Abstract—This paper presents a new tool for assessment and therapy in post-stroke upper-limb rehabilitation and a new wireless sensor technology to enhance rehabilitation robotics based on the ZigBee network of wearable Inertial Measurement Units (IMU) and Surface Electromyography (sEMG) sensor nodes. These sensor nodes will allow the measurement of kinematic and electrical muscle activity of patients in continuous therapy motion over all body segments as a Body Sensor Network (BSN). The IMU Sensor design was based on a direction-cosine-matrix DCM. The system validation was achieved with an optical motion tracking system in which cameras and IMU sensors recorded upper limb positions simultaneously during a standard gesture of reaching and grasping. The comparison between elbow flexion-extension angle in reaching and grasping movements obtained from both techniques shows equivalence. The analysis of IMU data signals for several movements demonstrates high repeatability intra and inter-subjects.

I. INTRODUCTION

A stroke is the consequence of cell death within the brain relating to either internal bleeding or a blockage in one of the two main supplying arteries. Currently, it represents a major problem in clinical medicine being a leading cause of disability in the developed world [1]. Worldwide, 15 million people suffer a stroke, of which, 5 million die and another 5 million are left permanently disabled. Strokes are uncommon in people under 40 years; when they do occur, the main risk factor is high blood pressure [2]. Strokes remain a major public health concern in the United States, with more than 795,000 cases diagnosed annually. Over 50% of subjects present some paresis [3]. Treatment and intervention is often primarily focused on lower limbs, to develop and improve walking ability, with intervention for upper limb dysfunction being secondary. However, upper limb dysfunction can be equally disabling, affecting daily functional tasks that involve reaching, grasping and manipulating objects [4].

Many stroke victims regain their abilities in daily life activities, however; they have quantitative limitations, such as reduced speed and accuracy despite neurorehabilitation. Clumsy and slow performance may be a considerable impediment to these patients, especially when fine manipulations are required. Research on the factors involved in chronic functional disability of upper limb movements and dexterity would help to enhance treatments and to elucidate the pathophysiology of post-stroke motor recovery [5].

There is great potential for the use of rehabilitation robotics to assist in motor therapy. Some of its benefits include introducing the ability to perform precise and repeatable therapeutic exercises, reduction of the physical burden on participating therapists, incorporation of interactive virtual reality systems, and the collection of quantitative data that can be used to optimize therapy sessions and assess patient outcomes [6]. In this application there is a need to obtain variables such as joint angular accelerations and angular velocities to close control loops aiming to achieve specific goals.

These sensors should have specific characteristics such as a small size, a long battery life, the ability to extract a wide range of parameters from human motion, easy adaptability to an orthotic frame, suitable bandwidth and other preferred features. These characteristics made Micro Electro-mechanical Systems (MEMS) inertial sensors attractive for the rehabilitation robotics field [7]. These sensors such as accelerometers, gyroscopes and magnetometers can combine their measures through Inertial Measurement Unit (IMU), in order to get kinematic data measures, typically acceleration, speed, position, and orientation.

In addition, sEMG (surface Electromyography), which represents activated states of muscles, has been widely used to assess progress in the process of rehabilitation and human-robot interaction. The motor recovery process can be monitored during post-stroke robot-assisted training through quantitative assessment by the sEMG parameters [8]. The additional patient control intention, through continuous myoelectric control, could provide more interaction during
the whole motion, which might be beneficial in promoting the restoration of motor functions for post-stroke patients [9].

II. BACKGROUND

Nowadays, there are a variety of IMU sensors available for wireless data transmission [10] [11], and many sensing methods have been researched in order to reduce the inaccuracies inherent in the MEMS sensor measurements [11] [12]. There are sEMG sensors implemented over several wireless protocols [13]. The aim of this work was focused on the development of a real-time IMU and sEMG sensor, small enough to be integrated with easy adaptability into clothing or an orthotic frame used by the patient. It can be used to close control loops aiming to achieve specific goals through rehabilitation robotic systems. At the same time, these sensors can be used for on-line monitoring the therapy evolution for both the medical staff and the patient, through biofeedback tools.

ZigBee technology defines the network, security and application framework for an IEEE 802.15.4-based system. These capabilities enable thousands of devices to be connected on a single wireless network. The growing expansion of ZigBee Alliance in the healthcare space has resulted in the development of the ZigBee Health Care public application profile (ZHC). It was designed for use by assistive devices operating in non-invasive health care. ZHC provides a standard for exchanging data with a wide industrial field of application between a variety of medical and non-medical devices [14].

This work aims to develop a monitoring architecture for rehabilitation therapy robotics based on the ZHC network of wearable IMU and sEMG sensor nodes. This proposed sensor is developed in a single chip mixed microcontroller ARM Cortex M4 Kinetis K53 microcontroller. It is the latest embedded processor by ARM, specifically developed to address digital signal control markets. The K50 MCU family provides designers with an Analog Measurement Engine consisting of integrated operational and transimpedance amplifiers and high-resolution ADC and DAC modules [10].

The ZHC network was based on the MC13224 Platform-in-Package incorporating a complete, low power, 2.4 GHz radio frequency transceiver, 32-bit ARM7 core based MCU, hardware acceleration for both the IEEE 802.15.4 MAC and AES (Advanced Encryption Standard) security [16].

A. IMU Sensor

The IMU Sensor design is based on the implementation of a Direction-Cosine-Matrix (DCM) for IMU application in model planes and helicopters developed in previous works [17] [18] [19] [20]. The orientation can be described by three consecutive angular rotations named Euler angles: 1) yaw \( \psi \), rotate the segment about its z axis, 2) pitch \( \theta \), rotate the segment about its y axis and 3) roll \( \phi \), rotate the segment about its x axis (Fig. 2a). To describe the motion of a human body segment, it is necessary to define two coordinate systems, the Global System Coordinate (GSC) of the laboratory and the Local System Coordinate (LSC) of the body segment (sensor location). These reference frames have right-handed coordinate systems and are both defined by three orthogonal axes (Fig. 2b).

The 3x3 matrix allows to transform the coordinates from the LSC to the GSC. It is possible to rotate vectors by multiplying them by a DCM, where \( Q_{\text{LSC}} \) and \( Q_{\text{GSC}} \) are vectors measured in the LSC and the GSC, respectively. The
relation between the DCM and Euler angles is shown in (1),
where \( sT \) and \( cT \) mean \( \sin T \) and \( \cos T \) respectively. 

\[
R = \begin{bmatrix}
  c \theta c \psi & s \psi & -s \theta c \psi - c \theta s \psi \\
  c \theta s \psi & s \psi & -c \theta s \psi - s \theta c \psi \\
  s \theta & c \phi & c \psi
\end{bmatrix}
\] (1)

R allows a vector to rotate measured in the LSC to the GSC. R matrix has orthogonally conditions. The signals of the gyroscope update the DCM. The rate of change of a rotating vector from kinematics shown in (2), where \( \omega(t) \) is the rotation rate vector obtained from Gyros in the LSC and \( r(t) \) the rotation vector in GSC.

\[
\frac{dr(t)}{dt} = \omega(t) \times r(t)
\] (2)

In (3), the rotation form GSC with approximate integration is shown. From LSC, the GSC is rotating equal and opposite to the rotation of the LSC. So, the GSC axes can track as seen in the LSC by flipping the sign of the gyro signals. The negative sign is combined with the cross product swapping the order of the operands.

\[
r(t) = r(0) + \int_0^t \omega(\tau) \times r(\tau) d\tau
\] (3)

In (4), the equation for updating the direction cosine matrix from gyro signals is shown.

\[
r_{gsc}(t + dt) = r_{gsc}(t) + r_{gsc}(t) \times d\theta(t)
\] (4)

The values of 1 on the diagonal of the matrix in (5) represent the first term in (4). The smaller, off-diagonal elements represent the second term in (4). The R Matrix is updated with incremental rotations every sample time.

\[
R(t + dt) = R(t) \begin{bmatrix}
  1 & -d\theta_x & d\theta_y \\
  d\theta_x & 1 & -d\theta_z \\
  -d\theta_y & d\theta_z & 1
\end{bmatrix}
\] (5)

Therefore, it is possible to get the orientation in (6) (roll, pitch and yaw) from the last row and the first column of the matrix in (1).

\[
\theta = -\sin^{-1}(R[3,1]) \\
\phi = \tan^{-1}(R[3,2]/R[3,3]) \\
\psi = \tan^{-1}(R[2,1]/R[1,2])
\] (6)

The final IMU design over the K53N512 microcontroller is shown in Fig. 3. The 3 Axis Gyroscope Data is the primary source of orientation. This angular rate output is integrated in every axis, in order to update the kinematic of the movement through DCM routine implementation. Additionally, this algorithm does a matrix normalization, in order to avoid that the numerical error in the integration would gradually violate the orthogonal constraint that the DCM must satisfy. It is necessary to perform adjustments to the elements of the matrix to satisfy the constraints. Fig. 3 shows the block diagram of the IMU sensor.

Fig. 3. Block diagram of the IMU sensor design.

In the same way, it is necessary to apply gyroscope drift detection, as the gyroscope offset will gradually accumulate errors in the DCM elements. This feature is realized in three stages: first, orientation reference vectors are used to detect orientation error by computing a rotation vector that will bring the measured and computed values of reference vectors into alignment. These vectors are obtained from two inertial sensors: a magnetometer is used to detect yaw error, and an accelerometer is used to detect pitch and roll. Then, the rotation error vector feeds a proportional plus integral (PI) feedback controller to produce a rotation rate adjustment for the gyros. Finally the rotation error is subtracted to the actual gyro signals. The system sampling frequency is established at 60 Hz to be compatible with the sampling of the optical tracking systems used in experimental validations.

### B. sEMG Sensor

The developed sEMG sensor has a mixed acquisition and processing signal through powerful Analog Measurement Engine and DSP capabilities of the Kinetics K53N512 microcontroller, and with an integrated instrumentation amplifier implemented with the microcontroller’s internal amplifiers. It consists of three operational amplifiers: two of them are used for buffering input signals which are transimpedance amplifiers named TRIAMP1 y TRIAMP2. The third one is for differential signal gain, which is an operational amplifier named OPAMP1 (Fig. 4).

Fig. 4. Block diagram of the EMG sensor design.

Although the instrumentation amplifiers have an excellent noise-rejection ratio, it is still necessary to filter some noise interferences, such as electric installation, movement, respiration, electromagnetic interference, and electromagnetic emissions from electronic components. A band-pass filter (2-500Hz) helps in those sources attenuation. This filter is supported by an operational amplifier named OPAMP2. There is a notch filter that helps to eliminate a specific frequency. In this case, the frequency
is 50/60 Hz. This filter use is specifically orientated to eliminate any noise related to an electrical power source. The microcontroller uses the signal coming out of the notch filter as a reference upon which a base level is set. In this application a delivery feedback signal from the output of the internal DAC is managed to adjust the base level. Finally, the signal resulting from all these blocks is sent to the microcontroller for sampling and measurement. Every one millisecond, a new 16 bits sEMG sample is obtained. The digitalized signal is stored as an array, in order to be filtered by a FIR Filter in the surface sEMG Band-Pass and thus complete a high performance noise elimination.

C. ZIMUED Network

The ZigBee network has three types of devices: End Devices (ZED), Routers (ZR) and only one Coordinator (ZC). We implemented a ZIMUED network based on ZHC profile and Freescale ZigBee Stack for the ZigBee chip MC13224. This network is a start configuration where a ZIMUED Coordinator receives patient’s signals data from several ZIMUED End Devices (Fig. 5). The network protocol proposed is based on defining a specific and unique period designated to the transmission of each ZED, thus improving the delivery packet success rate.

Additionally, it includes a small battery supply for portable applications.

The second board is a ZHC board (Dimensions: 41 x 48 mm) installed over the sensor board, shown in Fig. 7. It has an MC13224 microcontroller and there are three possible antennas to evaluate different environments for monitoring applications: an external SMA (SubMiniature version A) Antenna for extended distance communications, a Chip Antenna for reducing board size and an F-Antenna (Antenna painted over the board, it is shaped like the letter F) for reducing external components. This board has pushbuttons and LEDs for ZigBee network initializations. Additionally, it has different communication busses with the Sensor board through two connectors.

IV. System Implementation

The wearable sensor system implemented consists basically of two boards: Sensor Board and ZHC Board. These version boards were designed for evaluation. In the next version we are reducing the size and integrating only one single board.

The sensor board is shown in Fig. 6. (Dimensions: 43 x 60 mm) It has a KN53N512 microcontroller, which is connected by I2C bus with the inertial sensors for IMU processing. The sEMG sensor receives the signal from external electrodes to be connected to the front-end microcontroller. It has a USB connection for debugging purposes and massive storage capabilities for a micro SD Card. It has two connectors to integrate with a ZHC Board.

In Figure 8, a ZIMUED Network application is shown. In every ZIMUED End Device, the Sensor Board periodically sends the IMU and sEMG measures to the ZHC Board, which sends them, using ZigBee Communication as a star network to the ZIMUED Coordinator. It receives the measure packets of all End Devices, and sends the information to the PC, to be visualized and processed. The K53N512 and MC13224 are connected by I2C bus. The IMU measure is obtained by the K53N512 every 60 Hz, and consists of 3 bytes (roll, pith and yaw), then it sends the kinematic measure to the ZHC Board. The sEMG measure is obtained every 1 kHz in 16 bits. It stores 15 sEMG samples before sending them to ZHC Board in order to reduce the traffic and interruptions. All ZHC Boards start transmissions in its unique time in the ZIMUED network.

Fig. 6. IMU and EMG Sensor Board.

Fig. 7. ZHC Board over the IMU and EMG Sensor Board.

Fig. 8. ZIMUED Network Configuration.
Finally, the signals obtained from two ZIMUED End Devices are shown in Fig. 9, corresponding to a recording of the sEMG and IMU signals over the arm and forearm respectively, in continuous motion with a repetitive upper limb standard gesture of reaching and grasping. The continuation of this research contemplates the sEMG assessment in relation to motion analysis to quantify and evaluate muscular activity during the gesture as an additional tool in upper limb rehabilitation.

![Fig. 9. EMG signals from upper limb muscles and angles recorded by the IMU.](image)

V. EVALUATION

The evaluation was performed with simultaneous recording IMU-ZIMUED and an optical system for motion analysis. The gesture to reach and grasp an object was chosen because of the importance of this movement, as mentioned in the introduction. The sampling of the optical tracking systems is done at 60 Hz, during every frame the positions of the 10 reflective markers are registered, they were located at: C7: Cervical 7, ACL, ACR: Acromion Left and Right, EST: manubrium, OL_R: Elecranon law, ER_R: Radial styloid, EU_R: Ulnar styloid, 3M_R: Head of metacarpal 3, RAD_R Marker Radial Arm and TRA_R between styloid (Fig. 10).

![Fig. 10. Location markers and IMU sensor for simultaneous recording.](image)

The development of an anthropometric model to estimate joint centers of the shoulder, elbow and wrist is required to assess the positions of the segments; this work was based on the model proposed for Rab et al. [21]. In addition, the joint centers the generation of a local system of reference axes, embedded in the center of gravity of every segment, which is necessary to find the spatial orientation of each segment involved in the movement of reaching and grasping.

![Fig. 11. Sequence of images of the gesture of reaching and grasping and back to rest position. (bottom) Virtual reality for the patient feedback (biofeedback).](image)

The evaluation of the results consists of repetitions of the standard movement by several subjects being measured by both techniques. The movement of reaching and grasping began from the rest position with the forearm on the table, at angle of 90 degrees approximately with respect to the arm before reaching and grasping an object, and then returning it to starting position. In Figure 11 a detailed image sequence of the movement can be observed. At the bottom of the figure the visual feedback (biofeedback) to be submitted to the patient during rehabilitation is shown.

![Fig. 12. Registers of both IMU and Videography systems. Each color represents the same gesture acquired simultaneously. The IMU registers are shown in dashed lines, while solid lines correspond to the optical video system measurements.](image)

The complete data measured by the IMUs (consisting of 9 subjects with more than 10 repetitions in/of each one of them) was evaluated, calculating the elbow angle as a direct quantification of flexion-extension motion. The next figures show these evaluation parameters, all acquired from the IMUs system; first (Fig. 13), 14 curves of data for the same subject, followed by 9 curves, each one from a different subject (Fig. 14). High repeatability inter and intra subject is observed without artifacts or signal distortion.
VI. CONCLUSIONS

In this paper, we show a new tool for assessment and therapy in post-stroke upper-limb rehabilitation and new wireless sensor technology to enhance rehabilitation robotics. In the evaluation, the integrity of biomedical data and reliability in the communications was ensured; the wireless sensor nodes did not have any lost packet during the test. We also performed a comparison with optical system tracking, with successful results. It also demonstrates a huge potential both as a tool to provide visual feedback for robot-assisted arm training for Patients After Stroke,” IEEE Transactions on Neural Systems and Rehabilitation Engineering, vol. 16, no. 4, pp. 371-379, Aug. 2008.


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This system can be use in test and pathologic analysis in daily life activities and ergonomic analysis where it is difficult to implement with wired sensors and sEMG active electrodes. Finally, the Wearable ZigBee Sensor System in medical robotics can improve the bioinspired control in order to integrate the patients and robotic signals in different kind of therapies.

To continue with this project, we will test this wearable system with patients in the Sabana University Hospital, where we are working with stroke patients with dysfunction in upper limbs.

VII. REFERENCES