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DESIGN AND VALIDATION OF A LOW−COST WIRELESS ELECTROMYOGRAPHY SYSTEM

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ABSTRACT

Electromyography (EMG) is being explored for evaluating muscle activity. For gait analysis, EMG needs to be small, lightweight, portable device, and with low power consumption. The proposed superficial EMG (sEMG) system is aimed to be used in rehabilitation centers and biomechanics laboratories for gait analysis in Iraq.

The system is built using MyoWare, which is controlled by using STM32F100 microcontroller. The sEMG signal is transferred via Bluetooth to the computer (about 30m range) for further processing. MATLAB is used for sEMG signal conditioning. The overall system cost (without computer) is about $80. The proposed system is validated using wired NORAXON EMG using the mean root mean squared method. The results showed a root mean squared error between (0.216% − 43.14%).

Keywords: Electromyography, Wireless transmission, MyoWare, STM32F100 microcontroller.

INTRODUCTION

In recent years, electromyography (EMG) system was commonly used in biomechanical analyses, rehabilitation devices, sport, physiotherapy, and prosthetic control and where it is very important to evaluate the performance of the muscles throughout the task **(Hu *et al.*, 2009; Song *et al.*, 2008; Toro *et al.*, 2019)** based on electrical signal variations **(Farina *et al.*, 2004; Molinari *et al.*, 2006)**. In order to achieve the best outcomes, the system should have high accuracy with minimum movement restriction. Unfortunately, those equipment are not available in the rehabilitation centers in Iraq due to the higher costs **(Fuentes Del Toro *et al.*, 2019)**, as shown in (Table 1).

There are many different ways for EMG signal acquisition, the most widely used one is the superficial electromyography (sEMG) which is the less invasive method in comparison to the others, e.g. needles. sEMG is considered as valid as other methods by some researchers **(Adriano *et al.*, 2012)**, taking into account that the acquired signal must be cleaned.

Table (1): Comparison between the prices of commercial and the low cost EMG system.

Equipment		Prices ($)
Commercial Equipment	Delsys Trigno	(around) 23,440.96
	Cometa	(around) 17,580.72
Researchers Low-Cost Equipment	Bitalino	Up to 175.807
	Myoware EMG + Arduino Mega	117.219
	Myoware EMG + STM32F100	80

In the last few decades, implementing a portable, lightweight sEMG and other biosignals **(da Silva *et al.*, 2014; Ehrmann *et al.*, 2022)** with minimum noise possible was the goal for many researchers.

(Ohyama *et al.*, 1996) developed active, wireless sEMG electrodes accompanied with a built-in transmitter, which measure the signals without the need for any skin preparation due to impedance transformation. **(Yen *et al.*, 1999)** developed a wireless sEMG transmission system using an analog signal processor chip. **(Moon *et al.*, 2004)** designed a preamplifier and a wireless sEMG.

Even though the above mentioned studies showed successful development of wireless EMG system, all of the designs had limitation: either large size of the system limits the movement of the subjects or poor signal-to-ratio (SNR) of the system. **(Youn & Kim, 2009)** developed a successful wireless sEMG by using the tms320f2812 processor for A/D conversion and digital signal processing. Three electrodes were used to obtain the sEMG signal measurement consists of disposable, Ag/AgCl electrodes and a Bluetooth module for wireless data transmission. (**Fuentes Del Toro *et al.*, 2019)** designed a low cost wireless sEMG system using Arduino Mega board and MyoWare EMG sensor. Their proposed design was validated by comparing the results with commercial EMG system (Delsys Trigno). There is an excellent agreement between both systems, but there are also some limitations; the signal delay (<1 s) and noise due to the hardware and assembly in the proposed system.

This paper presents a lightweight, low cost wireless sEMG measurement system using Bluetooth communication technology. More filtering and processing techniques were used to reduce the noise presented in previous designs. The performance of the developed system was validated in comparison with it of a commercially available EMG measurement system, using the mean root mean squared (mRMS) values of the signal.

MATERIALS AND METHODS

Design considerations

For gait analysis applications, the EMG system should be small, lightweight, wireless, and accurate for better analysis and interpretation. To be a portable device, the power consumption should be small, so that it can be powered with batteries.

A non-invasive method of measurement using *Ag/Ag Cl* surface electrodes is used for convenience, and as a result, the proposed device must have high sensitivity and immunity to noise signals.

Hardware design

For accomplishing the design requirements, the system consists of two main blocks, which are data acquisition and data logger as shown in (Figure1 and 2).

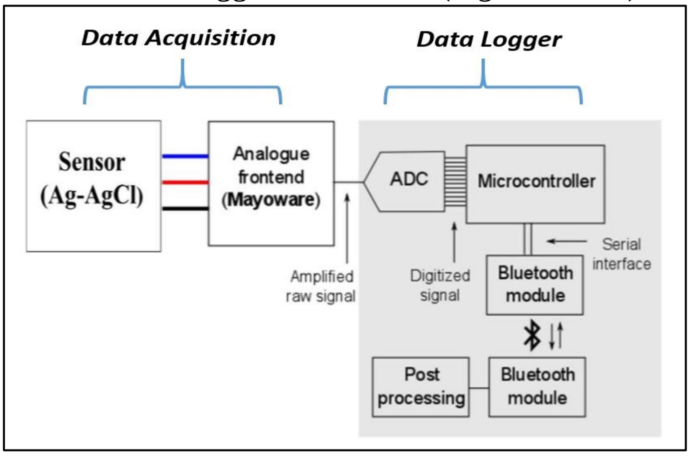

Figure (1): EMG system architecture.

The system includes:

1. Sensors (Ag-AgCl)
2. EMG (MYOWARE Muscle sensor)
3. STM32F100 Microcontroller
4. Data transmitter/ receiver (HC-05 Bluetooth)
5. Personal computer (Laptop) for data acquisition and analyses.

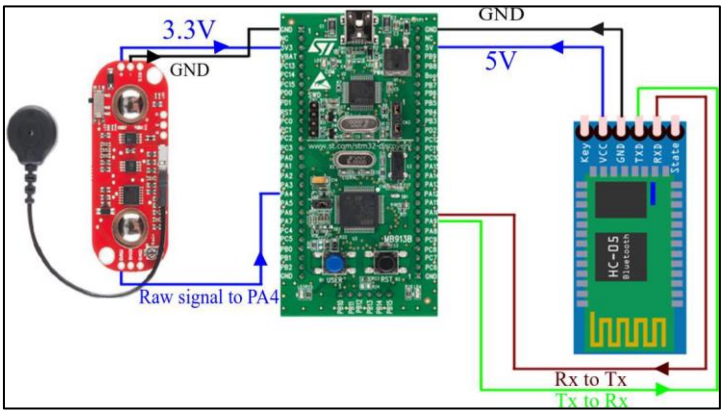

Figure (2): Hardware setup.

The overall system is shown in (Figure 3). MyoWare muscle sensor was used as a data acquisition equipment. It has an internal analog filter and amplifier, therefore, it provides a ready analog signal to be processed by the 12 bit ADC of the microcontroller board with a sampling rate of 1 kHz. The Bluetooth module is used to transmit the sampled data to the computer with baud rate of 115200 bps. The flowchart of the c code is shown in (Figure 4). The overall system runs on 5V from a rechargeable battery. The main characteristics of MyoWare EMG sensor, STM32F100, and HC-05 Bluetooth are listed in (Table 2).

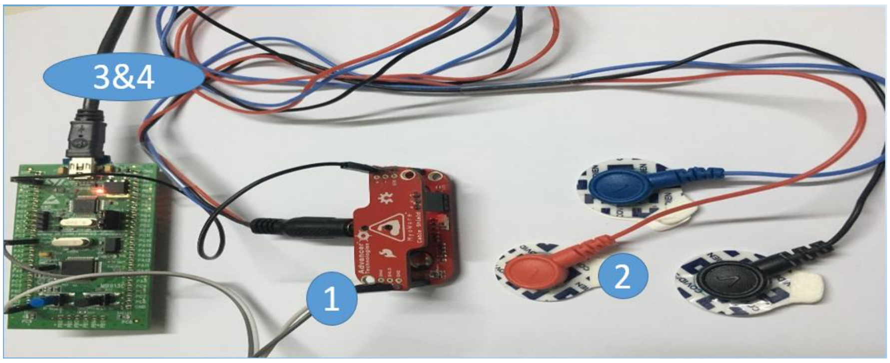

Figure (3): The proposed EMG system.1:MyoWare EMG, 2: *Ag/Ag Cl* surface electrodes, 3 and 4: STM32F1 microcontroller with Bluetooth (HC-05) (at the rare side).

Comparing STM32F100 microcontroller to the commonly used Arduino, STM32F100 has many outstanding features with lower cost. (Table, 3) summarized the comparison between the features of STM32F100 and Arduino.

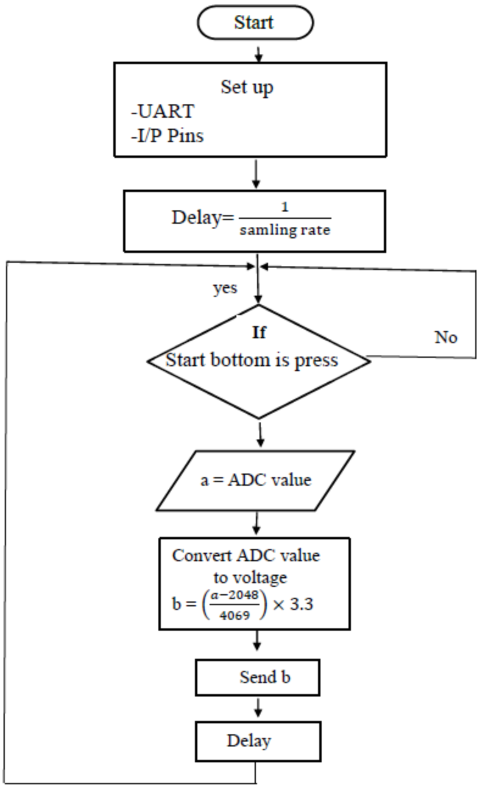

Figure (4): The flowchart of the microcontroller program.

Table (2): Technical specifications of STM32F100, MyoWare sEMG Sensor, and HC-05 Bluetooth.

STM32F100	
Microarchitecture	ARM Cortex-M3
CPU clock rate	24 to 72 MHz
V_{in}	5V
V_{out}	5V and 3.3V having inside regulator (it can support up-to100mA output current)
Flash memory	128 kBytes
SRAM	8 kBytes
ADC	12bit
MyoWare sEMG Sensor	
Supply	+3.1 to +5.9 V
Output modes	EMG envelope/Raw EMG
Size	2.08×5.23 cm
HC-05 Bluetooth	
Supported baud-rate	9600, 19200, 38400, 57600, 115200, 230400, and 460800 bit-per-seconds
Maximum range	30 m
Operating voltage	3.3V
Input power supply	3.6-6V , prohibit more than 7V
Weight	3g
Size	26.9mm × 13mm × 2.2mm
Flash memory size	8 Mbit

Table (3): Comparison between STM32F100 and Arduino.

Parameters	**STM32F100**	**Arduino**
Debugging	Yes	No
Bus architecture	32 bit	8 bit
Processing speed	Faster	Lower
ADC sample width	12 bit	10 bit
ADC resolution	High	Low

Digital signal processing of EMG signal

The received raw signal characterized by arbitrary positive and negative peaks, therefore, it requires further processing for interpretation. The digital signal analysis procedure (Figure 5) were performed offline using MATLAB (R2017a).

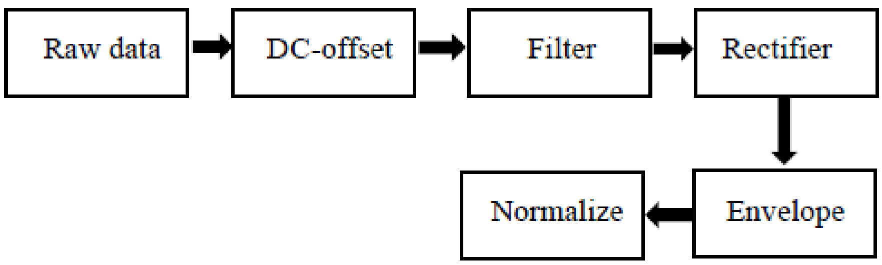

Figure (5): Block diagram for EMG signal analysis

First of all, the signal bias should be removed as it caused by the difference in electrical impedance between the skin and electrodes, the device components, and from the internal circuitry of the sensor (MyoWare). Offset cancellation involves subtraction the mean value of the signal from the signal, this is simply done by the MATLAB function *detrend*.

The second step involves filtering the signal using 2nd order, Butterworth bandpass filter with the range of 20 Hz to 500 Hz. The filter was applied in both forward and backward direction for phase shift cancellation.

To get the shape (envelope) of the signal, a full wave rectification is used. In this way, calculations will be more accurate. The rectified signal is low pass filtered to generate a linear envelope which is easier to interpret, also, it can be useful for possible future applications. For this purpose, a 4th order, double sided Butterworth, low pass filter with cut off frequency of 6 Hz was used. For comparing the EMG signal to a reference one, the signal is normalized to its maximum value (**Ghazwan *et al.*, 2017**), that is;

$$EMG_{normalized} = \frac{EMG_{envelope}(n)}{M}$$

Where M is the maximum value of the EMG signal (envelope).

EXPERIMENTAL WORK

The experimental procedure was performed in the Biomechanics laboratory at Prosthetics and Orthotics Engineering department/ Al-Nahrain University. Before applying electrodes, the skin was cleaned with alcohol to eliminate impurities and reduce skin impedance. EMG is recorded in differential mode, i.e. measuring the voltage difference between two electrodes. The electrodes were attached to the following:

A. Upper muscle Biceps branchii muscle during contraction.

B. Lower muscle gastrocnemius muscle during normal gait walking at self-selected speed.

The EMG signal is recorded for 10 seconds, from about 255 cm distance from the computer. The same experiment was repeated using NORAXON EMG system for performance evaluation. Both signals were processed using the same MATLAB program. For validation, mRMS (**Renshaw *et al.*, 2010**) was used for all recorded EMG signals.

RESULTS AND DISCUSSION

The output of signal processing steps using MATLAB (Figure, 5) are depicted in (Figure, 6). The output processed vector is the envelope of the random EMG signal. The envelope obtained using 4th order, double-sided low pass filter with 6 Hz cut off frequency. Although the shape now can be easily distinguished, it could not be compared visually with other tests without normalization. The results of the indicator-based comparison of the commercial and low-cost systems are listed in (Table, 4). In comparison to the commercial EMG, the mRMS measurement produced higher normalized activation intensities.

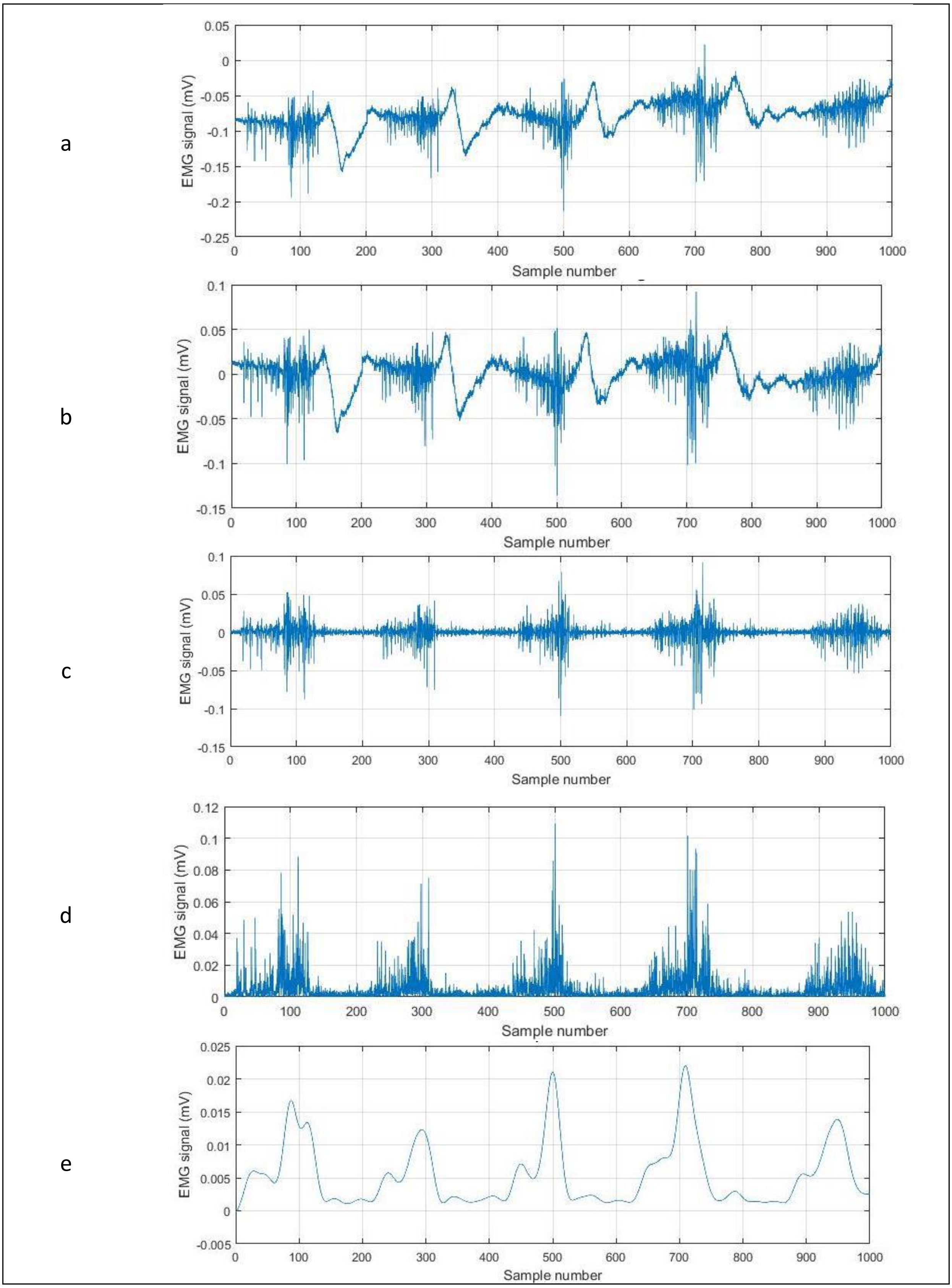

Figure (6): Digital signal processing of EMG signal. a: raw EMG signal; b: DC–offset cancelation; c: band pass filtered signal, c: signal rectification, d: envelope detection.

Table (4): The mRMS values of the portable EMG and NORAXON EMG.

Test	NORAXON	Proposed EMG	Error (%)
Test A			
Right	0.3283	0.4160	26.71
	0.3881	0.5113	31.74
Left	0.5164	0.5406	4.680
	0.3696	0.3704	0.21
Test B			
Right	0.4608	0.3581	22.28
	0.4619	0.2626	43.14
Left	0.5080	0.3422	32.63
	0.5031	0.3307	34.26

This study shows the design methodology of a low cost wireless sEMG system and MATLAB software as a signal processing software. The main idea is to be an economical alternative to commercial systems, while respecting the cutting-edge limitations of low-cost technology. Therefore, we assessed its reliability using commercial wired surface EMG system under controlled experimental conditions.

Two dynamic exercises were performed, Biceps branchii muscle contraction, as an example of low movement speed, and recording the gastrocnemius muscle contraction during walking as a relatively fast movement. The mRMS of the sEMG recordings showed a root mean squared error between (0.216 − 43.14%), where largest errors occurs in the fast movement. This means that the proposed sEMG is not adequately sensible to detect the peak to peak variations properly for fast movements.

The use of wireless sEMG system reduced the impact of two extrinsic noise sources, power line noise and cable motion artifact. The remaining noise is caused by electro-chemical noise at the skin-electrode contact and thermal noise caused by the electronics.

EMG signal is subjected to random errors that are inherent in the measurement process and are difficult to be eliminated no matter how carefully the experiment is conducted. For instance, electrode type, and performing the same exact movement during the experiment. As a result, when repeating the same experiment, different recordings were obtained. Further, difference filtering and processing techniques should be further explored such as the interference of the heart beat **(Strzecha *et al*., 2021)**.

It is worth noting that the mRMS is a reliable measure that reduces the effects of movement artifact, but it is less sensitive to changes in the EMG signal and may conceal variations in muscle activation intensity throughout test conditions.

CONCLUSION

The wireless EMG validation shows promising results. However, the system needs further improvements by means of noise reduction and using inline signal visualization and processing. The system can be further developed by combining further processing and classification techniques. After all, the proposed wireless sEMG is suitable for biomechanics laboratories and rehabilitation centers.

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