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A wireless, minaturized multi-channel sEMG acquisition system for use in dynamic tasks

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Abstract— Nowadays, the detection of surface EMG (sEMG) signals is almost exclusively based on a single or a few electrode pairs. However, in the last two decades limitations of bipolar sEMG signals emerged. To increase the amount and reliability of information extracted from sEMG, linear electrode arrays and two-dimensional detection systems have been proposed. The aim of this work was the development of a wearable wireless, 32-channels sEMG acquisition system. The developed system performs the conditioning, sampling and wireless transmission of 32 monopolar sEMG channels and 3 auxiliary signals, sampled at 2.048ksps with 16 bit resolution. The system wirelessly transmits the acquired signals to either a mobile device (smartphone or tablet with Wi-Fi connectivity) or a personal computer for real time visualization and storage. The developed system has been tested in clinical and sport scenarios showing good performances in wearability and movement artefact robustness.

Keywords— multi-channel surface EMG; wearable; muscle; rehabilitation; sport

I. INTRODUCTION

Nowadays, the detection of surface EMG (sEMG) signals is almost exclusively based on a single or a few electrode pairs for the purpose of (1) estimating muscle force, (2) providing biofeedback about the level of muscle activity, (3) identifying activation patterns of muscles (4), controlling a prosthesis, a robot or some artificial device.

However, in the last two decades limitations of bipolar sEMG signals emerged. Global signal features extracted from the interference EMG are influenced by both central and peripheral properties of the neuromuscular system. Moreover, several works in literature demonstrated heterogeneous activity within muscles during voluntary contractions. For these reasons, being able to sample the distribution of surface EMG with an appropriate spatial sampling seems essential [1].

Moving from isometric to dynamic contractions, the complexity of the interpretation of sEMG signals considerably increases. Changes in sEMG signals become related to the continuous modifications in the force output, muscle fiber length, and relative position of surface electrodes and sources occurring during a dynamic task [2].

To increase the amount and reliability of information extracted from sEMG, linear electrode arrays and two-dimensional detection systems have been proposed [3]. The use of multiple sEMG electrodes (High-Density surface EMG - HDsEMG) allows the extraction of anatomical and physiological information either at the muscle or at the motor unit level with applications in several fields ranging from clinical neurophysiology to the control of prosthetic devices.

Even though multi-channel sEMG open new perspectives in the non-invasive neuromuscular system assessment, several problems related to the signal acquisition and interpretation are still open especially in dynamic conditions. Among them the setup complexity and the lack of wearable multi-channel sEMG systems.

Several systems have been developed for specific applications (e.g. prosthetic control) [4] or to meet specific requirements such as minimizing energy consumption [5] or guarantee high reliability of the acquired signals in terms of noise reduction and rejection of the powerline interference [6].

The aim of this work was the development of a wearable wireless and miniaturized 32-channel sEMG acquisition system for use in dynamic tasks. The miniaturization of the system is particularly important to minimize movement artifacts in the case of dynamic contractions.

II. MATERIALS AND METHODS

A. General Archietrcture

Fig. 1 shows the high-level block diagram of the proposed system. The core of the system is a wireless miniaturized 32-channel sEMG amplifier (Sensor Unit - SU) for the monitoring of different muscles.

The Sensor Unit performs the conditioning, sampling and wireless transmission of 32 monopolar sEMG channels and 3 auxiliary signals, sampled at 2.048ksps with 16 bit resolution. The SU wirelessly transmits the acquired signals to either a mobile device (smartphone or tablet with Wi-Fi connectivity) or a personal computer for real time visualization and storage. The connection with the receiver is implemented through a wireless access point. The system is entirely realized using COTS (commercially available off-the-shelf) components.

B. Sensor Unit Design

The Sensor Unit module consists of four main building blocks showed in Fig. 2: (1) the bio-signal acquisition unit, (2)

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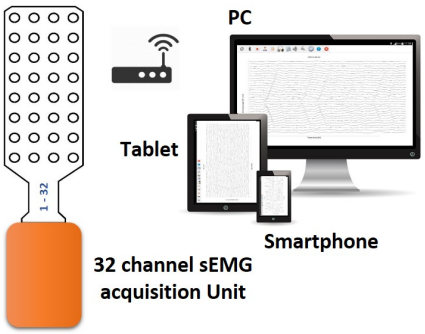

Fig. 1. System architecture. The developed system is composed by one sensor unit that conditions, samples, and transmits up to 32 monopolar sEMG and 3 auxiliary signals sampled at 2048 ksps with 16 bit resolution and one mobile device (notebook, tablet or smartphone) that receives wirelessly the sampled signals. The transmitter and the receiver are connected through a wireless access point.

the Wireless Microcontroller Unit (MCU), and (3) the Power Management Unit. Each building block is described in more details in the following. The bio-signal acquisition unit implements the conditioning, sampling, and digitalization of up to 32 monopolar sEMG channels and three non-conditioned channels that can be used for the acquisition of auxiliary signals (such as biomechanical or trigger signals). It is based on the Intan RHD2132 chip (Intan Technologies).

The RHD2132 chip is a low-power and ultra-compact (size 9x9 mm) bio-signal acquisition system, integrating 32 monopolar AC-coupled analog front-ends with fixed gain (192V/V) and programmable bandwidth (0.1Hz – 20kHz), three non-conditioned auxiliary channels, a 35 channels analog multiplexer, a 16bit resolution successive-approximation (SAR) A/D converter and a SPI communication interface. Considering the characteristics of the sEMG signals, the chip has been configured with a bandwidth between 10Hz and 500Hz.

The sampling of 35 signals (32 EMG and 3 auxiliary channels) at 2.048ksps with 16 bit resolution requires a constant data throughput of about 1.1Mbps. In order to achieve the required data throughput and to allow the connection of the system to mobile devices without an ad-hoc receiver, the Wi-Fi transmission protocol has been chosen and the Texas Instruments CC3200 system on chip wireless MCU has been selected. The CC3200 device is a wireless MCU that integrates an ARM Cortex-M4 MCU core (Main Processor, MP) running at 80 MHz and an additional dedicated ARM MCU (Network Processor, NWP) that acts as a Wi-Fi network processor subsystem including an embedded TCP/IP stack. This characteristic allows running data sampling (Sampling task managed by MP) and data transmission (Transmission task managed by NWP) in parallel. The Sampling and Transmission tasks run on the FreeRTOS Real Time Operating System embedded on the CC3200 device.

In order to meet the requirement of 1.1Mbps constant data throughput and maximize it, a data packet of 20 samples per channel has been chosen. When 20 new samples per channel are available, MP creates the data packet and informs the NWP

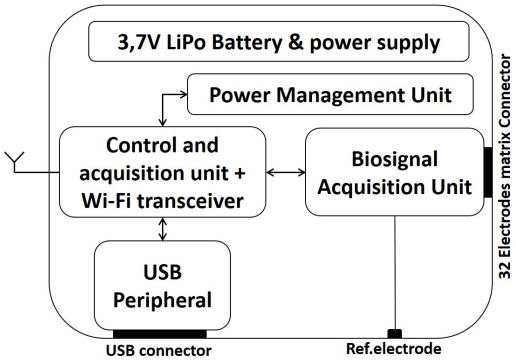

Fig. 2. Block diagram of the Sensor Unit. The control and acquisition unit is based on a wireless Microcontroller (Texas Instruments CC3200) It communicates with the Intan RHD2132 biosignal acquisition system (through a SPI bus) with the USB Peripheral interface and provides to send wirelessly the sampled data. The Biosignal Acquisition unit provides amplification (192 V/V), conditioning (10Hz-500Hz) and quantization of 32 sEMG signals plus 3 auxiliary signals. The Power supply unit provides a regulated 3.3V power supply while and the Power Management Unit allows to charge the system through the USB connector. The USB peripheral is also used for debug purposes.

a new data packet is available for transmission. Data packets are composed by 700 WORDs containing the sampled data (20 samples x 35 channels).

Because MP and NWP share the same RAM, a circular buffer 20 packet long has been used to avoid the simultaneous access to the same memory location from the Sampling and the Transmission tasks.

The system is powered through a 1-Cell LiPo Battery while the 3.3V regulated power supply voltage is obtained through a TPS62172 (Texas Instruments) DC-DC converter featuring an efficiency higher than 90%. An USB LiPo battery charger chip is embedded on the system in order to charge the battery directly from a USB port or a common phone charger. The USB communication, used for debug purposes, has been obtained using a serial port controller providing bridging between an USB port and the UART serial port of the CC3200 chip.

C. *Software Development*

A standalone software has been developed for the acquisition and online visualization of the sEMG signals on the receiver. The software has been designed using the C++ multi-platform Qt libraries (Android, Windows, Linux and MAC).

D. *Experimental tests*

The system has been tested in isometric and dynamic conditions. Signals in isometric conditions were recorded from the biceps brachii muscle during elbow flexion at 20% MVC using a 32 electrodes linear array with 5mm inter-electrode distance. In order to verify the effectiveness of the system in field applications, we tested it in sport and clinical scenarios. For sport scenario, we monitored the activity of the rectus femoris muscle during cycling on a bike. Signals were recorded using a 32 electrodes linear array with 5mm inter-electrode

distance. For the clinical scenario, the system was tested during the gait analysis monitoring the activity of gastrocnemius muscle using a 32 electrodes linear array with 5mm inter-electrode distance.

III. RESULTS

A. Bio-signal Acquisition Unit characterization

The band-pass gain of each channel was measured applying an 80Hz, 2mVpp sinewave to the input of the RHD2132 front-end and calculating the ratio between the output and the input amplitude (G_i=$Vout_{ipp}$/Vin_{ipp}). The measured in-band gain for each channel resulted 183±1V/V within a bandwidth between 10Hz and 500Hz.

The bandwidth of each monopolar front-end was measured applying a 2mVpp sinewave to the input and varying the input frequency in order to find those values corresponding to -3dB attenuation with respect to the nominal gain of each channel. The measured input voltage range resulted above 10 mVpp for each channel.

The noise level on each channel has been measured shorting the amplifiers inputs to ground, recording the signals for a 60s long session and calculating the corresponding RMS value. The mean noise resulted 1.8±0.2μV_{RMS}.

B. Wireless MCU and characterization Units

The generated Wi-Fi data throughput has been analyzed by means of a Wi-Fi packet sniffer and calculated as in Eq. (1).

$$Throughput = \frac{T_{TX}}{N_{RX} * 1000} \quad (1)$$

The throughput is expressed in kilo-bits per seconds (kbps) while T_{TX} and N_{RX} are the acquisition time (s) and the number of received bits respectively during a five minutes long acquisition session.

The resulting data throughput obtained sending only one fixed size data packet was 5.5 Mbps, allowing the transmission of up to 170 EMG signals sampled at 2ksps with 16-bit resolution. These results fit the goal of the project.

The loss of packets was tested by means of a packet sniffer during a five minutes long measurement session repeated five times. The distance between the SU (transmitter) and the router (receiver) was of 3m. The worst result corresponds to a packet loss of less than 0.016% (48ms in 300s).

The Wi-Fi latency can vary from some to a few dozens of milliseconds depending on the network load. To mitigate this phenomenon and use the system in biofeedback applications, we decided to use a dedicated access point for the communication between the system and the PC. Furthermore, the connection with the access point has been established on the Wi-Fi channel having the higher signal strength. With these precautions, the measured Wi-Fi latency was below 10ms. This result is not critical for biofeedback applications.

C. System prototype

The Sensor Unit design was kept as simple as possible using only commercially available off-the-shelf (COTS) components and minimizing the system's encumbrance in order to improve wearability and reduce the movement artifacts during signal acquisition. The prototype of the Sensor Unit module is composed by two interconnected PCBs (Fig. 3.a and Fig 3.b). The first PCB, contains the connector for the electrode array, the connector for the three auxiliary channels and the Intan RHD2132 analog front-end.

The second PCB contains the CC3200 microcontroller, the circuits for the USB communication, the power supply and the battery charging unit.

The system has been powered through a 1-cell LiPo battery with capacity of 370mAh. Two prototypes have been mounted and encapsulated in a 3D printed case. The total encumbrance of the system is 3.7cm x 3cm x 2cm. The measured current consumption was of 119 mA during continuous transmission. Considering the chosen battery, the estimated battery life is about 3 hours. The time required to recharge completely the sensor unit module is of two hours.

D. Experimental tests

Fig. 4.a and Fig 4.b show the sEMG signals acquired from one subject during an isometric elbow flexion using a 32 electrodes linear array with 5mm inter-electrode distance positioned on the Biceps brachii. Fig. 4.c shows the PSD of the channels number 2 and 20 calculated on the monopolar EMG signals over a 250ms long epoch. It is possible to observe the absence of appreciable power line interference.

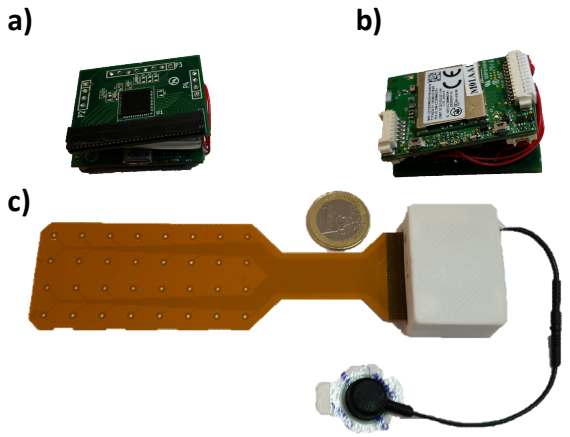

Fig. 3. The sensor unit prototype is composed by two interconnedted PCBs mounting the a) biosignal acquisition unit and b) the wireless MCU, the power supply and the battery charging circuit. The battery is mounted between the two interconnected PCBs. c) The system prototype was encapsulated in a 3D printed case and connected to a flexible kapton matrix (8x4 electrodes with 10 mm interelectrode distance). The overall system's dimensions are 37 mm x 33 mm x 20 mm.

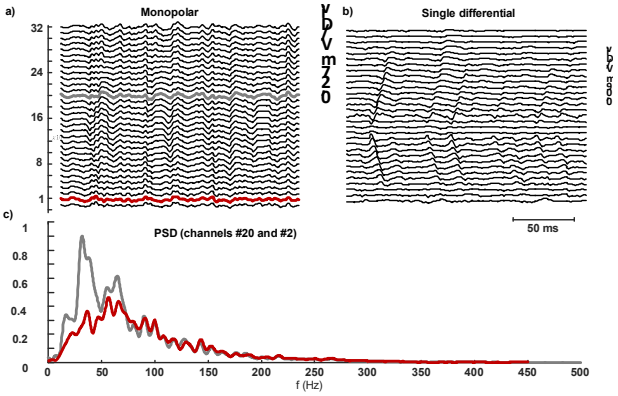

Fig. 5. Example of sEMG signals recorded from the biceps brachii muscle during an isometric contraction at 20% MVC. Signals were recorded using a 32 electrodes linear array with 5 mm inter-electrode distance On the top the monopolar and single differential signals are shown. On the bottom the Power Spectral Density (PSD) of the channels number 2 and 20 calculated on the monopolar EMG signals over a 250 ms long epoch is shown. It is possible to observe the absence of appreciable power line interference. From the differential signals it is possible to identify the innervation zone (between channels 13 and 14) and the propagation of single MAUPs.

IV. Discussion and Conclusions

This work describes the development and test of a wearable 32 channels sEMG acquisition system. The system is compact and when worn it does not interferes with the movements of the subject. The tests performed on the Intan RHD2132 single-chip analog front-end and the CC3200 wireless MCU satisfy the system requirements. Motion artifacts are almost absent also during dynamic tasks.

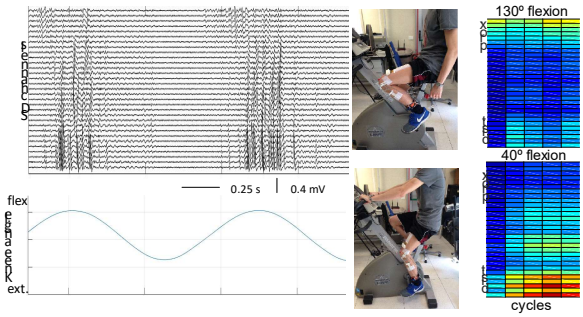

Fig. 4. Example of the use of the 32-channels sEMG system during cycling. In order to study the localized rectus femoris activation as a function of the cycling phase, a linear array with 32 electodes and interelectrode distance of 5 mm was positioned on the muscle covering its whole length. On the left the SD signals for two cycles together with the knee angle are showed. It is possible to appreciate the good quality of the signals without movement atrefacts. On the right are shown the RMS maps (125 ms epoch) estimated during five consecutive cycles corresponding to knee flexion angle of 130º and 40º. It is possible to observe a more proximal activation for 130º and a more distal activation for 40º.

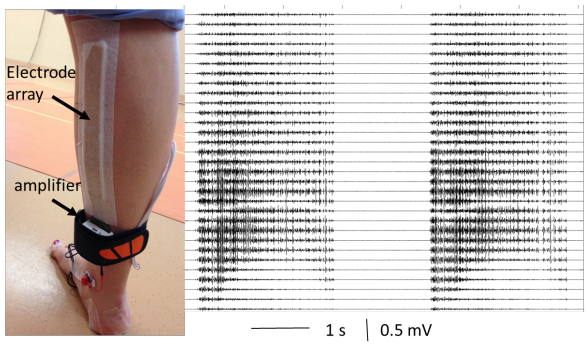

Fig. 6. Example of the use of the 32-channels sEMG system during gait analysis. In order to study the effect of electrode positioning on the estimation of activation intervals and intensity of gastrocnemiuos muscle during gait, a linear array with 32 electodes and interelectrode distance of 5 mm was positioned on the medial gastrocnemious muscle covering the whole length of the muscle and connected to the amplifier (on the left). SD signals recorded during two steps are reported on the right. It is possible to appreciate the good quality of the signals without movement artefacts and the effect of the shortening/lengthening of the muscle on the detected signal in the distal region.

Movement artefacts are due to (1) changes in skin thickness during movement (2) reciprocal movement of the gel-electrolyte skin interface and (3) cable movement. The two first causes can be reduced by scrubbing the skin and using Ag-AgCl electrodes. while the latter is due to the tribolelectric noise generated by the friction and deformation of cable insulation acting as piezoelectric transducer. The integration of the conditioning electronics as close as possible to the electrodes matrix without the use of connecting cables implemented in the proposed system allows to drastically reduce this problem. The developed wearable wireless 32-channel sEMG system open new perspectives in the use of HD-EMG in dynamic conditions for the study of neuromuscular system.

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