An improved solution for knee rehabilitation at home

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ABSTRACT
The paper illustrates a prototype of an end-to-end solution to support knee rehabilitation at home. The solution has been designed by a multidisciplinary team, including engineers, physical therapists, physicians and ergonomists. The correct execution of the rehabilitation exercises is evaluated in real time by monitoring the range of the knee joint angles by means of Inertial Measurement Units (IMUs) worn by the patient on the shank and thigh. The parameters of the kinematic model of the lower limb, necessary for knee angle estimation, are estimated by an innovative functional calibration algorithm, which is detailed in the paper. The main advantages of this algorithm is its robustness to incorrect IMU placement and to the specific anatomical parameter of the users. Details of the solution architecture and its sub-systems are presented along with preliminary results of the comparison with a reference optoelectronic motion capture system.

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, Tele-rehabilitation, Joint Coordinate System, Joint Angle

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1. INTRODUCTION
Knee rehabilitation therapies are usually deployed in hospital and ambulatories under the supervision of the physiotherapist, which controls the correct execution of movements and adapts the exercises to the regained level of function [1][2]. The request for this kind of services is rising, mainly due to aging population, while national healthcare systems are not extending their budgets, therefore the patients have often to wait weeks or months before to start a rehabilitation program. To address this problem, ICT technologies could be applied to monitor the correct execution of rehabilitation exercises at home. The proposed solution should be reliable, flexible, scalable, easy-to-use and cheap. In the case of knee rehabilitation, reliability is strictly related to joint angle evaluation during the rehabilitation exercises. In this field, among the several techniques available (see [4] for a review), wearable sensor networks based on Inertial Measurement Units containing only accelerometers and gyroscopes (IMU), or also Magnetometers (MIMU) respond to most of the above mentioned requirements. IMU/MIMUs are suited to deploy tele-rehabilitation service [3], and several examples of their applications in knee-joint angle monitoring have been described [5][6][7]

However some drawbacks limit IMU and MIMU measurement reliability: MIMU are sensitive to magnetic field distortions easily found in home environment; low frequency noise of gyros impacts on long term estimation of IMU/MIMU orientation when algorithms based on gyro signal integration are applied [14]; joint angle estimation based on IMUs relative orientations is biased by incorrect wearing of the IMUs.

In this paper we extend and improve the work presented in [13] on the development of a post-surgical knee tele-rehabilitation service. First, we propose a new joint angle estimation algorithm that overcame some of the aforementioned problems. The algorithm is intended for IMUs, thus avoiding MIMUs magnetic field disturbances. Second, the algorithm uses a derivative approach to knee joint angle estimation, based on a kinematic model of the shank-knee-thigh [5][6][12]. This approach overcomes the problems of drifts of the algorithms based on gyro signal integration. Third, the algorithm relays on a simple functional calibration procedure to be performed initially by the patient. The procedure allows the estimation of both the incorrect positioning of the IMUs as in [13], and the estimation of model parameters required by the algorithm. This last estimations represent an important improvement respect to [5] for an unassisted use of the application, because it does not require skilled users for IMUs placement nor an a priori knowledge of model parameter values related to the specific patient anatomy. Finally, we have improved the tele-rehabilitation solution described in [13] with the integration of a cloud platform to collect data about rehabilitation programs and results, accessible by the physical therapist via a web application.

2. KNEE JOINT ANGLE ESTIMATION

The case of study is based on a knee rehabilitation protocol, including a number of exercises for the post-surgical rehabilitation for Anterior Cruciate Ligament reconstruction, described in [13]. The knee joint angles which can be reached during rehabilitation exercises have to be measured and checked, and they must stay within minimum-maximum allowable ranges to avoid dangerous effects [1][2]. Joint angle measurements obtained at different times can be compared if they are referred to the same body fixed reference frames [8], called Cartesian Bone-embedded Anatomical Frames or BAFs [16].

We assume a simplified model for the shank-knee-thigh limbs made by two segments (femur and tibia) with the associated femur and tibial BAFs (K and A frames in Fig. 2) linked by a spherical joint in Ko, origin of the K frame. In this case the knee joint angle estimation problem is equivalent to estimates the rotation matrix $\mathbf{R}_k$ (and its three Euler angles) which brings the K frame onto the A frame translated in Ko .

At home, rehabilitation implies that the hardware setup has to be managed by unskilled users, which cannot guarantee a perfect positioning of the two IMUs S1 and S2 with respect to any body fixed landmark point. Referring to Fig. 2 we can assume that their positions $\mathbf{r}_1$, $\mathbf{r}_2$ and their relative orientations $\mathbf{R}_1$, $\mathbf{R}_2$ with respect to K and A BAFs remain constant during the execution of the exercises. This is a reasonable hypothesis if the IMUs are tightly fastened to the limb. Incorrect IMU placements make the $\mathbf{r}_1$, $\mathbf{r}_2$, $\mathbf{R}_1$, $\mathbf{R}_2$ model parameters values approximately known and this produces joint angle estimation errors.

To overcome these problems we first use the functional calibration procedure presented in [13] (and briefly reviewed here) to estimate $\mathbf{R}_1$, $\mathbf{R}_2$ and to align the orientation of S1,S2 frames to the K and A BAFs. This simplifies the model in Fig. 2 to the model of Fig. 3. Second we introduce the kinematic equation of the simplified model and its constraints, then we extend the functional calibration procedure to estimate $\mathbf{r}_1$, $\mathbf{r}_2$. in the simplified model. Finally, we solve the kinematic equations where all parameters are known for the joint angle matrix ($\mathbf{R}_k$).

2.1 Functional calibration procedure

The patient wears the thigh and shank IMUs, respectively S1 and S2, equipped with three-axial accelerometers and gyros. In the first step of the calibration procedure he/she must stand up, stay well balanced on the two legs (Fig.1a) and keep this position for a few seconds. In the second phase, he/she starts to perform two sets of movement sequences: one consists of hip abduction/adduction (Fig.1b); the other one is a hip extension/flexion (Fig.1c). For small range rotations these sequences of movements can be assumed respectively around the anatomical axis x and z of the K and A BAFs.

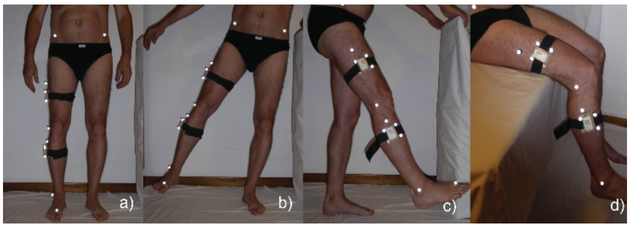

Fig. 1 A subject with IMUs and the reflective markers in standing posture (a), hip ab/adduction (b), hip flex-extension (c) and shank dandling (d)

The recordings of the two IMUs signals acquired for a few seconds in the static and dynamic phases are then processed to estimate the relative orientations between the IMUs and 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, and the longitudinal anatomical axes $\hat{\boldsymbol{y}}_{\boldsymbol{K}}$ in the knee frame K and $\hat{\boldsymbol{y}}_{\boldsymbol{A}}$ in the ankle frame A are aligned with $\hat{\boldsymbol{g}}$ by definition. In the dynamic calibration phases, the leg oscillation movements correspond to an angular rotations $\boldsymbol{\omega}(\boldsymbol{t})_{H,}$ around the (quasi stationary) hip joint. The angular velocities $\boldsymbol{\omega}(\boldsymbol{t})_{K,}$ $\boldsymbol{\omega}(\boldsymbol{t})_{A}$ must have the same value in the knee and ankle BAF reference frames, because the frames rotate together rigidly around the hip joint. Furthermore, their corresponding unit vector directions $\widehat{\boldsymbol{\omega}}_K$, $\widehat{\boldsymbol{\omega}}_A$ in the knee and ankle BAFs are aligned respectively to the anatomical axes $\hat{\boldsymbol{z}}_K$, $\hat{\boldsymbol{z}}_A$ in the case of hip ext/flexion and to the anatomical axes $\hat{\boldsymbol{x}}_K$, $\hat{\boldsymbol{x}}_A$ in the case of abd/adduction. We can use these constraints to estimate the constant rotation matrices $\mathbf{R}_1$, $\mathbf{R}_2$ representing the misalignments between S1, S2 and their respective BAFs. This is done by solving a general least square fit problem:

find the rotation matrix **R** *which minimizes the loss function L, defined by* Eq. (1):

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

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 BAF and sensor S frames respectively*.

We specialize the solution to the current case by substituting for the pair $\hat{\boldsymbol{b}}_i$, $\hat{\boldsymbol{s}}_i$ the pairs $\hat{\boldsymbol{y}}_{i\,K,A}$, $\hat{\boldsymbol{g}}_{stat\,i,K,A}$ (static case); $\hat{\boldsymbol{z}}_{i\,K,A}$, $\widehat{\boldsymbol{\omega}}_{abd-add\;K,A}$ (abd/add movements); $\hat{\boldsymbol{x}}_{i\,K,A}$, $\widehat{\boldsymbol{\omega}}_{fles-est\;K,A}$ (flex/ext movements), then solving for $\mathbf{R}_1$ (K frame) and $\mathbf{R}_2$ (A frame) (see [13]). We align the orientation of S1 and S2 to the K and A BAFs by using $\mathbf{R}_1$ and $\mathbf{R}_2$. In this way we reduce the original model (Fig.2) to the simplified one (Fig. 3).

2.2 Kinematic model of the lower limb

During hip abduction/adduction (see Fig. 1b) the segments HK and KA are aligned and S1, S2 rotate around H at the same angular velocity $\boldsymbol{\omega}_{S1} = \boldsymbol{\omega}_{S2} = \boldsymbol{\omega}_{12}$.

The measured accelerations $\boldsymbol{a}_{S1}, \boldsymbol{a}_{S2}$ in the S1, S2 frame are related to the acceleration $\boldsymbol{a}_{\boldsymbol{H}}$ in the H frame by the following kinematic equations (See [15]):

$$\boldsymbol{a}_{S1} = \boldsymbol{a}_{\boldsymbol{H}} + \dot{\boldsymbol{\omega}}_{S1} \times \hat{\boldsymbol{r}}\,|r_{S1}| + \boldsymbol{\omega}_{S1} \times (\boldsymbol{\omega}_{S1} \times \hat{\boldsymbol{r}})|r_{S1}| \quad (2)$$

$$\boldsymbol{a}_{S2} = \boldsymbol{a}_{\boldsymbol{H}} + \dot{\boldsymbol{\omega}}_{S2} \times \hat{\boldsymbol{r}}\,|r_{S2}| + \boldsymbol{\omega}_{S2} \times (\boldsymbol{\omega}_{S2} \times \hat{\boldsymbol{r}})|r_{S2}| \quad (3)$$

Where r_{S1} and r_{S2} represent respectively the distance between S_1 and H and S_2 and H, as shown in Fig. 3, while $\hat{\boldsymbol{r}}$ is the unit vector corresponding to the rotation around H.

But $\boldsymbol{\omega}_{S1} = \boldsymbol{\omega}_{S2} = \boldsymbol{\omega}_{12}$ (stiff linked segments), then:

$$\boldsymbol{a}_{S2} - \boldsymbol{a}_{S1} = [\dot{\boldsymbol{\omega}}_{12} \times \hat{\boldsymbol{r}} + \boldsymbol{\omega}_{12} \times (\boldsymbol{\omega}_{12} \times \hat{\boldsymbol{r}})](|r_{S2}| - |r_{S1}|) \quad (4)$$

with $(|r_{S2}| - |r_{S1}|) = |r_1| + |r_2| = r_{Tot}$

We can estimate r_{Tot} by solving a general least square fit problem:

find r_{Tot} *which minimizes the loss function* L, *defined by* Eq.5:

$$\mathrm{L}(r_{Tot}) = \sum_{i=1}^{n} a_i \left\| \widehat{\Delta \boldsymbol{a}}_i - \widehat{\boldsymbol{b}}_i r_{Tot} \right\| \quad (5)$$

over the set of i = 1..n *sample acquired, with* a_i *weights* and

$$\widehat{\Delta \boldsymbol{a}}_i = \boldsymbol{a}_{S2} - \boldsymbol{a}_{S1},\ \widehat{\boldsymbol{b}}_i = [\dot{\boldsymbol{\omega}}_{12i} \times \hat{\boldsymbol{r}} + \boldsymbol{\omega}_{12i} \times (\boldsymbol{\omega}_{12i} \times \hat{\boldsymbol{r}})]$$

are measured quantities sampled at time i.

We note that thank to the sensor alignment given by the estimated $\mathbf{R}_1$ and $\mathbf{R}_2$, $\hat{\boldsymbol{r}}$ is coincident with one of the sensor frame axis.

We then perform a third functional calibration movement. Sitting on a table with the shank dangling down, the user oscillates the shank only, with the knee and the thigh at rest (Fig. 1d).

For this case, the following kinematic equation provides the acceleration measured in K by a virtual sensor. The acceleration $\boldsymbol{a}_{\mathrm{K\,S2}}$ is given with respect to the S2 reference frame with origin in K as:

$$\boldsymbol{a}_{\mathrm{K\,S2}} = \boldsymbol{a}_{S2} + \dot{\boldsymbol{\omega}}_{S2} \times \hat{\boldsymbol{r}}_2 |r_2| + \boldsymbol{\omega}_{S2} \times (\boldsymbol{\omega}_{S2} \times \hat{\boldsymbol{r}}_2)|r_2| \quad (6)$$

where $\boldsymbol{a}_{S2}$ is the acceleration measured by the physical sensor on the thigh. The same acceleration $\boldsymbol{a}_{\mathrm{K\,S1}}$ measured in the S1 (steady shank) reference frame with origin in K is $\boldsymbol{g}$. Since the point K should have only one physical acceleration, $\boldsymbol{a}_{\mathrm{KS1}}$ and $\boldsymbol{a}_{\mathrm{KS2}}$ must be the same, apart from the different orientation of their frames due to shank rotation around K, which is represented by the rotation matrix $\boldsymbol{R}$:

$$\boldsymbol{a}_{\mathrm{K\,S2}} = \boldsymbol{R}\boldsymbol{a}_{\mathrm{K\,S1}}$$

$$\boldsymbol{a}_{\mathrm{K\,S2}} = \boldsymbol{R}\boldsymbol{g} = \boldsymbol{a}_{S2} + \dot{\boldsymbol{\omega}}_{S2} \times \hat{\boldsymbol{r}}_2 |r_2| + \boldsymbol{\omega}_{S2} \times (\boldsymbol{\omega}_{S2} \times \hat{\boldsymbol{r}}_2)|r_2| \quad (7)$$

By taking the squared modulus of eq. (7), (with $|\boldsymbol{R}|^2 = 1$) we get:

$$|\boldsymbol{a}_{\mathrm{K\,S2}}|^2 = |\boldsymbol{g}|^2 = (|\boldsymbol{a}_{S2}|^2 - 2\,\boldsymbol{a}_{S2}\,\boldsymbol{a}_{\omega}\,|r_2| + |\boldsymbol{a}_{\omega}|^2\,|r_2|^2) \quad (8)$$

where $\boldsymbol{a}_{\omega} = \dot{\boldsymbol{\omega}}_{S2} \times \hat{\boldsymbol{r}}_2 + \boldsymbol{\omega}_{S2} \times (\boldsymbol{\omega}_{S2} \times \hat{\boldsymbol{r}}_2)$

The eq. (8) can be solved for $|r_2|$ by quadratic least square fit or can be simplified taking into account that the last term $|\boldsymbol{a}_{\omega}|^2\,|r_2|^2$ on the right side of eq. (8) is generally smaller than the first two:

$$|\boldsymbol{a}_{\mathrm{K\,S2}}|^2 = |\boldsymbol{g}|^2 \approx (|\boldsymbol{a}_{S2}|^2 - 2\,\boldsymbol{a}_{S2}\,\boldsymbol{a}_{\omega}\,|r_2|) \quad (9)$$

Then we can estimate r_2 by solving a least square fit problem:

find r_2 *which minimizes the loss function* L, *defined by*:

$$\mathrm{L}(r_2) = \sum_{i=1}^{n} a_i\ \|c_i - d_i r_2\| \quad (10)$$

over the set of i = 1..n *sample acquired*, *with* a_i *weights* and

$$c_i = |\boldsymbol{g}|^2 - |\boldsymbol{a}_{S2\,i}|^2\,; \qquad d_i = (-2\,\boldsymbol{a}_{S2\,i}\,\boldsymbol{a}_{\omega\,i})$$

Finally we can calculate r_1 as the difference between r_{Tot} and r_2.

Now all the lower limb model parameters $\mathbf{r}_1$, $\mathbf{r}_2$, $\mathbf{R}_1$, $\mathbf{R}_2$ are known, and we can say as in [5][12] that the accelerations $\boldsymbol{a}_{\mathrm{KS1}}, \boldsymbol{a}_{\mathrm{KS2}}$ measured by two *virtual sensors* in K, as expressed in the S1 and S2 frames respectively, are related to the measured accelerations $\boldsymbol{a}_{S1}, \boldsymbol{a}_{S2}$ in S1, S2 according to the following formulas:

$$\boldsymbol{a}_{\mathrm{KS1}} = \boldsymbol{a}_{S1} + \dot{\boldsymbol{\omega}}_{S1} \times \hat{\boldsymbol{r}}_1 |r_1| + \boldsymbol{\omega}_{S1} \times (\boldsymbol{\omega}_{S1} \times \hat{\boldsymbol{r}}_1)|r_1| \quad (11)$$

$$\boldsymbol{a}_{\mathrm{KS2}} = \boldsymbol{a}_{S2} + \dot{\boldsymbol{\omega}}_{S2} \times \hat{\boldsymbol{r}}_2 |r_2| + \boldsymbol{\omega}_{S2} \times (\boldsymbol{\omega}_{S2} \times \hat{\boldsymbol{r}}_2)|r_2| \quad (12)$$

Since the point Ko should have the same physical acceleration, $\boldsymbol{a}_{\mathrm{KS1}}, \boldsymbol{a}_{\mathrm{KS2}}$ must be the same, apart from the different orientation of their frames, which is expressed by the knee joint angle rotation matrix $\mathbf{R_k}$:

$$\boldsymbol{a}_{\mathrm{KS1}} = \mathbf{R_k}\,\boldsymbol{a}_{\mathrm{KS2}} \quad (13)$$

We can then use (13) to evaluate $\mathbf{R_k}$ and the associated three Euler angles of the knee joint over the set of *i* = 1..n samples acquired. After the functional calibration procedure the values $\boldsymbol{a}_{\mathrm{KS1}}, \boldsymbol{a}_{\mathrm{KS2}}, \mathbf{R_k}$ and the angles of the knee joint can be evaluated efficiently in real-time from equations (11), (12) and (13).

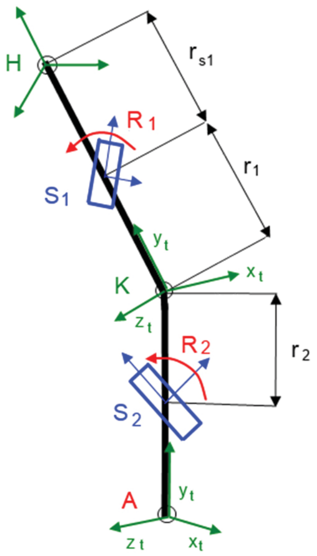

Fig. 2 Lower limb model

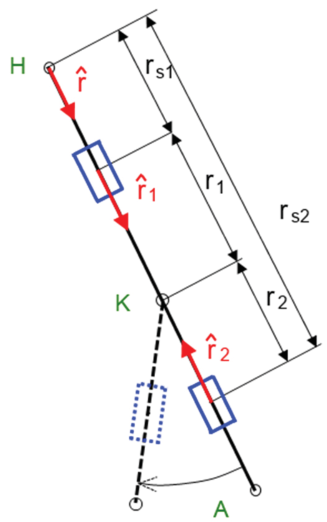

Fig. 3 Lower limb simplified model

3. THE END-TO-END SOLUTION

The calibration and estimation algorithms described before have been included in the tele-rehabilitation service Fisio@Home. The main idea behind Fisio@Home is to develop a system that is easily usable by physicians, therapists and patients. In order to achieve this target, the user interface and the functionalities were developed involving physicians and therapists of C.T.O. (Centro Traumatologico e Ortopedico) of Turin, organizing both focus groups and trial sessions with real patients. Details of the acquisition system and the mobile application used by the patients are given in [13]. Therefore the following subsections will be focused on requirements and functionalities of the web application developed for physicians and therapists and only a short reference to the patient-side subsystem will be included.

3.1 The acquisition sub-system

The aim of the patient sub-system is: (a) to assist the patients during the rehabilitation sessions with personalized exercise programs, with real-time feedback on incorrect postures and with history of assessments; (b) to assess his/her performance at the end of each exercise; (c) to send the assessments to a platform to make them available to therapists and physicians; (d) to enable communication between patients and physicians.

The assessments are mainly based on the measurement of the knee 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 [13].

The data for the real-time evaluation of the knee angle is provided by the 3-axial accelerometer and 3-axial gyroscope of two 9DoF Shimmer nodes [10], positioned on the lower limb as illustrated in Fig.1a. Data at 100 Hz are transmitted via Bluetooth to a tablet, where an Android application analyzes the data and provides real-time feedbacks to the patient, to help him/her to improve his physical training. Fig.4 shows a snapshot of the application home, which gives access to a number of tools, designed to help the patient during his/her rehabilitation program.

Fig. 4: Main functionalities of the Patient application

These tools are grouped in the following three main areas, from left to right:

Services in support of rehabilitation: they include a) the access to data related to the patient performances since the beginning of her/his rehabilitation program, b) some video tutorials about either the exercises to be performed at home or some good practices concerning the rehabilitation, c) the communication to an helpdesk for technical assistance and to a physician for health issues.

Access to the rehabilitation session: pushing the central button the IMUs automatically connect to the tablet, then the functional calibration procedure is performed and finally the application shows the list of prescribed exercises and provides a number of tools suitable to help the subject in the performance of the exercises, such as warning about incorrect executions, measure of the knee angle, counting of repetitions, etc., as shown in Fig. 5. Details about the biofeedback are provided in [13].

Fig. 5: Biofeedback of the Patient application

Messages from the physical therapist: an area on the right side gives evidence to suggestions and comments from the therapist, which can help the patient to make the rehabilitation more effective and to encourage him to comply with the prescribed physical therapy.

3.2 The clinical subsystem

The physicians and the therapists will be able to access the tele-rehabilitation platform through an easy-to-use web application that provides the following functionalities, reported in Fig.6:

User Management: this functionality allows therapists and physicians to address all the aspects related with patient registration, cancellation and personal data management (birth date, medical history, contacts).

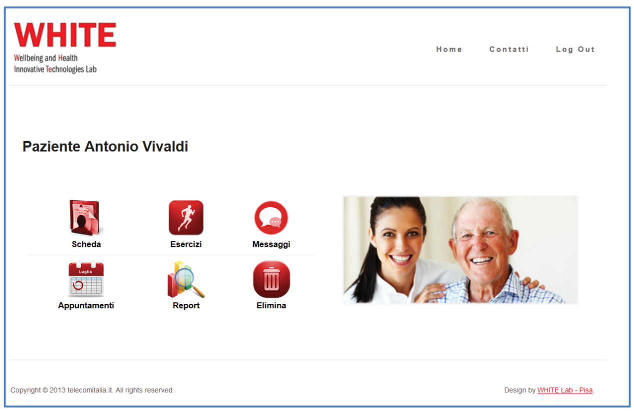

Fig. 6: Functionalities of the web application

Exercise Plan definition: with this functionality, physicians and therapists can remotely set a working plan for the patient, selecting between ten different exercises. For each of these it is possible to specify several parameters (duration and repetitions of the exercise, angle excursion in terms of minimum and maximum angles) together with a description on how to execute the working plan, as shown in Fig. 7. Each setting will be sent to the patient application on the mobile device through the Internet.

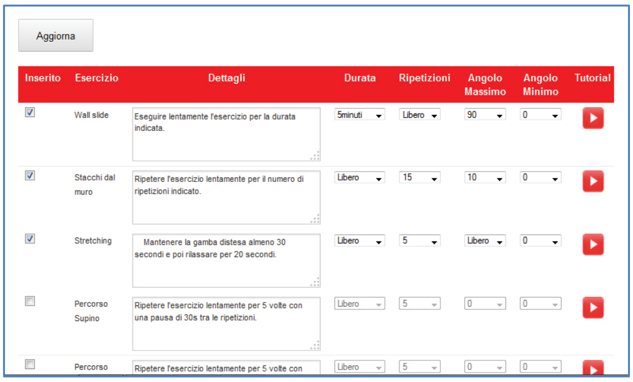

Fig. 7: Exercise plan definition

Results and Feedbacks visualization: the web app shows to the physician the results of the prescribed exercises that are executed by the patient, as illustrated in Fig. 8. For each exercise, a physician can analyze the time-based trend of the angle excursion that a patient achieved during the exercises, together with information about the related pain level. Moreover, for each exercise is showed an overall score that summarizes the quality of the execution, comparing the angular range performed during the exercise with the one prescribed by the doctor.

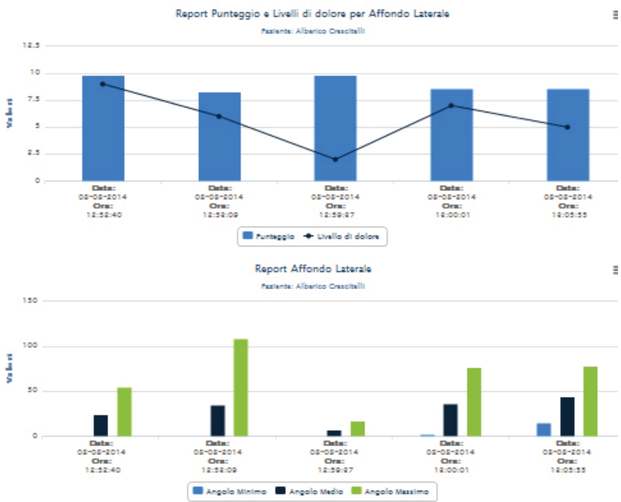

Fig. 8: Patient performance representation

Communication: the tele-rehabilitation platform allows patients and physicians to send messages each other. In this way, for example, a patient can quickly alert his physician or his therapist about unexpected difficulties in the execution of the exercises. In the same way, a therapist can send some recommendations on how to execute properly each exercise, based on the results showed in the visualization section.

3.3 Service Platform Architecture

As showed in Fig. 9, the platform that provides the web app can be structured in three fundamental layers, each of which performs a macro functional task:

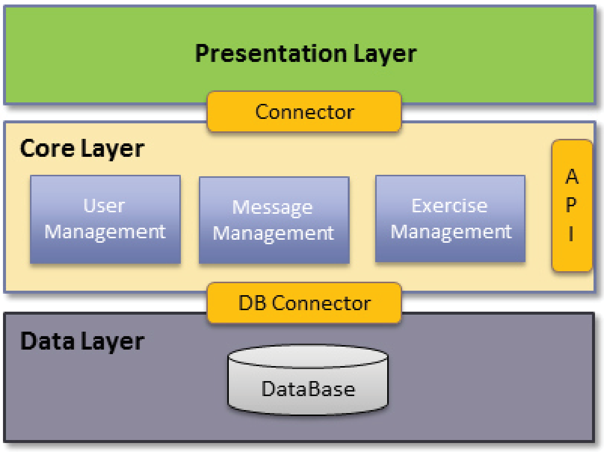

Fig. 9: F@H Service Platform Architecture

Data-Layer, for the storage of information collected by the IMU sensors and processed by the mobile device. This layer involves all the aspects related with the management of data persistence and it is focused on how to get and store information that comes from the mobile app and all the data (e.g. patient working plan and personal data, messages) provided by the doctor. At this level were preferred the choice of open source solutions for both database engine (e.g. MySQL) and data access technologies (e.g. JDBC Connector).

Core-Layer, for data processing. This layer involves the development of the business-logic core of the service, in order to process the data stored in the Data Layer. The components of this layer use an application server that provides the development and the execution functionalities of the application. The chosen technology is Apache Tomcat. Moreover, the Core-Layer exposes APIs in order to provide access to the platform by the mobile app.

The Core-Layer provides three fundamental functionalities:

- **User Management**: block that handles the functionality of adding, editing and removing users. The user information is divided in two categories: personal data and medically relevant data (e.g. anamnesis).
- **Message Management**: involves the logic related to the message exchange between user and doctor.
- **Exercise Management**, related to two basic functionalities for the doctor: setting a working plan for the user (in terms of number of exercises, execution mode and time) and data management for the results (score, angular range, feedbacks).

Presentation-Layer, for data visualization. This layer gives to the doctor the view of the information stored through the Data-Layer and processed through the Core-Layer. The front-end is a website developed using server-side Java-based technologies (e.g. Java Server Pages, JavaServer Pages Standard Tag Library) that act as Connector with the Core Layer. Communications with mobile app on tablet/smartphone are achieved using connections secured by application layer protocols (e.g. HTTPS).

4. PRELIMINARY RESULTS

To assess the performance of the proposed functional calibration algorithms an optoelectronic system (ELITE, BTS, Italy; 8 TVC 100 Hz) was used as gold standard. Three healthy subjects, two males and one female, participated to this preliminary study.

Two supports, each of them carrying an IMU and three reflective markers (used to define the 3D position/orientation of the IMUs), where attached by elastic straps on the thigh and the shank respectively (See Fig. 2). A set of reflective markers where placed on the body according to the ISB standard for lower extremity measurements [9]. The subjects performed both the static and the three dynamic functional calibration sequences of movements before each rehabilitation exercise. This sequence was repeated four times for each subject, changing the inclination of the IMUs, to simulate potential wearing errors. To evaluate the accuracy of joint angle measurements and to obtain corrected measurements, the relative rotations provided by the estimated matrices $\mathbf{R}_1$, $\mathbf{R}_2$ where used to align the sensor and the BAF frames. The $\mathbf{r}_1$, $\mathbf{r}_2$, parameter were used to estimate the knee joint angles during some representative rehabilitation exercises (See Eq. 11, 12, 13). In order to compare the angle estimated by the algorithm and the angle measured by the optoelectronic system, the quaternion representation $\mathrm{Q_{EST}}$ of the rotations $\mathbf{R_k}$ was used to compare the estimated $\mathrm{Q_{EST}}$ and the measured $\mathrm{Q^*_{OPT}}$ knee joint angles by means of the quaternion product $\otimes$ [14].

$$\mathrm{Q_{REL}} = \mathrm{Q_{EST}} \otimes \mathrm{Q^*_{OPT}} \qquad (14)$$

In eq. (14) $\mathrm{Q_{REL}}$ is the *total* angular difference, which can be expressed compactly in angle-axis notation. This means that the associated Euler difference angles are in general smaller than this upper bound. Mean, standard deviation and absolute maximum of the total angle differences $\Delta\theta_{knee}$ for the knee JCS angles during the execution of typical rehabilitation exercises are shown in Table 1. The values are large but in the acceptable range of values according to the therapists requirements. Furthermore, it should be taken into that the measurements done by therapists with goniometers are often affected by larger errors.

Table 1. Angular differences

	Mean	**Standard Deviation**	**Abs. maximum**
Knee [deg] (exercises)	**6.4**	**3.8**	**9.4**

5. SUMMARY AND FUTURE WORKS

This paper has presented a prototype service for knee tele-rehabilitation, focusing on the automatic evaluation of the range of motion and in particular on an innovative functional calibration procedures, and illustrating the tele-rehabilitation platform and the web app used by the therapist to check the performances of his/her patients, concerning the rehabilitation exercises performed at home.

The first 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. In particular the precision obtained in the evaluation of the knee range of motion seems to be suitable for the remote monitoring of knee rehabilitation programs. However, since the test was performed with a very limited number of healthy subjects in a lab, extended evaluations with patients at home are still required.

6. ACKNOWLEDGMENTS

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