Single-channel electromyography based on Arduino for analysis of the swing phase in normal gait

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Single-channel electromyography based on Arduino for analysis of the swing phase in normal gait

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***Abstract*—This paper presents a thorough evaluation of a single-channel electromyography (EMG) signal acquisition board based on Arduino technology to characterize the swing phase of normal gait. A total register of 231 EMG signals corresponding to 11 lower limb muscles (7 right and 4 left) of 21 subjects was generated. For each record, the subject made a coordinated paused walk based on a sequence of audible tones to indicate each step. The maximum amplitude (MA) defined as the maximum value of the amplitude of the rectified and smoothed EMG signal corresponding to each of the swing sub-phases (initial, mid and terminal) were compared for the left and right limbs. The MA between swing sub-phases were also examined. Results show that the amplitudes obtained in each swing sub-phase of the gait records are coherent with the normal swing phase. These findings allow to recommend the use of the EMG acquisition prototype for studies addressed to the detection of motion intention.**

***Index Terms*—EMG, lower limbs, swing phase, motion intention**

I. Introduction

Nowadays the development of lower limb exoskeletons is a very popular research topic of wearable robots since it emerges as a promising tool in physical rehabilitation with the aim of recovering or improving gait. Several studies have been made in the development of robotic exoskeletons, such as a generalized control system for lower extremity exoskeletons based on hybrid control [1], the design of a device for knee rehabilitation [2]; the design of an exoskeleton with three degrees of freedom and the development of a control strategy based on torque feedback, which adapts to the movement requirements of the user [3].

EMG signals are one of the most useful biological signals for the control of wearable devices by detecting the intention of the user's movement [4]. EMG signals processing has allowed the development of numerous clinical applications and devices, such as gait analysis, muscle rehabilitation, orthetic devices and prostheses control among others [5]–[7].

Walking enables the displacement of the body's center of gravity with minimal energy demand and requires the integrity of the complex neuromuscular network. Gait analysis includes two main phases: stance and swing. Stance phase occurs when the foot remains in contact with the floor, so the weight transfer of the body to the opposite limb occurs, while overtake on the support foot. This phase constitutes 62% of the cycle. Swing phase, occurs when the foot loses contact with the floor, the limb remains in the air and advances forward, representing 38% of the cycle [8], [9]. During the swing phase, the muscular action responds primarily to the main functions of each one of the muscles involved, whether to accelerate or decelerate. The stability and synergy of the hip, knee and ankle joints, as well as the trunk and upper extremities are fundamental during the entire gait cycle.

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In this work, a single-channel EMG signal acquisition board based on Arduino technology is assessed [6]. The main drawback of this system corresponds to its sequential data acquisition feature in contrast with the simultaneous data acquisition of more robust platforms, which demands to repeat the walking sequence for the acquisition of a single record for each muscle. The device is evaluated for the analysis of muscle activation in the swing phase, which is an important process in determining the start of the gait cycle and the intention of the neuromusculoskeletal movement. The electromyography was evaluated using an EMG database recorded to determine the threshold of the intention of movement of lower limbs during gait [10], [11].

This paper is organized as follows: Section II describes the single-channel electromyography acquisition system. Section III presents the results. Section IV a discussion of the findings. Section V presents the conclusions.

II. Methods

A. Single-channel electromyography

The EMG signal acquisition system has a general scheme divided into four stages: transduction, amplification, A/D conversion, and visualization (Fig. 1). A brief description of every stage follows:

- *Transducer*: Ag-AgCl surface electrodes were used in the transducer stage with a passive electrode cable, used for reasons of compatibility with the shield EKG-EMG. This cable connects the electrodes with the processing board, which is composed by an audio shielded cable with a stereo connector in one end, and in the other, three crocodile connectors.
- *Amplifier*: The shield EKG-EMG is an extension module manufactured by Olimex, and compatible with Arduino. The board converts 2-point analog differential signals

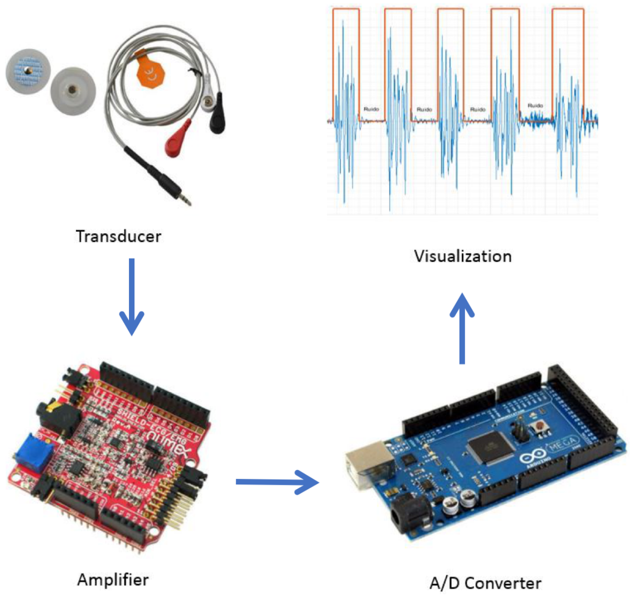

Fig. 1: EMG signal acquisition block diagram

into a single data stream as output (a single channel). The board architecture consists of open hardware and software; therefore, users have access to all design documents. The board total gain is the product of the gains of each discretization stage. By default, the total gain is about 2848. The maximum analog voltage = 3.0V/2848 ≈ 1mV [12].

- <u>A/D converter</u>: The DAC of the Arduino Mega 2560 is used with a resolution of 10 bits and a conversion rate of 1 KHz.
- <u>*Visualization and storage*</u>: This stage is developed in MATLAB by using serial communication between the PC and the Arduino microcontroller at a rate of 115200 bauds.

B. Experimental methodology

A total of 21 subjects (14 men), age (21.52 $\pm$ 2.4 years) without pathologies in the lower limbs participated voluntarily in the experimental tests. The purpose of the study and the experimental procedure were explained to all the subjects. All the participants read and signed an informed consent prior to the data recording.

EMG signals were recorded from 7 muscles of the left lower limb: rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), sartorius (S), biceps femoris (BF), semitendinosus (ST) and semimembranosus (SM); and 4 muscles of the right lower limb: rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF) [13], [14].

Three superficial electrodes per muscle were used, which were placed by a clinical expert. The pair of differential electrodes were located at the motor points of each muscle, while the reference electrode was placed on the back of the knee. Each test subject made a coordinated paused walk based on a sequence of audible tones that indicated the time intervals for stepping forward. All the tests consists of 10 or 11 steps. Each step started always with the left leg. The records were made sequentially. The average one-channel EMG acquisition time was 58 $\pm$ 3 s. In total, 231 EMG records corresponding to 11 muscles (7 right and 4 left) of 21 subjects were processed. The annotations [1] of contraction episodes, both activation and rest of the EMG signals, were made visually by two experts. The contraction episode corresponds to the swing phase (Fig. 2) which is approximately 38 % of the gait cycle.

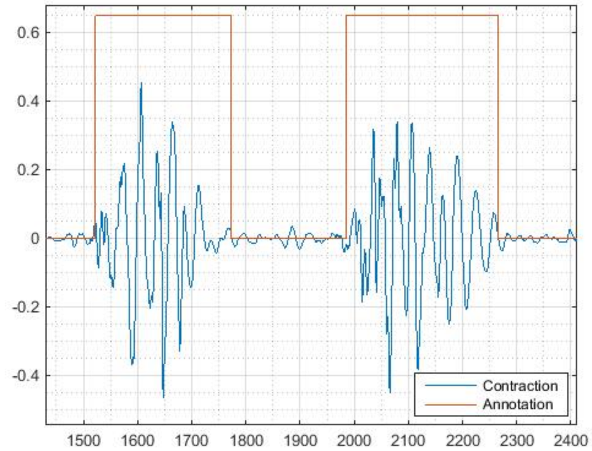

Fig. 2: EMG signal of Rectus Femoris (RF) muscle: contractions and annotations

C. The three sub-phases of the swing phase

The swing phase corresponds to 38% of the gait cycle and includes three sub phases [9], [15] (Fig. 3):

1) <u>Initial Swing</u> (IS): occurs when the toes leave the floor and ends when the knee reaches maximum flexion during walking. The thigh is directly below the body and parallel to the contralateral lower limb, which at this moment supports the body weight. IS correspond to 34 % of the swing phase. The muscles involved in the IS are: *knee joint* (BF, and S); *hip joint* (S and RF).
2) <u>Mid Swing</u> (MS): starts at the maximum flexion of the knee and culminates when the tibia is placed perpendicular to the floor. MS corresponds to 26% of the swing phase. The muscular activity of the MS involves: *knee joint* (BF, ST, and SM); *hip joint* (psoas major and iliac).
3) <u>Terminal Swing</u> (TS): begins in the vertical position of the tibia, continues as the knee extends fully and ends when the heel makes contact with the floor. TS correspond to 40% of the swing phase. The muscle activity involves: *knee joint* -quadriceps (RF, VM, and VL), and hamstrings (BF, ST, and SM)- *hip joint* -hamstrings (BF, ST, and SM).

[1] An annotation hereafter is defined as the beginning and ending of a step in a time scale, which is coordinated with the audible tones.

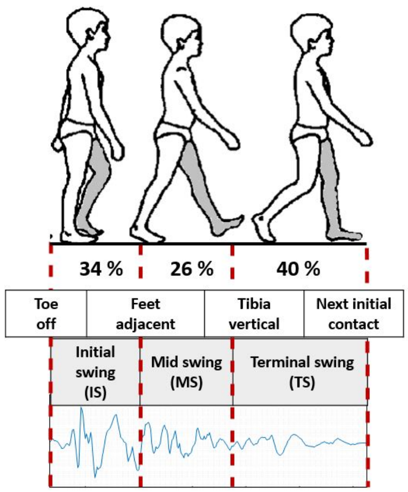

Fig. 3: Swing phase

D. Maximum amplitude estimation

MA is defined as the maximum value of the amplitude of the rectified and smoothed EMG signal corresponding to each of the swing sub-phases.

The methodology for estimating the amplitude values of the EMG signals can be summarized in the following stages:

1) Preprocessing: each EMG signal passes through a band pass filter (25-450 Hz), then through a rectification process applying the absolute value in order to obtain a completely positive signal, finally a mobile averaging window $(n = 25)$ is applied. In this way, a smoothed EMG signal is obtained.
2) Sub-phases: from the EMG annotations, the start and end points of each contraction were obtained, each swing phase was divided into three stages IS, MS and TS as explained in section II-C.
3) MA estimation: MA was estimated for each segment (IS, MS, TS). For all subjects $3N$ values of MA corresponding to the 3 segments of his/her N contractions are obtained; from these values the median is calculated as the value reported for each subject. This value is obtained for each of the 4 muscles (RF, VL, VM and BF from right and left extremity) in a methodical way for every subject from the 21 records.

Wilcoxon rank test were used to measure the differences between MA muscle (left and right) and between subphases (IS, MS, TS). Statistical significance was accepted at $p < 0.05$.

III. Results

Table I presents the average MA values for the four muscles analyzed and the p values between the right and left limb. Fig. 4 shows the variation of MA as the swing phase advances through its three sub-phases (IS, MS, TS). It is important to note that the left limb reaches greater amplitudes in the first two sub-phases. Fig. 5 shows MA for S, ST and SM in each swing sub-phases, it can be seen that in the MS phase there is a lower variance of amplitudes in comparison with the other two phases.

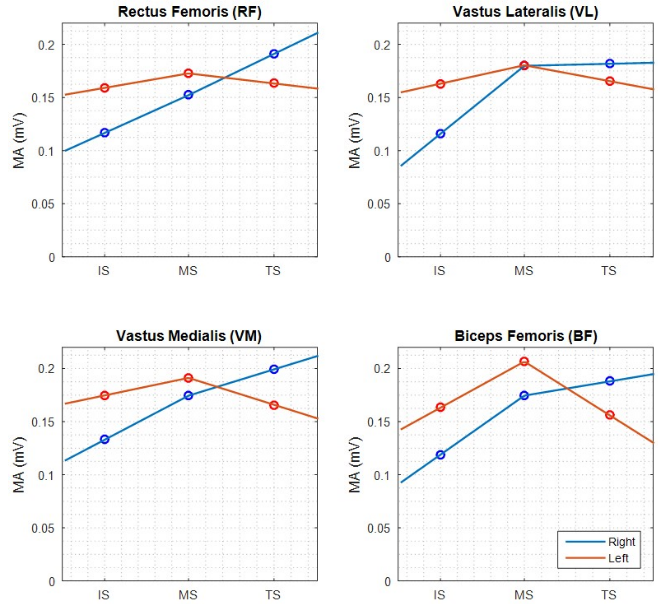

Fig. 4: MA values for muscles: Rectus femoris (RF), Vastus lateralis (VL), Vastus medialis (VM), Biceps femoris (BF)

IV. Discussion

In the IS phase, MA of the left leg muscles is significantly greater especially for the biceps femoris (BF) and vastus lateralis (VL), this is due to the fact that the hip flexor moment generated by these muscles favors knee flexion and with this, the advancement of the lower extremity; as well as the vastus medialis (VM) and rectus femoris (RF) since the concentric action of these muscles promotes the angular acceleration in the direction of the flexion of the hip and knee joints.

In the MS phase, this difference is only observed for the BF, which is expected, since it is the most involved muscle during this phase because it starts to act eccentrically to slow the extension of the knee.

During the TS sub-phase, all the muscles studied are involved, however the difference becomes smaller and is significant for two muscles: biceps femoris (BF) and vastus medialis (VM), since in the first half of this phase they slow down the flexion of the muscles. The knee promotes joint stability, in the second half they contract concentrically to facilitate the full extension of the knee in synchrony with the pelvis and prepare the next initial contact. As expected, the average amplitude of the left extremity is greater than that of the right as each step starts with the left leg.

During the IS sub-phase the sartorius muscle produces the knee and hip flexion. The sartorius presents no changes in the MA, possibly because is the longest muscle in the human body and its MA is large during the swing phase. The SM and ST muscles hardly variate its MA during the IS sub-phase, while during the MS sub-phase they have an eccentric action to decelerate the extension of the knee. These facts are coherent with results observed in Fig. 5. SM increases between IS and MS $(p = 0.15)$ and decreases between MS and TS $(p = 0.06)$; Semitendinosus (ST) increases between IS and MS $(p < 0.05)$.

TABLE I: Maximum Amplitude (MA) for each muscle: Rectus femoris (RF), Vastus lateralis (VL), Vastus medialis (VM) and Biceps femoris (BF). Swing sub-phases: Initial Swing (IS), Medium Swing (MS) and Terminal Swing (TS)

Muscle		**MA(mV)**			**p**		
		IS	***MS***	***TS***	***IS***	***MS***	***TS***
RF	R	0,1166 ± 0,3837	0,1522 ± 0,4027	0,1910 ± 0,3627	0,0012	0,0527	0,0527
	L	0,1591 ± 0,4891	0,1727 ± 0,407	0,1632 ± 0,5572			
VL	R	0,1159 ± 0,5214	0,1798 ± 0,4697	0,1817 ± 0,3714	0,0035	0,5629	0,0663
	L	0,1630 ± 0,454	0,1804 ± 0,3161	0,1654 ± 0,3249			
VM	R	0,1330 ± 0,4741	0,1745 ± 0,399	0,1992 ± 0,3476	0,0018	0,1249	0,0103
	L	0,1747 ± 0,3941	0,1912 ± 0,3758	0,1659 ± 0,3756			
BF	R	0,1189 ± 0,4921	0,1745 ± 0,4544	0,1879 ± 0,4076	0,0061	0,0346	0,0147
	L	0,1632 ± 0,5264	0,2066 ± 0,4457	0,1559 ± 0,4223			
S	L	0,1654 ± 0,4508	0,1649 ± 0,4183	0,1870 ± 0,4934			
ST	L	0,1543 ± 0,5674	0,1970 ± 0,2728	0,1853 ± 0,4640			
SM	L	0,1723 ± 0,5662	0,1882 ± 0,3408	0,1605 ± 0,4548			

The EMG signal acquisition methodology by using the single-channel board used in this work has several limitations: it is not possible to analyze the signals simultaneously, protocol and acquisition is long and repetitive, and subjects come to be fatigued, which causes a slight deficiency in the quality of recorded signals. However, the acquisition scheme and processing signal proposed in the present work has allowed to describe satisfactorily the swing gait cycle.

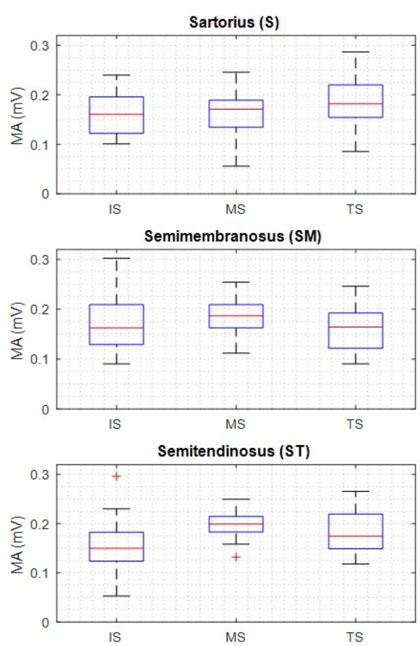

Fig. 5: MA values for muscles: Sartorius (S), Semitendinosus (ST) and Semimembranosus (SM) in each swing sub-phases

V. Conclusions

The results obtained show that the EMG acquisition prototype allows to characterize the swing phase. This finding is fundamental in the development of exoskeletons since it will allow the motion intention detection, which coincides with the beginning of the swing phase. It is expected that the prototype can be used for research purposes to analyze the intention of movement of the muscles involved in the flexion of the knee during gait. In a near future, five extra channels will be appended to this system, since the Olimex's Arduino stackable shield is extensible to 6 channels.

Acknowledgment

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