A BSN based service for post-surgical knee rehabilitation at home

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ABSTRACT

The paper illustrates a prototype of an end-to-end service to support unassisted rehabilitation of motor functions. The core of the developed solution is a system able to analyze the patient movements during the execution of the prescribed exercises. The motion analysis relies on real-time evaluations of biomechanical parameters, derived from inertial and magnetic data provided by a wireless body sensor network, worn by the patient while he is doing his exercises. The method uses an approach based on complementary filters and addresses a number of challenges, such as compensating the incorrect positioning of motes and managing perturbations of the Earth magnetic field. Besides, the solution provides the patient with a number of "coaching functions", aimed at helping him in getting the best from his training at home, and with a videoconferencing tool to be used whenever the direct evaluation of the therapist is needed. Although this system can have a wider application field, in this work the focus is on the knee rehabilitation after the anterior cruciate ligament (ACL) reconstruction, in order to demonstrate the suitability of this solution to address specific clinical requirements. Preliminary results on this case study are provided.

Categories and Subject Descriptors

J.3 [**Computer Applications**]: Life and Medical Sciences–*Health* (http://www.acm.org/about/class/ccs98-html)

General Terms

Algorithms, Management, Measurement, Performance, Design, Human Factors, Verification.

Keywords

Knee Rehabilitation, Inertial Measurement Unit, Functional Calibration, Orientation Estimation, Complementary Filters, Tele-rehabilitation, Joint Coordinate System, Joint Angle, Anterior Cruciate Ligament

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1. INTRODUCTION

Knee injuries are one of the most frequent problems of the musculoskeletal system reported in primary care. The prevalence is 48 per 1000 patients per year [1]. In 9% of these cases, there is a damage to one or more ligaments, of which the anterior cruciate ligament (ACL) is the most commonly injured. The ACL is a primary stabilizer of the knee, thus a rupture can lead to functional instability (i.e. giving-way episodes). Conservative or surgical treatment of this instability is indicated for regaining pre-injury level of function [1] [2]

After ACL reconstruction, the speed and safety with which an athlete returns to sports depends largely on the rehabilitation protocol [4]. Nevertheless, considering the large differences in clinical and outpatient protocols, there is no consensus regarding the content of such a rehabilitation program and then differences between rehabilitation departments can be observed.

In any case, the common practice is to perform the postoperative rehabilitation in the hospital gym, under the supervision of the physiotherapist, which controls the patients for the correct execution of movements and adapts the exercises to the regained level of function. Patients are also supposed to perform a number of specific exercises at home, but usually compliance cannot be controlled, because no system for the remote assessment of patient movement is available. A reliable solution for the automatic evaluation of movement parameters not only would enable to control the compliance and to evaluate the improvements over time, but it would help to motivate the patient, it would reduce the time and costs associated to the transportation to the gym, and last but not least it would lead to reduce the number of sessions in hospital, with a consequent improvement of the quality of life of patients and a cost saving for the healthcare organization, while keeping the current quality level of care or even improving.

Because of technological advances in microelectronics, sensing, communication and data analysis, wearable sensor networks are now much less invasive, more reliable and affordable and then suitable for developing tele-rehabilitation applications, customized to specific physiotherapy protocols [5][6].

In the case of ACL, and more in general of knee rehabilitation, exercises consist of sequences of lower limb movements along prescribed spatial patterns, where, to assure the correctness and the effectiveness of the exercise, the knee and ankle joint angles which can be reached during motion are constrained within minimum-maximum allowable ranges [3][4]. Hence kinematic analysis of lower limbs, in particular the measurement of knee and ankle 3D angles has become an important area of research in biomechanics [14][17]. Many approaches have been proposed to analyze body joint angles [8], among the others: optical, magnetic and ultrasound motion tracking systems[7], video image processing[9], electrogoniometers[10][9], textile fabrics with integrated sensing devices (e.g. conductive fibers)[11][10], **I**nertial **M**easurement **U**nits containing accelerometers and gyroscopes (IMU), eventually with **M**agnetometers (MIMU), supporting wireless transmission of their sensor data [12][16][18].

Standard optoelectronic and magnetic motion capture systems provide accurate information on body joint angles, but are expensive lab equipment, which must be used by highly trained staff. Electrogoniometers are more invasive than the above mentioned approaches, while the reliability and accuracy in angle measurement obtained by fabrics with integrated sensing devices are still under assessment.

IMUs and MIMUs are cheap, small and can be easily worn by users in ambulatory or home environments. They provide continuous monitoring of the user kinematic over long period of time, and in principle, in an unlimited measurement volume[19][20]. On the other hand, MIMUs provides their orientation respect to an Earth-fixed reference frame, defined by gravity and local magnetic field, but are sensitive to magnetic field distortions easily found in home environment. IMU are unaffected, but they are unable to sense heading and their orientation is provided respect to a their specific initial orientation frame, therefore an incorrect wearing of IMU can strongly impact the results.

In this paper we describe an end-to-end solution for unsupervised knee rehabilitation. The case of study is based on a protocol, described in *Section 2*, developed at Centro Traumatologico e Ortopedico (C.T.O.) of Turin and currently in use. *Section 3* illustrates an adaptive algorithm to process MIMU's data to derive reliable joint angle measurements, also in presence of magnetic field distortions and wrong mote positioning. Then, *Section 4* provides an overview on the architecture of the overall system and some insights on the patient's application, while preliminary results are discussed in *Section 5*. Finally, conclusions and future works are outlined in *Section 6.*

2. KNEE REHABILITATION PROTOCOL

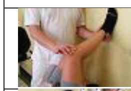
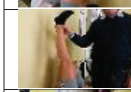
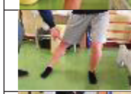
In the physiatrist department of the C.T.O. hospital in Turin, the experimental data were combined with information from background literature to develop an optimal evidence-based rehabilitation protocol to be applied *in hospital* after ACL reconstruction. This post-operative protocol is articulated in a number of phases, starting since the day after the surgery until the complete recovery of knee functionality. At the end of each phase the progresses are evaluated by the physiatrist, around 7^{th} , 15^{th} , 45^{th}, 90^{th} and 150^{th} day after surgery. The work described in this paper focus on the first three phases, corresponding to the first 45 days of rehabilitation.

Since the first day after the operation, the patient should start to perform vascular exercises and isometric co-contraction, to prevent circulation problems and loose of strength. Starting from the middle of the first week, the patient can start exercises, with a limited range of knee flexion and extension, typically between 0° and 90°, done while sitting or lying on a bed. Only after a couple of weeks the patient can start exercises which must be performed while standing and loading the operated limb.

Among the exercises commonly performed at the C.T.O gym for this protocol, a team of therapists have selected those that are the most appropriate for a non-assisted performance and for each of them they have produced video demonstrations and a set of instructions for the patients. In addition a set of parameters to be measured, associated to corresponding reference ranges, were defined in order to automatically check the correct execution of each exercise and derive an objective scoring for each performance.

Table 1 illustrates a few of these exercises, parameters and acceptable value ranges, selected for the tele-rehabilitation.

Table 1. Examples of adopted exercises

	Exercise	Parameters	Range of values [1]
	**Wall Sliding**	Knee angle Duration	0° - 90° > 5'
	**Hip Flexion**	Knee angle Ankle angle	0° - 20° 50° - 90°
	**Stretching**	Knee angle Ankle angle	0° - 20° 50° - 90°
	**Semi-Squat**	Knee angle # Repetitions	0° - 90° 5 - 10
	**Lateral Lunge**	Knee angle # Repetitions	0° - 60° 5-10
	**Front Lunge**	Knee angle # Repetitions	0° - 60° 5-10

At the end of each exercise the patients are asked to evaluate the level of pain, perceived during the execution of the exercise.

3. JOINT ANGLE ESTIMATION

In the design of the joint angle measurement system we adopted a MIMU based body sensor network approach, for the reasons discussed in the Introduction. MIMUs provides their orientation respect to an Earth-fixed reference frame defined by gravity and local magnetic field, which is an advantage respect to IMUs, which provides their orientation respect to their specific initial orientation frame. A drawback of MIMUs is their sensitivity to magnetic field distortions, which can be present in home environment.

An important aspect to be considered in joint angle measurements is that to be useful, measures obtained by different measurement systems at different times should be comparable on a common basis.

The International Society of Biomechanics (ISB) deals with this requirement by defining standard procedures for joint angular estimation based on relative orientations between Cartesian **B**one-embedded **A**natomical **F**rames (BAFs)[21].

In principle, the relative orientation of two body segments linked by a spherical joint can be evaluated from the relative orientation

[1] Range of values must be defined case by case. The table includes the most commonly used values, applicable at least two weeks after the operation

of the two MIMUs fixed at each segment, but the MIMUs reference frame (MT, MS, MF, Fig.1) generally does not coincide with the BAF frame (O_T, O_S, O_F, Fig.1), and further, their relative orientation changes at every repositioning throughout the sessions. This problem, if not compensated for, precludes the accurate measurement of joint angles [22][15]. Using the ISB reference system for measurement comparison is then necessary to estimate the relative orientation between the IMU/MIMUs and BAFs frames (Fig.1). To this purpose, functional calibration procedures have been proposed [22][15][14].

These approaches allow joint angle estimation in anatomical coordinates both when magnetic measurements are not available (IMU) or unreliable (MIMU).

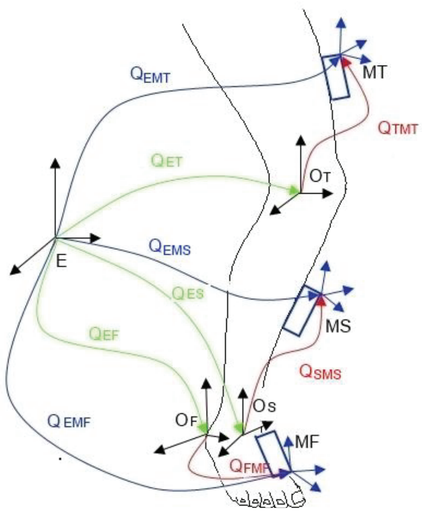


Figure 1. MIMUs, BAFs and Earth Reference Frames

3.1 ISB Reference Model

The knee and ankle joint angles relevant in the rehabilitation exercises are defined according to ISB standards and are based on relative orientations between femur, tibial, and foot BAFs frames[21] (See Fig.1) and their relative rotations[23]. Hence knee and ankle angle measurements in the ISB reference system requires to estimate the relative orientations between the three MIMUs fasten by elastic straps on the thigh, shank and foot and their respective BAFs frames (Fig.1). It must be pointed out that, disregarding these relative orientations (Q_{SMS},Q_{TMT},Q_{FMF}) can lead to large errors in angle estimation, due to potential errors in positioning of sensors from session to session.

3.2 The orientation algorithm

In this application the wearable system is based on three MIMUs (see. Sect. 4. for HW/SW details) worn on the thigh, the shank and the foot (Fig.1). Many orientation estimation algorithms have been presented in the last decade for real-time body tracking and joint angle measurements [12][14][15][17][18], (see [13][16] for a review). Here we adopt an approach based on complementary filter, similar to [17][18], chosen for its simplicity and performance, and comparable in accuracy to Kalman approaches[17]. The algorithm estimates the quaternions Q_{EMT},Q_{EMS},Q_{EMF} of the orientations of the IMU/MIMU respect to the Earth frame (Fig.1), and then by a functional calibration procedure the anatomical referenced quaternion Q_{ET},Q_{ES},Q_{EF}, from which the knee and ankle joint angles are finally obtained.

As in[18], every quaternion $\boldsymbol{q}_{est}$ is estimated by the fusion of the two separate orientation calculations, as indicated in Eq. (1):

$$\boldsymbol{q}_{est,t} = \gamma_t\, \boldsymbol{q}_{\nabla,t} + (1-\gamma_t)\boldsymbol{q}_{\omega,t}, \quad 0 \le \gamma_t \le 1 \tag{1}$$

where $\boldsymbol{q}_{\nabla}$ is a low frequency estimate based on magnetic and gravitational fields, and $\boldsymbol{q}_{\omega}$ a high frequency estimate based on angular rate. The subscripts t indicate the value calculated at time t.

The quaternion $\boldsymbol{q}_{est}$ can be computed by numerically integrating the (discrete-time) quaternion derivative, according to Eq. (2):

$$\boldsymbol{q}_{est,t} = \boldsymbol{q}_{est,t-1} + \dot{\boldsymbol{q}}_{est,t}\,\Delta t \tag{2}$$

where Δt is the sampling period and $\dot{\boldsymbol{q}}_{est}$ is given by Eq. (3)

$$\dot{\boldsymbol{q}}_{est,t} = \dot{\boldsymbol{q}}_{\omega,t} - \beta(t)\, \dot{\boldsymbol{q}}_{\nabla,t} \tag{3}$$

The result in (2) is then normalized to be a unit quaternion of rotation [18]. $\dot{\boldsymbol{q}}_{est,t}$ is the quaternion representing the rate of change of the orientation, obtained by fusing the angular rate $\dot{\boldsymbol{q}}_{\omega,t}$ and the magnetic and gravity info contributions $\dot{\boldsymbol{q}}_{\nabla,t}$ by means of the parameter $\beta(t)$. In [18] the parameter β is constant, while we define it dynamically. As further explained below, in quasi-static conditions the optimal $\beta(t)$ should compensate the gyro drift and accumulated orientation errors by using the $\dot{\boldsymbol{q}}_{\nabla,t}$ contribution, while in dynamic condition only a small contribution from $\dot{\boldsymbol{q}}_{\nabla,t}$ is given because the gravity reference in the acceleration measures is corrupted by unknown component due to motion.

The quaternion derivative $\dot{\boldsymbol{q}}_{\omega,t}$ in (3) can be expressed by Eq. (4):

$$\dot{\boldsymbol{q}}_{\omega,t} = \tfrac{1}{2}\, \boldsymbol{q}_{est,t-1} \otimes \boldsymbol{\omega}_t \tag{4}$$

where $\boldsymbol{\omega}_t$ is the angular rate, measured by the gyroscope, $\boldsymbol{q}_{est,t-1}$ is the previous estimate of orientation, and the symbol $\otimes$ represents the quaternion product.

In Eq. (3) $\dot{\boldsymbol{q}}_{\nabla,t}$ is a normalized direction in quaternion space, which starting from $\boldsymbol{q}_{est,t-1}$ reaches $\boldsymbol{q}_{opt}$, the optimal $\boldsymbol{q}$ which minimizes the difference between **m**S, the measured field (magnetic or gravity) in the sensor frame, and **m**E, the measured field in the Earth frame, mapped by q in the sensor frame, as indicated in Eq. (5):

$$\boldsymbol{q}_{opt} = \min_{\boldsymbol{q}} \left\| \boldsymbol{q}^{*}_{est,t-1} \otimes \boldsymbol{m}\mathrm{E} \otimes \boldsymbol{q}_{est,t-1} - \boldsymbol{m}\mathrm{S} \right\| \tag{5}$$

where $\boldsymbol{q}^{*}$ is the conjugate of the quaternion $\boldsymbol{q}$, while $\boldsymbol{m}$S and $\boldsymbol{m}$E are the three components measurement vectors represented as quaternions by the insertion of a 0 as first component.

We have introduced two improvements to the algorithm illustrated in [18]: a dynamic behavior for the parameter β and a weight on m**S**, to cope with situations when magnetic field is distorted.

A gating process on acceleration $\boldsymbol{a}$ and angular rate $\boldsymbol{\omega}$ signals defines dynamically $\beta(t)$. Under static conditions, that is when the measured acceleration is close to gravity acceleration **g** and the measured angular rate $\boldsymbol{\omega}$ is low, $\beta(t) = \beta max$, so that more importance is put on $\dot{\boldsymbol{q}}_{\nabla,t}$ acceleration and magnetic field, while when movements are detected, $\beta(t) = \beta min$ limits the contribution of $\dot{\boldsymbol{q}}_{\nabla,t}$ in favor of the angular rate info $\dot{\boldsymbol{q}}_{\omega,t}$.

if ($1-\boldsymbol{a}_{th} < |\boldsymbol{g} + \boldsymbol{a}| < 1+\boldsymbol{a}_{th}$) AND ($|\boldsymbol{\omega}| < \boldsymbol{\omega}_{th}$)
 $\beta(t) = \beta max$
else
 $\beta(t) = \beta min$
end

with $\boldsymbol{g}$ gravity acceleration and $\boldsymbol{a}_{th}$ and $\boldsymbol{\omega}_{th}$ the gating thresholds (e.g. $\boldsymbol{a}_{th} = 0.1$g, $\boldsymbol{\omega}_{th} = 4°$/sec).

The second improvement introduces a weight w_m in the calculation of $\boldsymbol{q}_{opt}$ which depends on the reliability of the magnetic measurements $\boldsymbol{m}$Em, $\boldsymbol{m}$Sm. In the sensor calibration phase, made *in-lab* with a non-distorted magnetic field (Earth field), a reference value $m\mathrm{Sm}_{REF} = |\boldsymbol{m}\mathrm{Sm}_{CAL}|$ for the magnetic field modulus is evaluated. In operative conditions at home the actual magnetic field modulus | $\boldsymbol{m}$Sm | is then compared to $m\mathrm{Sm}_{REF}$ to detect magnetic field distortions and thus to weight accordingly less the magnetic data.

In working conditions, a large percentage of unreliable magnetic samples indicate that the MIMU behaves similar to an IMU; this trigs a warning to the user to perform a functional calibration sequence to recover the lacking heading information of the MIMU, as described in Sec. 3.3.

3.3 The Functional Calibration Procedure

An at-home rehabilitation application cannot rely on skilled users to be effective. This is important especially for the set-up calibration phase, which is necessary for correct angle measurement and must not require a skilled user intervention. Solutions based on MIMUs and mechanical devices have been proposed [24]; they estimate anatomical axis, but require skilled staff to be employed and are not practical for in-field applications. At-home user setups cannot guarantee a perfect positioning of IMU/MIMUs respect to any body fixed landmark point. We can only assume the relative orientations between IMU/MIMUs and BAFs remain constant during the execution of the exercise (i.e. quaternions Q_{SMS}, Q_{TMT} and Q_{FMF} in Fig.1 are constants), an assumption generally verified in our context. The need for a practical calibration procedure, which can be easily performed by the patient at home, can be satisfied by a combination of the reference and the functional MIMU/IMUs calibration methods [22].

To define a suitable and practical calibration procedure for our context, we extend these original approaches by including different type of movements, info acquired from MIMUs and calibration procedures.

In the proposed approach the patient first wears the thigh, shank and foot MIMUs, and assumes a standard anatomical position, standing up, well balanced, and with foot parallel to the sagittal plane (Fig.2a). Then he starts to perform two sets of movement sequences: one consists of hip abduction/adduction (movement in the coronal plane, see Fig.2b); the other one is a hip extension/flexion (movement in the sagittal plane, see Fig.2c). In either these cases, the patient can lean on a chair or a table in order to help the balance, he may not do any movement of either the knee or the ankle joint, as illustrated in Fig.2b and Fig.2c. For small range rotations these sequences of movements can be assumed around the anatomical axes.

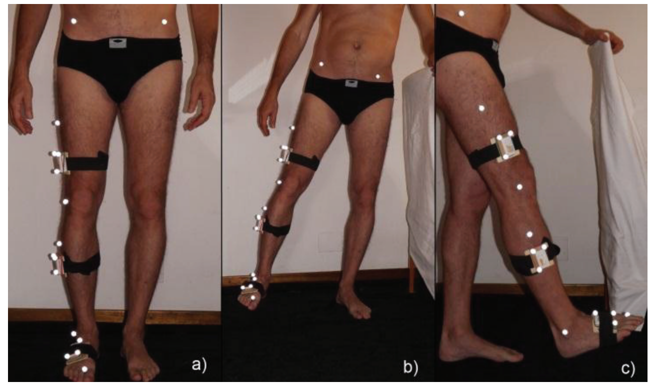


Figure 2. A subject equipped with MIMUs and the reflective marker set of the reference system in standing posture (a), hip abduction/adduction movements (b), and hip extension/flexion (c)

The recordings of the three MIMUs signals acquired for a few seconds in the static and dynamic phases are then processed to estimate the quaternions of the relative orientations between the three couple MIMU-BAFs as in the following.

In the static phase (stand-up, the gravity $\mathbf{g} = |\,\mathbf{g}\,|\,\hat{\boldsymbol{g}}$ is the only detected acceleration) the longitudinal anatomical axes of the femur $\hat{\boldsymbol{y}}_{\boldsymbol{T}}$, tibial $\hat{\boldsymbol{y}}_{\boldsymbol{S}}$ and foot $\hat{\boldsymbol{y}}_{\boldsymbol{F}}$ are all aligned with $\hat{\boldsymbol{g}}$ by definition, that is $\hat{\boldsymbol{g}}_{\boldsymbol{E}}$ has the same orientation as $\hat{\boldsymbol{y}}_{\boldsymbol{T}}$ in the JCS knee frame (OT, $\hat{\boldsymbol{x}}_{\boldsymbol{T}}$, $\hat{\boldsymbol{y}}_{\boldsymbol{T}}$, $\hat{\boldsymbol{z}}_{\boldsymbol{T}}$), and the same holds for the ankle and the foot frames. We can impose this constraint to estimate the rotation matrix $\mathbf{R_{TMT}}$ associated to Q_{TMT} by solving the following Wahba problem [25]:

find the rotation matrix $\mathbf{R}$ which minimizes the loss function L, defined by Eq. (6):

$$\mathrm{L}(\mathbf{R}) = \sum_{i=1}^{n} a_i \left\| \hat{\boldsymbol{b}}_i - \boldsymbol{R}\,\hat{\boldsymbol{s}}_i \right\| \tag{6}$$

over the set of i = 1..n sample acquired, with a_i weights and $\hat{\boldsymbol{b}}_i$, $\hat{\boldsymbol{s}}_i$ unit vectors in body and sensor frames.

Then we can write (in compact notation) for the knee, ankle and foot reference frames the following set of three functions to be minimized :

$$\mathrm{L_g}(\mathbf{R}_{\mathrm{T,S,F}}) = \sum_{i=1}^{n} a_{i\,T,S,F} \left\| \hat{\boldsymbol{y}}_{i\,T,S,F} - \boldsymbol{R}_{TMT,SMS,FMF}\,\hat{\boldsymbol{g}}_{i,MT,MS,MF} \right\| \tag{7}$$

In the dynamic calibration phase the leg oscillation movements in the sagittal plane correspond to an angular rotation $\boldsymbol{\omega}(\boldsymbol{t})_{E,}$ around the (quasi stationary) hip joint. The angular velocities $\boldsymbol{\omega}(\boldsymbol{t})_{MT}$, $\boldsymbol{\omega}(\boldsymbol{t})_{MS}$, $\boldsymbol{\omega}(\boldsymbol{t})_{MF}$ as measured by gyroscopes must have the same value in the knee, ankle and foot reference frames and in the three corresponding BAFs (apart from constant rotations represented by $\mathbf{R_{TMT}}$, $\mathbf{R_{SMS}}$, $\mathbf{R_{FMF}}$), because the six frames rotate together rigidly around the hip joint. In the knee, ankle and foot BAFs their unit vector directions $\hat{\boldsymbol{\omega}}_{sagit\,T}$, $\hat{\boldsymbol{\omega}}_{sagit\,S}$, $\hat{\boldsymbol{\omega}}_{sagit\,F}$ correspond to the three anatomical axes $\hat{\boldsymbol{z}}_T$, $\hat{\boldsymbol{z}}_S$, $\hat{\boldsymbol{z}}_F$ axes.

Then we can write for the knee, ankle and foot reference frames the following new set of three functions to be minimized:

$$_{\mathrm{sagit}}(\mathbf{R}_{\mathrm{T,S,F}}) = \sum_{i=1}^{n} b_{i\,T,S,F} \left\| \hat{\boldsymbol{z}}_{i\,T,S,F} - \boldsymbol{R}_{TMT,SMS,FMF}\,\hat{\boldsymbol{\omega}}_{sagit\,T,S,F} \right\| \tag{8}$$

Analogously, when the leg performs abduction/adduction movements in the coronal plane, we can use the same arguments as before to conclude that the unit vector directions $\hat{\boldsymbol{\omega}}_{cor\,T}$, $\hat{\boldsymbol{\omega}}_{cor\,S}$, $\hat{\boldsymbol{\omega}}_{cor\,F}$ correspond to the three anatomical axes $\hat{\boldsymbol{x}}_T$, $\hat{\boldsymbol{x}}_S$, $\hat{\boldsymbol{x}}_F$ thence giving the following functions to be minimized:

$$\mathrm{L_{cor}}(\mathbf{R}_{\mathrm{T,S,F}}) = \sum_{i=1}^{n} c_{i\,T,S,F} \left\| \hat{\boldsymbol{x}}_{i\,T,S,F} - \boldsymbol{R}_{TMT,SMS,FMF}\,\hat{\boldsymbol{\omega}}_{cor\,T,S,F} \right\| \tag{9}$$

We note that the static and dynamic phases give information about three different BAFs axes, avoiding singularities in the set of nine loss functions. These are minimized by SVD [25] and solved for $\boldsymbol{R}_{TMT}$, $\boldsymbol{R}_{SMS}$, $\boldsymbol{R}_{FMF}$. The rotation matrices $\boldsymbol{R}_{EMT}$, $\boldsymbol{R}_{EMS}$, $\boldsymbol{R}_{EMF}$ associated to estimated quaternions Q_{EMT},Q_{EMS},Q_{EMF} and the constant matrices $\boldsymbol{R}_{TMT}$, $\boldsymbol{R}_{SMS}$, $\boldsymbol{R}_{FMF}$ are than combined to give the absolute orientation matrices of the BAFS, as illustrated in Eq.(10):

$$\begin{aligned} \boldsymbol{R}_{ETknee} &= \boldsymbol{R}_{EMT}\boldsymbol{R}_{TMT}^{T} \\ \boldsymbol{R}_{ESankle} &= \boldsymbol{R}_{EMS}\boldsymbol{R}_{SMS}^{T} \\ \boldsymbol{R}_{EFfoot} &= \boldsymbol{R}_{EMF}\boldsymbol{R}_{FMF}^{T} \end{aligned} \tag{10}$$

from which the JCS knee and ankle angles are easily derived by applying the ISB definitions. We note that this functional calibration approach avoid the use of magnetic measurements. This was chosen because the functional calibration allows the correct measurement of the JCS angles even when magnetic field measurements are unreliable.

4. THE REHABILITATION PLATFORM

The final goal of the work presented in this paper is to develop an end-to-end platform for tele-rehabilitation, extending the current functionalities of Nuvola IT Home Doctor, the Telecom Italia e-health platform [26].

The target service will enable the patient to keep track of his rehabilitation activities at home or outdoor and sending evaluations of them to a cloud platform, which can be accessed by physicians and therapists. These evaluations are calculated on the basis of kinematics data, collected by means of a Wireless Body Sensor Network (WBSN) and transmitted via Bluetooth to a smartphone or a tablet, where an application not only analyzes the data but also provides real-time feedbacks to the patient, to help him to improve his physical training. Then, a number of additional services will be supported, such as tutoring, video-call with therapist, management of appointments at hospital, etc.

Since this activity builds on well-established work on the cloud platform and data reporting, herewith we will focus on the most innovative parts of the end-to-end solution, which are the acquisition system and the patient application.

4.1 The acquisition system

Kinematics data are collected by means of a WBSN, made by three 9DoF Shimmer nodes[27], positioned as illustrated in Fig.2a. Each node consists of: a tri-axial accelerometer (Freescale MMA7361) with full-scale range 1.5g/6g; a tri-axial gyroscope (InvenSense 500 series) with full-scale range from -500/ +500°/sec; a tri-axial magnetometer (Honeywell HMC5843) with full-scale range from -0.7/4.5Ga; a TI MSP430 microcontroller; a Bluetooth radio (Roving Networks RN-42); a IEEE 802.15.4 radio (TI CC2420) and a 450mAh rechargeable Li-ion battery.

In our implementation we used the Bluetooth radio for the communication between the MIMUs and the gateway, an Android tablet (Samsung Galaxy Note 10.1). The MIMUs were synchronized during the setup phase and no significant drifts on timestamps were observed in the time frame of acquisitions. Data about each rehabilitation session can be sent to the cloud platform either via 3G or Wi-Fi connection.

4.2 The application

The patient's application was developed taking into account two main goals: the *"coaching" function*, aimed at supporting the patient in his rehabilitation program and the *"assessment" function*, aimed at evaluating and communicating to the therapist the patient progresses over time.

Concerning the coaching function, many different features where developed, including but not limited to: a *tutoring video library*, illustrating the best practices to avoid complications after the operation and the correct execution of the exercises; an automatic *control on the correct positioning of the MIMU*; a number of *warning on the correct execution* of the exercise, as illustrated in Fig.3; a *feedback about the trend of rehabilitation*, as illustrated in Fig.4; *personalization of training plan*; possibility of making *video-call with the therapist* to receive direct indication about the exercises; *management of the appointments* with therapist and physiatrist; etc.

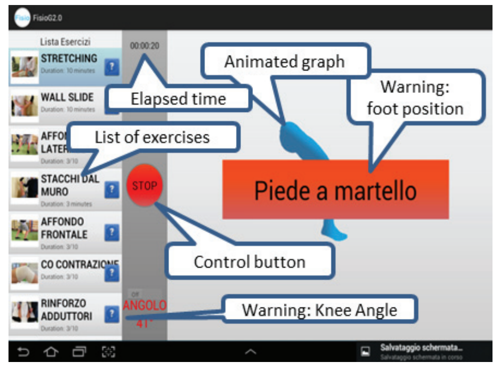


Figure 3. GUI sample

Figure 4. GUI sample

The assessment function is mainly based on the measurement of the knee and ankle angle. Other evaluation parameters are related to compliance (e.g. number of rehabilitation sessions per day, duration of the exercises) and to the perceived level of pain, that the patient must evaluate at the end of each exercise, using an 11-levels *Numerical Rating Scale* of pain [28].

An overall score for each exercise is calculated and made available to therapist through the e-health platform. For a detailed control on the execution of each exercise additional information is stored, including joint angles and videos shot by the webcam, during the exercises.

5. PRELIMINARY RESULTS

To assess the performance of the proposed measurement system a Vicon© MX13 setup with 6 MX synchronized cameras was used as reference. Four healthy subjects, three males and one female, participated to this study.

Three supports, each of them carrying a MIMU and three reflective markers, where attached by elastic straps on the thigh, the shank and the foot of the subjects (See Fig. 2). A set of reflective markers where placed on the body according to the ISB standard for lower extremity measurements [23]. The subjects performed the functional calibration procedure before each rehabilitation exercise. This sequence was repeated four times for every subject, changing the positioning of the mote and the location of the subject within the lab. To evaluate the accuracy of angle measurements, the relative rotation between the BAF estimated by the algorithm under test, provided by the quaternion $Q_{EST,}$ and the BAF calculated by the optical reference system, provided by quaternion $Q_{OPT,}$ were calculated. The quaternion representation of BAFs was used and the transformation $Q_{REL} = Q_{EST} \otimes Q_{OPT}^{*}$ was expressed compactly in angle-axis notation [7]. Mean, standard deviation and absolute maximum of the time samples of the total angle differences $\Delta\theta_{knee}$ and $\Delta\theta_{ankle}$ for the knee and ankle JCS angles during the execution of the rehabilitation exercises are shown in Table 2. Some values are large but below the conservative threshold values established by medical staff.

Table 2. Angular errors of the measurement system

	Mean	Standard Deviation	Abs. maximum
Knee	**2.4**	**1.8**	**6.2**
Ankle	**3.1**	**2.4**	**8.1**

6. SUMMARY AND FUTURE WORKS

In this paper an optimized system for joint angle evaluation and its application to the case of unassisted rehabilitation after Anterior Cruciate Ligament reconstruction has been presented. It relies on the analysis of magnetic and inertial data, based on the complementary filters approach. In order to cope with typical problems, such as perturbation of the Earth magnetic field and incorrect mounting, a suitable strategy has been proposed.

The results show that the system is suitable to evaluate the patient compliance, at least with the adopted exercises, which are representative for knee rehabilitation protocols applicable after ACL reconstruction. In addition the graphical interface of the application resulted intuitive and easy to use to the subjects. However the test was performed with a very limited number of healthy subjects in a lab, therefore the validation of the end-to-end service requires a more extended evaluation. A suitable number of operated patients, using the system at home, should be engaged in order to thoroughly evaluate the effectiveness of the proposed solution.

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8. REFERENCES

[1] Plas C., Dingjan R., Hamel A. et al. (1998) Nederlands *Huisartsen Genootschap standaard traumatische knieproblemen* M66. Available via DIALOG. http://nhg.artsennet.nl/upload/104/standaarden/M66/std.htm. Accessed June 1998

[2] Shelbourne, K.D., Nitz, P. 1990. *Accelerated rehabilitation after anterior cruciate ligament reconstruction,* Am J Sports Med June 1990 vol. 18 no. 3 292-299

[3] Trees A., Howe T., Dixon J., White L. (2005) *Exercise for treating isolated anterior cruciate ligament injuries in adults* (review). Cochrane Database Syst Rev 4:1–41

[4] Cascio B., Culp L., Cosgarea A. (2004) *Return to play after anterior cruciate ligament reconstruction.* Clin Sports Med 23:395–408

[5] Patel et al. *A review of wearable sensors and systems with application in rehabilitation*. Journal of NeuroEngineering and Rehabilitation 2012 9:21

[6] Giggins O., Kelly D. and Caulfield B. , *Evaluating Rehabilitation Exercise Performance Using a Single Inertial Measurement Unit ,* Proceedings PervasiveHealth 2013, pp. 49-56, Venice, Italy, May 5-8 2013

[7] Kuipers J.B., *Quaternion&Rotation Sequences: A Primer with Applications to Orbits, Aerospace and Virtual Reality*, Princeton Univerity Press, Princeton, New Jersey, 1999

[8] Welch G., Foxlin E., *Motion Tracking: No Silver Bullet, but a Respectable Arsenal*, Computer Graphics and Applications, IEEE vol.22 , n. 6 , Nov.-Dec. 2002 , 24 - 38

[9] Huiyu Zhoua, Huosheng Hub, *Human motion tracking for rehabilitation—A survey,* Biomedical Signal Processing and Control, vol 3, no. 1, 2008, Pages 1–18

[10] Wang P. T., King C. E., Do A. H., Nenadic Z., *A durable, low-cost electrogoniometer for dynamic measurement of joint trajectories*, Medical Engineering & Physics 33 (2011) 546–552

[11] Gibbs P. T. and Asada H. H., *Wearable Conductive Fiber Sensors for Multi-Axis Human Joint Angle Measurements* , Journal of NeuroEngineering and Rehabilitation. 2005 Mar 02;2(1):7

[12] Kun L., Inoue Y., Shibata K., and Enguo C., *Ambulatory Estimation of Knee-Joint Kinematics in Anatomical Coordinate System Using Accelerometers and Magnetometers*, IEEE Trans. Biom. Eng., vol. 58, no. 2, Febr. 2011 , 435-442

[13] Peng Cheng, and Bengt Oelmann, *Joint-Angle Measurement Using Accelerometers and Gyroscopes—A Survey*, IEEE Trans.on Instr. Meas., VOL. 59, NO. 2, Feb. 2010, 404-414

[14] Favre J., Jolles B.M., Aissaoui R., Aminian K., *Ambulatory measurement of 3D knee joint angle*, Journal of Biomechanics 41 (2008) pp.1029–1035

[15] Favre J., Aissaoui R., Jolles B.M., deGuise J.A., Aminian K., *Functional calibration procedure for 3D knee joint angle description using inertial sensors* , Journal of Biomechanics 42 (2009) 2330–2335

[16] Young A. D., *Comparison of Orientation Filter Algorithms for Realtime Wireless Inertial Posture Tracking*, 2009 Wearable and Implantable Body Sensor Networks, 2009. BSN 2009. Sixth International Workshop on, Berkeley, CA, 3-5 June 2009, pp. 59-64

[17] Young A. D., 2009. *Wireless Realtime Motion Tracking System using Localised Orientation Estimation*. Doctoral Thesis,University of Edinburgh, http://www.era.lib.ed.ac.uk/handle/1842/4637

[18] Madgwick S. O. H., Harrison A. J.L., Vaidyanathan R., *Estimation of IMU and MARG orientation using a gradient descent algorithm*, 2011 IEEE International Conference on

Rehabilitation Robotics Zurich, Switzerland, June 29 - July 1, 2011

[19] Fergus P., Kafiyat K., Merabti M., Taleb-bendiab A., El Rhalibi A., *Remote Physiotherapy Treatments using Wireless Body Sensor Networks ,* Proceeding IWCMC '09, pp.1191-1197, Leipzig, Germany, June 21-24, 2009.

[20] Hadjidj A. et al. *Wireless sensor networks for rehabilitation applications: Challenges and opportunities*, Journal of Network and Computer Applications, 36 (2013) 1-15

[21] Wu G. et al., *ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion – partI: ankle, hip and spine*, Journal of Biomechanics, Vol. 35, Issue 4, April 2002, p.543-548

[22] deVries W.H.K.,Veeger., H.E.J., Cutti A.G., Baten C., vanderHelm F.C.T., *Functionally interpretable local coordinate systems for the upper extremity using inertial & magnetic measurement systems*, Journal of Biomechanics 43 (2010) 1983–1988

[23] http://www.vicon.com/ Plug-in Gait Manual, accessed May 2013

[24] Picerno P. et al., *Joint kinematics estimate using wearable inertial and magnetic sensing modules*, Gait Posture, 2008 Nov., 28 (4), 588-95

[25] Markley F.L. and Mortari D., *How to estimate attitude from vector observations*, Astrodynamics 1999, vol. 103, 1999, pp. 1979-1996, AAS 99-427

[26] Nuvola IT Home Doctor, http://www.nuvolaitaliana.it/wp-content/uploads/2012/04/Sito-NI-_-format-scheda-approfondimento-Nuvola-It-Home-Doctor1.pdf, Accessed June 2013

[27] http://www.shimmer-research.com/p/products/sensor-units-and-modules/wireless-9dof-kinematic-sensor, Accessed June 2013

[28] Tasmuth T, Blomqvist C, Kalso F: *Chronic post-treatment symptoms in patient operated in different surgical units*. Eur J Surg, 1999; 25:38-43