds of movements:

1) three rotations on all three axes of the hip joint approaching the maximum ROM allowed by the hip joint itself;
2) three rotations on all three axes of the hip reaching half the maximum ROM allowed by the hip joint;
3) only three intra-extra rotations of the thigh.

Each sequence was repeated 10 times in order to have ten estimates.

For each sequence a X_{stim} was calculated according to (2) and compared with the known X to verify dependency of the Park and Martin Algorithm by Range of Motion.

Finally, a comparison between data coming from IMUs aligned with the proposed sequence of movements and a magnetic motion capture system (liberty, Polhemus) with 0.15 deg RMS static accuracy has been carried out to verify the accuracy of the proposed procedure.

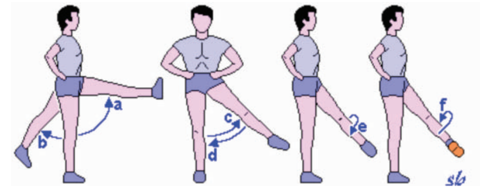

Fig. 2. Calibration movements: *a)* hip flexion; *b)* hip extension; *c)* hip abduction; *d)* hip adduction; *e)*;intra-rotation *f)* extra-rotation

III. EXPERIMENTAL RESULTS

To compare X_{stim} with the known X a distance d, or orientation error, has been defined. According to [14] we define such distance as the shortest single-axis rotation, or minimal geodesics, connecting X with X_{stim} :

$$d(X; X_{stim}) := acos(\frac{trace(X^T X_{stim}) - 1}{2}) \tag{3}$$

The distance was calculated for each trials. Its mean and Uncertainty Interval (UI) was reported in table I.

TABLE I

ERROR IN X ESTIMATION

Sequence	**Estimated X**	**±UI**
1	$\begin{pmatrix} 0.9988 & -0.0468 & 0.0129 \\ 0.0442 & 0.9868 & 0.1557 \\ -0.0200 & -0.1549 & 0.9877 \end{pmatrix}$	0.85 ± 0.20
2	$\begin{pmatrix} 0.9988 & -0.0396 & 0.0332 \\ 0.0307 & 0.9849 & 0.1702 \\ -0.0390 & -0.1690 & 0.9848 \end{pmatrix}$	0.68 ± 0.16
3	$\begin{pmatrix} 0.9898 & -0.0022 & 0.1423 \\ 0.0180 & 0.9899 & 0.1404 \\ -0.1412 & -0.1415 & 0.9798 \end{pmatrix}$	2.34 ± 0.66

Data show that a reduced range of motion does not significant affect the result of alignment procedure (see row 1 and 2 of TABLE I while it is important to perform rotation around the three anatomical axis to improve precision and accuracy of the algorithm.

The accuracy of the proposed alignment method was assessed comparing data coming from Xsens modules and data coming from a magnetic motion capture system (Liberty, Polhemus, USA) measuring knee flexion/extension. Polhemus trakers and Xsens modules were fasten to a wooden model of a human leg manually rotated at known positions. The knee physiological ROM during walking ($0\,^\circ$- $65\,^\circ$) was split in 13 intervals of $5\,^\circ$ each one. Ten repetitions of knee flexion were carried out for each interval. The angle of rotation was measured as:

$$R_B^A = (A_i)^T X B_i \tag{4}$$

$$\theta_x = \left\| \log(R_B^A) \right\| \tag{5}$$

see (A.2) for logarithmic map on $SO(3)$. The angle θ_x measured in (5) using Xsens data, was compared with the

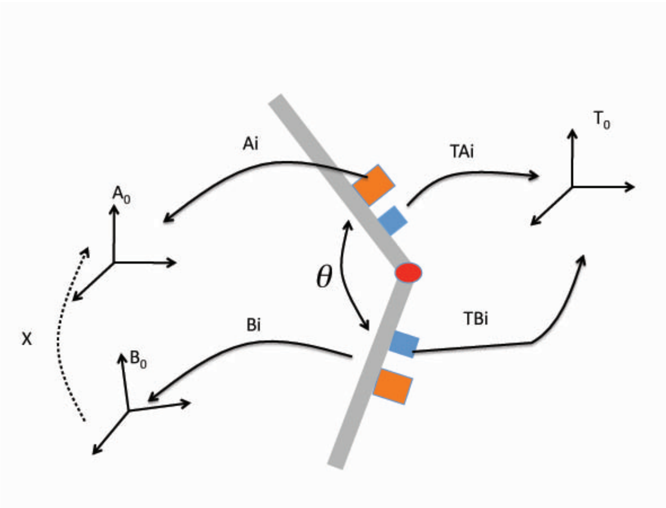

Fig. 3. Model for accuracy estimation and sensors displacement: in orange Xsens modules; in blue Liberty Polhemus Trackers

angle measured using Liberty data:

$$\theta = \left\|\log((TA_i)^T TB_i)\right\| \tag{6}$$

and error measured as:

$$e_i = \|\theta - \theta_x\| \tag{7}$$

For each interval mean error has been calculated. Results of this procedure are reported in TABLE II Mean static error

TABLE II
STATIC ACCURACY

Angular Displacement [°]	$Error \pm UI$ [°]
5	1.7 ± 0.1
10	1.8 ± 0.2
15	1.4 ± 0.2
20	2.6 ± 0.3
25	1.7 ± 0.7
30	1.8 ± 0.6
35	1.2 ± 0.3
40	0.9 ± 0.4
45	0.7 ± 0.3
50	0.7 ± 0.3
55	1.9 ± 0.4
60	2.2 ± 2.2
65	0.7 ± 0.2

in walking knee flexion range is $1.5 \pm 0.3°$ (c=0.95).

Using the same experimental setup ten knee flexion/extensions were carried out and error measured during movement .

Mean error estimated is $2.5 \pm 0, 1°$ (c=0.95), see TABLE III.

TABLE III
MEAN ERROR DURING MOVEMENT

Trial	$MeanError$[°]
1	2.6
2	2.2
3	2.6
4	2.2
5	2.8
6	2.7
7	2.6
8	2.6
9	2.4
10	2.7

IV. EXAMPLE OF IN-FILED USE

Although clinical validation is beyond the scope of this work, we shall here briefly provide an example of in-filed use of the proposed methodology with a healthy adult subjects. Subject was asked to wear two MTx sensors on their right leg: one on the thigh (A) and one on the shank (B). Each sensor was fasten to the subject's leg in order to avoid any relative movement, with the x-axis pointing forward, the y-axis pointing up and z-axis perpendicular to the sagittal plane. A full ROM calibration sequence (seq. 1, see sec. II) was performed to measure the misalignment matrix X. Subject was asked to walk for about 30 meters starting from a fully extended knee position with self-selected walking speed.The hip and knee flexion-extension angle was measured, according to [6], as the angle around the axis perpendicular to the sagittal plane respectively of matrix A_i and the matrix defined in (4). Results are shown in fig.4.

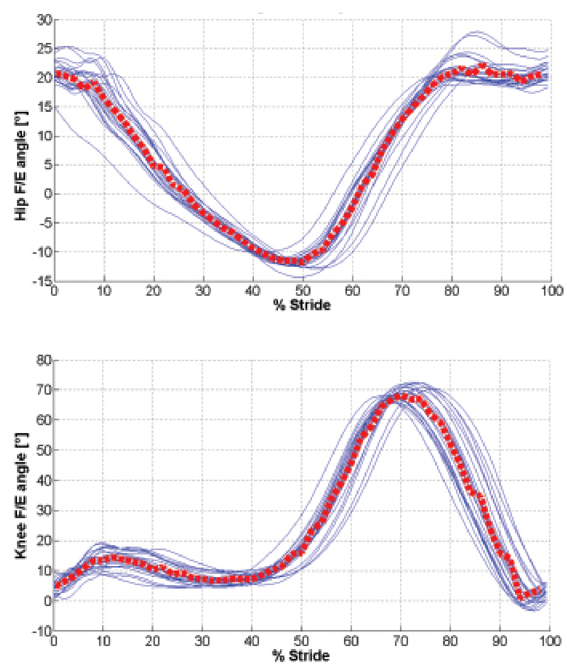

Fig. 4. Flexion/Extension angle of hip and knee joints: in blue the angle for each step; in red dotted line the mean angle.

V. CONCLUSIONS

In this work a calibration procedure for ambulatory measurement of limb kinematic using magneto-inertial sensors has been presented and discussed. A least-squares algorithm developed by Park and Martin has been used to assess initial misalignment between sensors used for kinematic reconstruction. The main advantage of this algorithm is its independency from the speed at which calibration movements are performed. To verify the effect of ROM on the calibration procedure a set of physiological movements with different ROM has been defined. A reduced ROM on the three main physiological axis of hip joint does not affect the calibration procedure. The angular error in knee flexion/extension was assessed comparing data coming from commercially available IMUs calibrated with the proposed procedure with data gathered by means of a magnetic motion capture system. Finally an example of in-field use of the proposed methodology was presented.

VI. ACKNOWLEDGMENT

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APPENDIX

The *Hat operator* maps a vector $a = [a_1 \ a_2 \ a_3]^T$ into a skew-symmetric matrix:

$$\widehat{\cdot} : a = \begin{bmatrix} a_1 \\ a_2 \\ a_3 \end{bmatrix} \longrightarrow \begin{bmatrix} 0 & -a_3 & a_2 \\ a_3 & 0 & -a_1 \\ -a_2 & a_1 & 0 \end{bmatrix} = \widehat{a} \quad \text{(A.1)}$$

The *Logarithmic map* [9] on $SO(3)$ maps a rotation matrix into a skew-symmetric one:

$$\widehat{r} = \log R := \frac{\theta}{2\sin\theta}(R - R^T) \quad \text{(A.2)}$$

where θ satisfies $1 + 2\cos\theta = \mathtt{trace}(R)$. The physical significance of this map is that any rotation R can be thought of as a pure rotation about a fixed axis r through an angle $\|r\| = \theta$.

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