The Development and Test of a Device for the Reconstruction of 3-D Position and Orientation by Means of a Kinematic Sensor Assembly With Rate Gyroscopes and Accelerometers

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***Abstract*—In this paper, we propose a device for the *P*osition and *O*rientation (P&O) reconstruction of human segmental locomotion tasks. It is based on three mono-axial accelerometers and three angular velocity sensors, geometrically arranged to form two orthogonal terns. The device was bench tested using step-by-step motor-based equipment. The characteristics of the six channels under bench test conditions were: crosstalk absent, non linearity $< \pm 0$, 1% fs, hysteresis < 0, 1% fs, accuracy 0, 3% fs, overall resolution better than 0, 04 deg/s, $2 * g * 10^{-4}$. The device was validated with the stereophotogrammetric body motion analyzer during the execution of three different locomotion tasks: stand-to-sit, sit-to-stand, gait-initiation. Results obtained comparing the trajectories of the two methods showed that the errors were lower than $3 * 10^{-2}$ m and 2 deg during a 4s of acquisition and lower than $6 * 10^{-3}$ m and 0.2 deg during the effective duration of a locomotory task; showing that the wearable device hereby presented permits the 3-D reconstruction of the movement of the body segment to which it is affixed for time-limited clinical applications.**

***Index Terms*—Accelerometer, accuracy, human movement analysis, simulation, 3-D rigid body position and orientation.**

I. Introduction

THE traditional approach used in movement analysis is based on the measurement of the position of markers affixed on body segments by means of optoelectronic technology [1]. Similarly this result can be achieved by means of ultrasound technology based on ultrasound emitting point-markers and microphonic sensors [27]. The main limitations of these techniques are the encumbrance and costs of the equipment. As alternative to the optoelectronic/ultrasound approach, in the seventies both Padgaonkar and Morris showed the potential of the accelerometric techniques. The first author introduced an analytical model aimed at reconstructing the three-dimensional (3-D) position and orientation (P&O) of a body segment using the signal provided by nine accelerometers for the angle measurement of the rigid body [2], the second author introduced the use of accelerometers in the assessment of human movement by means of an analytical model aimed at reconstructing the 3-D P&O of a body segment using the signal provided by six accelerometers [3]. The accelerometers are small and light enough to construct 3-D markers that can be rigidly and easily attached to a body segment without interfering with the execution of the physical exercise. The major drawback of accelerometric systems is due to the acceleration of gravity g, a fraction of which is measured by accelerometers, depending on their orientation as showed in our previous work [8]. The intrinsic limit in the estimation of orientation entails a significant limit in the correction of the sensor output to measure acceleration, and by applying appropriate integration algorithms, the relevant Positioning and Orientation vector. In [8] we also simulated devices which used 6 or 9 accelerometers [2], [3] used for the P&O trajectory reconstruction and showed that errors in the estimation of orientation and position are 10° and 0, 3 m, respectively, for an observation interval of one second; here we show that the predominant cause of error was due to the sensitivity of accelerometers to gravity. There are other kinematic sensors such as rate-gyroscopes which have the advantages to be insensitive to the influence of gravity. Wun in [12] showed an assembly comprised of three rate-gyroscopes, three accelerometers and optoelectronic markers used in a clinical application of reconstruction of body-center of mass acceleration. The use of kinematic sensors just like the other motion analysis methodologies is amply dependent on both the development of clinical applications and the development of microelectronic technology. Recently a growing interest for noninvasive patient monitoring has promoted a large development of portable/wearable sensors and systems as showed in the recent issue of the IEEE Engineering in Medicine and Biology magazine edited by Paolo Bonato [14]. The wonderful development of the Microelectronics thanks to the full custom, large scale and micro technologies now permits performances in layout miniaturization and precision never reachable before, with large perspectives for the implementation of assemblies based on kinematic sensors useful for the portable monitoring at a low cost. Accelerometers, inclinometers, and gyroscopes alone or combined have been increasingly used for the realization of portable sensors for the acceleration and orientation monitoring [15]–[17]. Recently the kinematic sensors have also been introduced in clinical applications where postural parameters are processed for biofeedback restitution to vestibular patients. Wall, for example, developed a balance prosthesis for the vibrotactile feedback [29] based on an inertial sensor composed

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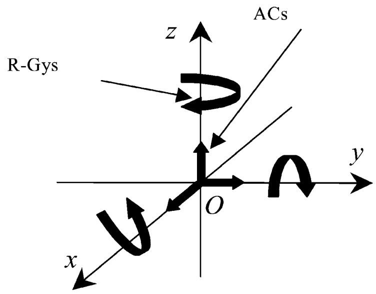

Fig. 1. Relative orientation of the sensors: 1, 2, 3 accelerometers; 4, 5, 6 gyrostars.

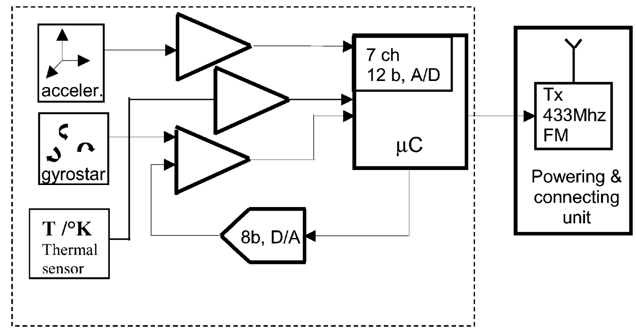

Fig. 2. Architecture of the wearable device.

of a triaxial accelerometer and gyroscope which furnished the head tilt as control variable to a matrix of vibrators worn by the subject. The growing interest in these technologies is documented by the fact that industries led by the low cost are recently developing solutions based on these sensors; see for example the solutions MT9-A furnished by the Xsens (Xsens, Enschede The Netherlands) [18] which realized a motion sensor based on three angular rate sensors three accelerometers and three magnetic sensors which by means of a dedicated algorithm reconstruct an angular displacement. Another important industrial solution is the system Is300pro (Intersense, Atlanta, GA) [28] based on inertial sensors, which furnishes a precise orientation reconstruction for virtual reality applications. All of the most commonly used kinematic sensors [19], [20] have a relative low cost if compared to optolectronic/ultrasound solutions [21]. All these solutions found in literature did not completely afford the problem of feasibility reconstruction range of P&O vector. In this article we developed a device based on an architecture of three rate gyroscopes and three accelerometers and performed a bench test and a clinical validation for the P&O reconstruction of locomotion tasks typical of human motion analysis.

II. Materials and Methods

A. Algorithms

The transducer consists of three mono-axial accelerometers (3031-Euro Sensors, USA) and three sensors of angular velocity (Gyrostar ENC-03J-Murata, Japan), assembled together and relatively oriented according to an orthogonal reference system. Fig. 1 shows the relative orientation of the sensors.

The actual body segment angular velocity vector $(\omega_x, \omega_y, \omega_z)$ is obtained by multiplying the relevant calibration matrix by the gyrostar tern output vector. The orientation of the segment is determined by the orientation matrix $[R]$, which is calculated by solving the following differential matrix-(1) [13], and then the three orientation angles (expressed in nautical angles) are obtained using (2) [13], see (1)–(2) at the bottom of the page.

It was found convenient to express the segment orientation in nautical angles: θ is the pitch angle, φ is the roll, ψ is the yaw angle.

The real instantaneous linear acceleration vector is obtained by (3).

$$\begin{bmatrix} ax \\ ay \\ az \end{bmatrix} = [R] \begin{bmatrix} ax' \\ ay' \\ az' \end{bmatrix} - g \tag{3}$$

where ax' ay' and az' are the acceleration vectors in the reference system solid with the device. The instantaneous motor acceleration vector is then double integrated to reconstruct the trajectory. All the algorithms were developed by means of Matlab R 12 software package (The Mathworks, Natick, MA). In particular the algorithm used to solve (1) and (2) was developed with the Simulink Tool box; the differential equation system, for the calculation of the orientation matrix, was solved by means of the ordinary differential equation system solver (ODE). The function *quadl()* we used for the integration is based on the adaptative Lobatto quadrature algorithm, with absolute tolerance error in the integral set to 10–9; the use of this function permits a more precise and fast computation than the use of *quad()* function, based on the adaptive Simpson quadrature algorithm. Another algorithm had to be integrated to all described measurement and computing algorithms to minimize the thermal drift of the gyrostars. The zeroing algorithm is based on a specific Thermal sensor (lm335-National Semiconductor, USA) and a tuning table compiled at different temperatures using a controlled oven to correct the offset-drift during calibration.

B. Device Construction

Fig. 2 shows the architecture of the wearable device. It is composed of two separate units: a sensor unit and a powering,

$$R^{-1*}\frac{dR}{dt} = \begin{bmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{bmatrix} \tag{1}$$

$$R = \begin{bmatrix} \cos(\phi)\cos(\theta) & \cos(\phi)\sin(\theta)\sin(\psi) - \sin(\phi)\cos(\psi) & \cos(\phi)\sin(\theta)\cos(\psi) + \sin(\phi)\sin(\psi) \\ \sin(\phi)\sin(\theta) & \sin(\psi)\sin(\vartheta)\sin(\varphi) - \cos(\phi)\cos(\psi) & \sin(\phi)\sin(\vartheta)\cos(\psi) - \cos(\phi)\sin(\psi) \\ \sin(\theta)\sin & \cos(\theta)\sin(\psi) & \cos(\theta)\cos(\psi) \end{bmatrix} \tag{2}$$

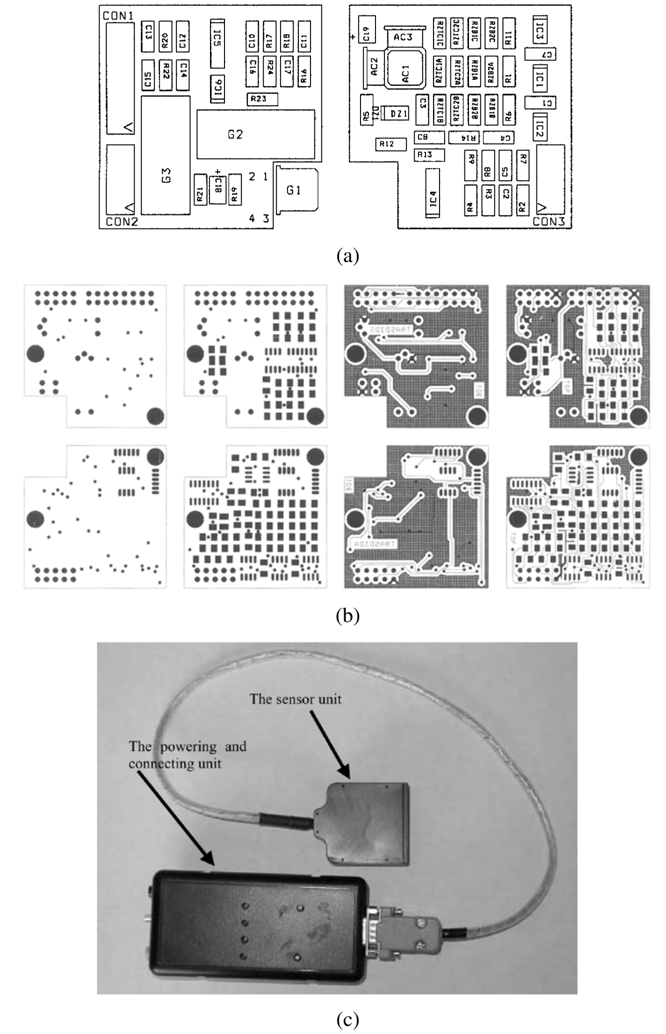

Fig. 3. (a) Optimal components arrangement. AC1, AC2, AC3 are accelerometers, G1, G2, G3 are rate gyroscopes. (b) The layers for the board manufacturer. (c) The complete device with evidentiated the sensor unit and the powering and connecting unit.

connecting unit. The latter contains the battery and the circuitry for communication at a 433 MHz \FM radio frequency.

The sensor unit was very small ($4 \times 5 \times 2$ cm). The sensor had a weight of 250 g, this mean that it is enough light for the applications of interest, in fact the weight is not different to the one of other wearable commercial sensors such as for example those for the daily cardiac monitoring; furthermore the duration of the specific clinical application conducted with this sensor is too short to fatigue the patient. The circuitry of the wearable consists of 7 signal conditioning chains (3 for the accelerometers, 3 for the gyrostars, 1 for the thermal sensor), a microcontroller that features a 12 b A/D converter and an 8 b D/A converter. The conditioning chains comprise seven amplifiers $(G = 10)$ and seven Voltage Controlled Voltage Sources (Sallen-Key cells) second-order filters with a Butterworth Low Pass Filter response at a cutoff frequency of 14 HZ, this frequency was obtained by means of P-Spice simulations.

Fig. 3(a) illustrates the placement and routing of the of the device components, with the standard circuit notation of components (IcXX are the integrated circuits, CXX are the capatacitances, RXX are the resistances, CONXX are the connectors).

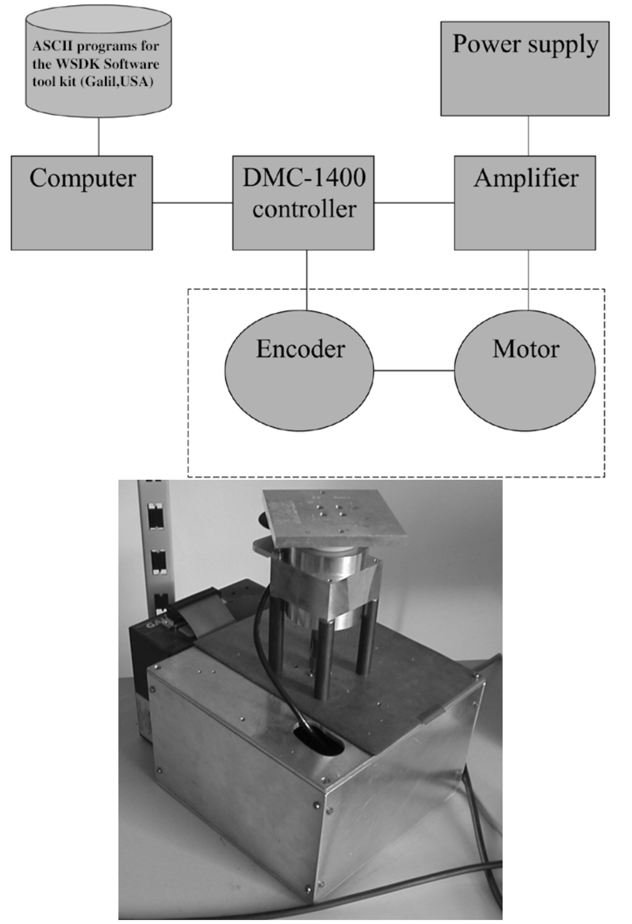

Fig. 4. (a) Complete architecture of the equipment for the calibration and generation of motion laws based on the DMC-1410 component (Galil, Rocklin, CA). (b) The complete equipment with the mechanical case and interface developed in our laboratory.

The sensor circuitry has been split in two separate boards: one for the gyrostar tern and the other for the accelerometer tern. The circuitry is assembled with surface montage technology, taking special care with the positioning of the sensors and their stability so as to ensure the proper functioning and reliability of the device; see Fig. 3(b) and (c).

C. Realization of a Dedicate Equipment for the Generation of Motion Tasks (Testing Equipment)

A dedicate equipment was developed in order to impose dedicate motion laws for calibration and for the bench test. The Core of the system were the DMC-1410 controller and step-by-step motor with encoder (Galil, Rocklin, CA) [22]

These elements were integrated together with

— power supply ± 12 V;
— PC with ISA bus;
— communication disk and Servo Design Software WSDK (Galil, Rocklin, CA);
— Amplifier AMP-1460.

The complete architecture is showed in Fig. 4(a). The mechanical case and interface were designed in our Laboratory; the complete realized equipment is showed in Fig. 4(b)

The system characteristics were the following.

Torque constant $\mathrm{Kt} = 0, 1$ Nm/A, system moment of inertia $J = 2 * 10^{-4}$, motor resistance $R = 2\ \Omega$.

Current amplifier gain $K_a = 4\ \mathrm{Amp/V}$, encoder line density $N = 1000\ \mathrm{Counts/rev}$, sample period $T = 0.1\ \mathrm{ms}$. Preliminary tests conducted in the maximal bench test conditions indicated an error lower than $1 * 10^{-3}\ \mathrm{m}$ and $2 * 10^{-2}\ \mathrm{deg}$.

D. Calibration and Performance Evaluation Set-Up Test

The instrument was calibrated in two phases: one static, the other dynamic. The calibration of the accelerometer tern was carried out during the static phase. The sensor unit was subjected to different g vectors by positioning each of its six faces on a horizontal plane; the output signals were averaged over a 6-s time interval.

The dynamic calibration was done with the Testing Equipment. A plate on which the device is affixed was rotated by means of the Testing equipment and to impose known rotational time laws with the necessary accuracy. Angular velocities ranging from 10° to 60°/s in both directions were imposed for each of the three orthogonal axes.

In both cases the calibration matrixes were computed by the *l*east *s*quares *m*ethod.

$[C_A]$ was the calibration matrix for the accelerometers channels obtained by means of (4), where $[L_A]$ was the matrix of the imposed quantities, which in this specific case were obtained when the sensor unit was subjected to different g vectors by positioning each of its six faces on a horizontal plane; $[S_A]$ was the matrix of the quantities assessed by the sensor unit, in the specific case g

$$[C_A] = [L_A][S_A]^T([S_A][S_A]^T) - 1. \quad (4)$$

$[C_{RG}]$ was the matrix for the rate-gyroscopes channels obtained by means of (5), where $[L_{RG}]$ was the matrix of the imposed quantities, which in this specific case were the angular velocities imposed by means of the dedicated equipment when the sensor unit was rotated with different angular velocities and with six different orientation, $[S_{RG}]$ was the matrix of the quantities assessed by the sensor unit, in the specific case the different angular velocities

$$[C_{RG}] = [L_{RG}][S_{RG}]^T([S_{RG}][S_{RG}][S_{RG}]^T) - 1. \quad (5)$$

Rotational and translational laws were imposed in a bench test in the range of the locomotor tasks (0.1, 1 Hz frequency; 0°, 60° rotational amplitude; 0.1, 0.5 m translational amplitude).

For validation purposes the instrument was used together with a classical stereophotogrammetric body motion analyzer VICON [21] with a field of view of six cameras, 1000 Hz of sample rate. Three markers at a known distance were affixed to the portable device choosing three non collinear positions because the knowledge of the position of at least three noncollinear points allows for the reconstruction of the P&O of the rigid body [23]; we avoided shadowing effects. We preferred to introduce a very simple preliminary test to account for the variability of environmental parameters which could increase the errors declared by the manufacturer in the specifics. The device was arranged on a fixed position on the extreme point of a pole (length 1 m) with an extreme vinculated to the ground by a rotative joint; a subject rotated the pole with a hand to preliminary determined constrained positions; the error of the VICON system in a rotation of 90°, with a trajectory length of $0, 5 * \Pi * \mathrm{m}$ was negligible, less than $5 * 10^{-4}\ \mathrm{m}$ and $10^{-2}\ {}^\circ$.

The sensor [see Fig. 3(c)] device was placed on a volunteer's back at position L5 on the subject's back, taking as a reference the subject navel to estimate the trunk flexion, firmly attached by means of a belt with a rigid pocket, the powering and connecting [see Fig. 3(c)] was attached to the trousers at the right side. This position is very near to the body center of mass and it was chosen to test protocols suitable for the future comparison to the data obtained from a force plate. A trigger signal was used to synchronise the sensor and the VICON system. The tasks chosen as significative were the following:

- — stand-to-sit;
- — gait-initiation, asking the subject to perform only a step, starting and ending with the feet aligned;
- — sit-to-stand.

The trajectory comparison was performed in a 4-s interval in the sagitall plane to test the feasibility of the reconstruction of these locomotory tasks in the plane of interest. We also tested the system to the effective time of a duration of a sit-to-stand, we used two different dedicated algorithms for the determination of the start and end positions of the trajectory of the sit-to-stand described by Cappozzo in [24] and by Giansanti in [25].

E. Extraction of Noise Parameters and Preliminary Simulations

A preliminary test was conducted on a lot of accelerometers and gyrostars before starting the development of the device.

During this preliminary bench test the following important noise parameters were delivered for both kinematic sensors: random noise, sensitivity error noise, offset error, and thermal drift [8].

Accelerometers: By means of bench test trials described in [13], [16] it was possible to set as offset noise $0.5 * 10^{-2}\ \mathrm{ms}^{-2}$ as off-set noise, $1 * 10^{-2}\ \mathrm{ms}^{-2}$ as random noise e_r, the constant γ of the sensitivity error was fixed at $2 * 10^{-4}$.

Rate Gyroscopes: The offset due to the thermal drift of the gyroscope component (ENC-03J, Murata) was $5\ \mathrm{mV}/^\circ\mathrm{C}$ corresponding to $4, 5\ \mathrm{deg/s}/^\circ\mathrm{C}$; the temperature variation was $0.02\ ^\circ\mathrm{C/s}$. After the execution of the bench test of the component it was possible to fix 0.1 deg/s as random error e_r^{rgy} and $5\ \mathrm{deg/s}/^\circ\mathrm{C}$ as the offset error $e_{0,}^{\mathrm{gyr}}$ the constant γ^{gyr} of the sensitivity error was fixed at 10^{-3}.

These errors were given as inputs to software simulations to estimate the performances of the algorithms and device before starting the hardware implementation.

The simulation was performed as follows.

1) The technical data obtained from the datasheets of the two different sensor components were used to realize two software models by means of the simulink packages of MatlabR12 (The Mathworks, Natick, MA).
2) The Positioning error was set to $p_e = 0.5 * 10^{-3}\ \mathrm{m}$ and the orientation error $\alpha_e = 0.06\ \mathrm{deg}$, these are the P&O errors which can be obtained by means of an automatic Pick and Place Montage.
3) All of the other component of the circuit were simulated by means of P-Spice simulations.

All of these packages were integrated by means of the Matlab R12 Basic Toolbox.

The simulation was limited to the simulation of the following motor tasks.

Rotation 60°, frequency 1 Hz.

Translation 0, $5 * m$, frequency 1 HZ

The simulations were run with all of the error conditions described above. The drift compensation and effect of calibration were considered during simulations. One-thousand-trial series were run for each of the conditions investigated. In each trial, a random value was assigned to each error source in the range indicated. The number one thousand was chosen since a pilot study showed that further trials did not modify the results. These motion conditions intended to simulate movement typical in the short time locomotion acts used in the ability/disability evaluations starting and ending by still with the starting position initially defined (sit-to-stand, stand to sit, gait initiation, standing the stairs). These acts which can be divided into translations and rotations can be considered with a good approximation planar with the movement located in the sagittal plane [8], [30]; these acts can be conveniently investigated in this plane where are principally located [30].

A preliminary optimization study conducted by means of the simulations permitted also to fix to 14 hz the threshold of the cutoff frequency of the second order filter developed by means of a Voltage Controlled Voltage Sources Sallen & Key cell. This result was also proved during the bench test conducted by means of a frequency swap.

III. Results

A. Calibration

The characteristics of the six kinetic channels obtained during calibration were as follows:

— crosstalk absent for both channels;
— nonlinearity $< \pm 0,\ 1\%$ fs for both channels;
— hysteresis $< 0,\ 1\%$ fs for both channels;
— accuracy 0, 3% fs for both channels;
— overall resolution was better than 0, 04°/s for rate gyroscope channels and $2 * g * 10^{-4}$ for accelerometer channels.

By applying Student's T test on 100 trials per condition we proved the stability of the instrumentation with $1^{\circ/\circ\circ}$ significance.

B. Static Performances

Assessment of Thermal Drift

Measurement on the selected rate-gyroscopes showed a 4, 9 $\mathrm{deg/s/^{\circ}C}$ drift at 20 °C.

In the operative range the following empirical law was found for the thermal drift

$$\mathrm{TD}\ (\deg/s/^{\circ}\mathrm{C}) = +\mathrm{T}(^{\circ}\mathrm{C}){*}0,2 + \mathrm{T}(^{\circ}\mathrm{C})^2{*}0,002 + \mathrm{T}(^{\circ}\mathrm{C})^3{*}0,000\,012. \quad (6)$$

It was the predominant cause of errors for this component which was corrected by means of the tuning table compiled with a controlled oven.

TABLE I
SIMULATION OUTPUT WITH CONDITIONS: 1 HZ, 60 DEGREES FOR ROTATION; 1 HZ, 0.5 M FOR TRANSLATION, INTERVAL DURATION 1 S

condition	ME Angle error	Maximal angle error	SD Angle error	ME Displacement error	Maximal displacement error	SD displacement error
Rotation	0.4 deg	0.5 deg	0.08 deg	$2*10^{-3}$m	$3*10^{-3}$m	$0.5*10^{-3}$m
Translation	0.1 deg	0.2 deg	0.05 deg	$4*10^{-3}$m	$5*10^{-3}$m	$1*10^{-3}$m

TABLE II
SIMULATION OUTPUT AT 4 S WITH CONDITIONS: 1 HZ, 60 DEGREES FOR ROTATION; 1 HZ, 0.5 M FOR TRANSLATION, INTERVAL DURATION 4 S

condition	ME Angle error	Maximal angle error	SD Angle error	ME Displacement error	Maximal displacement error	SD displacement error
Rotation	1.4 deg	2 deg	0.2 deg	$8*10^{-3}$m	$12*10^{-3}$m	$2*10^{-3}$m
Translation	0.4 deg	0.7 deg	0.1 deg	$12*10^{-3}$m	$18*10^{-3}$m	$3*10^{-3}$m

TABLE III
BENCH TEST WITH CONDITIONS: 1 HZ, 60 DEGREES FOR ROTATION; 1 HZ, 0.5 M FOR TRANSLATION, INTERVAL DURATION 4 S

condition	ME Angle error	Maximal angle error	SD Angle error	ME Displacement error	Maximal displacement error	SD displacement error
Rotation	1.5 deg	2 deg	0.3 deg	$9*10^{-3}$m	$12*10^{-3}$m	$2*10^{-3}$m
Translation	0.5 deg	0.7 deg	0.1 deg	$13*10^{-3}$m	$18*10^{-3}$m	$3*10^{-3}$m

TABLE IV
CLINICAL VALIDATION FOR THREE DIFFERENT LOCOMOTION TASKS, INTERVAL DURATION 4 S

condition	ME Pith Angle error	Maximal pitch angle error	SD Pitch angle error	ME Displacement error	Maximal displacement error	SD displacement error
Sit-to-stand	1.3 deg	1.8 deg	0.4 deg	$27*10^{-3}$m	$32*10^{-3}$m	$4*10^{-3}$m
Gait initiation	0.8 deg	1.3 deg	0.3 deg	$22*10^{-3}$m	$27*10^{-3}$m	$3*10^{-3}$m
Stand-to-sit	1.2 deg	1.7 deg	0.3 deg	$26*10^{-3}$m	$31*10^{-3}$m	$4*10^{-3}$m

Residual Error in Evaluating Orientation: The results showed that the maximal error in the estimation of orientation, after drift correction follows the empirical relationship

$$\Delta\vartheta(\deg) = 0,02 + 0,03t(s)^* + 0,0001^*t(s)^2. \quad (7)$$

The error under static conditions was 0, 3 deg at the end of a 10-s observation period.

- The Student's T was performed on 25 trials. The two laws were found to have a confidence of $1^{\circ/\circ\circ}$.

C. Simulation, Bench Test, and Clinical Validation

Simulation: The simulation algorithm was tested for numerical errors by running the simulations with the error sources set at zero. Maximum errors were found lower than $0.5 \cdot 10^{-2\circ}$ and 10^{-5} m for position and orientation, respectively. In Tables I and II, simulation results at 1 s and 4 s are shown.

Bench Test: Table III indicates the values obtained in the maximal bench test condition in a measurement interval of 4 s:

— 1 Hz, 60° for rotation;
— 1 Hz, 0.5 m for translation.

*M*ean *E*rror, *S*tandard *D*eviation, and maximal errors are referred to 100 trials.

Clinical Validaton: Table IV indicates for each locomotion condition (stand-to-sit, sit-to-stand, gait initiation) *M*ean *E*rror, *S*tandard *D*eviation and maximal error of the pith angle and trajectory displacement in the sagittal plane over 25 trials conducted on healthy people, as estimated by the VICON analyzer. Measurements were performed for an interval of 4 s.

The comparison between Tables II and III showed the correspondence between experimental data and simulation data. As expected the errors in a 1 s interval were lower. Experiments indicated that applying the algorithms to the effective duration of the locomotion task the error decreases, in fact the application of the algorithm to a 1.2 s movement of sit-to-stand gave 0.2 deg as pitch angle error, $6 * 10^{-3}$ m.

D. Device Life, Costs, and Maintenance

The sensor unit is prototypal, the estimated future cost of an internal production of one unit is 1950 euros; 250 euros is the cost of 16 hours of work of a technician to assembly and test the device unit and 1700 euros is the cost of the electronic components, boards comprised. To improve the life of the device we provided it by a $3 * 10^{-3}$ m thick elastic neoprene shell; we dropped it 200 times from an high of 1.5 m during a bench test and it did not show failures after reusing. The number of 200 is adequate if we consider the rarity of the event. To extend the life of the sensor we also used components with the failure rate typical of the military specifics and conducted an obsolescence study of the critical components to plane a large production. A preliminary conservative stress-analysis study showed 20 000 hours of life of the sensor. We also estimated that the cost of labour of the replacement of an integrated component with SMD package is lower than 3 Euros, due to the very short mean time of the substitution, (about 10 min) thanks to the use of our MANTIX VISOR (Vision Engineering, USA) specific for smd montage.

We should also consider that the substitution of the core components (accelerometers and rate-gyroscopes) has a little chance, in fact they are long life coming from aereo-space technologies.

IV. Discussions and Conclusions

The discussion on the advantages/disadvantages of the use of kinematic sensors has been opened from the seventies by Padgaonkar and Morris [2], [3]. In previous work we demonstrated that architectures with sole accelerometers did not allow us an accurate reconstruction of the trajectories [8]. Wun [12] introduced the use of an integrated kinetic sensor composed of rate gyroscopes and accelerometers and multiple optoelectronic markers in the the study of kinematic transients in locomotion and shown that accelerometers and angular rate sensors increased the accuracy in the determination of the the center of mass acceleration in clinical applications. Before this paper the problem of the range of feasibility of the complete P&O trajectory reconstruction by means of kinematic sensor assemblies was not completely afforded. The architecture introduced was simulated developed and tested in applications of trajectory reconstruction typical of transitory locomotion tasks usable in the ability/disability studies. As showed in the analysis, the introduction of gyrostars in the algorithm significantly improved the trajectory reconstruction thanks to their insensitivity to gravity but it was necessary the design and the introduction of a real-time algorithm for the drift compensation. Consideration of the errors obtained in the bench test conditions and those obtained in the locomotion condition (stand-to-sit, sit-to-stand, gait initiation) as recorded by the VICON system showed that the errors in trajectory reconstruction in the sagittal plane were lower than 3–4% during a 4 s of acquisition and lower than 0.9% if applied to the effective duration of the locomotory tasks, showing that the wearable device well satisfied the clinical requirements typical of the motion analysis in the evaluation of ability/disability. This study of the development of the device prototype indicates then a family of medical applications where it could have perspectives; these are those connected with the ability/ disability evaluation. As it is note, the main limitation of these clinical applications is that they are based on clinical trials with qualitative or partially quantitative observations of motor responses up to simple and well standardized motor tasks with short duration such as stand-to-sit, gait-initiation, standing a stair, sit-to-stand. Since these methods do not guarantee adequate sensitivity and reliability, quantitative measurements should be introduced in the evaluation process. Optoelectronic equipments are not adequate for their costs and encumbrance. Wearable sensors with the performances of the here described one could introduce the adequate and necessary quantitative measure to complete the analysis and evaluation. This quantitative measure could be for example also useful to create medical knowledge about the test itself and to develop an automatic clinical decision system based for example on the Neural Networks. Another advantage of this instrument rises if we consider that in the clinical validation no *a priori* knowledge specific to the locomotion task was used for trajectory reconstruction. It is well known that in model-dependent motion analysis methods physiological variability affects model accuracy, which entails the need for frequent customized calibration procedures and, at times, additional sensors [9], [10]. Not being model-dependent, our device is very practical for time-limited clinical applications with a large variety of patients. The possibility of full application in clinics of the movement analysis of portable instrumentation based on a sensor assembly like the simulated one could have relevant implications expecially if we consider that a kinematic sensor assembly like the one described in this paper can be up to twenty times less expensive than optoelectronic equipments such as Vicon [21] or Elite [26] and if we consider the continuous tails to the health care budgets. The continuous decreasing of costs in the development of technologies conjugated to the potentiality of the miniaturization technologies which permits year by year performances never thinkable before renders the kinematic sensor technology very interesting; we are already assisting for example to the realization sensor assemblies in a single chip for measuring angular velocity and acceleration such as for example the one realized by Veltink in [11] inspired to the vestibular system. We are also assisting to the integration of the inertial miniaturized sensors to ultrasound technology for the realization of systems for the P&O reconstrucion for application of Virtual Reality such as the IS900 system (Intersense, Atlanta, GA) with a cost lower than a full optoelectronic/ultrasound based system. The showed interest in the developing of integrated kinematic sensors based on different assemblies rised the necessity of a modular software for simulating the performances and furnishing indications for designers. Encouraged from all of these aspects we are now planning the development of a miniaturised device with the performances of the actual device combining solutions based on hybrid circuits realization and all on chip solution such as those premised now by the Very Large Scale Integration or Ultra Large Scale Integration technology.

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