Muscle fatigue estimation based on optical-fiber FMG sensors for FES rehabilitation systems

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Abstract: Systems designed to rehabilitate persons who have suffered a stroke or spinal cord injury (SCI) still rely on extensive physical therapy to recover. However, functional electrical stimulation (FES), in tandem with exoskeletons, or wearable robotic gloves, are consolidating as essential resources to help to address this condition. In contrast, for this type of rehabilitation to be effective, it is crucial to have muscle fatigue feedback, especially because SCI patients cannot feel muscle soreness. Since the efficacy of this treatment is limited by the ability of muscle fibers to perform a specific induced task, this work proposes a muscle fatigue estimator that relies on an optical-fiber-based force myography (FMG) transducer to ensure that electrical stimuli do not contaminate the muscles contractions captured by the transducer. In this way, this feedback module benefits since its activation method is entirely mechanical, presenting advantages compared to traditional methods based on surface electromyography (EMG). The results yielded satisfactory results, validating the relationship between the energy degradation of each muscle contraction with the time in which a group of muscles fibers was induced to perform a specific task.

Keywords: force myography (FMG); functional electrical stimulation (FES), muscle fatigue, rehabilitation

INTRODUCTION: The strongly impairing condition present in persons who survive a stroke or SCI leads them to complete or partially lose motor functions. The rehabilitation process of this type of condition is still addressed through conventional exercise therapy that brings the patients back to their psychophysical well-being and allows them to reintegrate into activities of daily living (ADLs).[1] The development of clinical neuro-science, robotics, and biomedical engineering, started to play an essential role in improving the quality of life of impaired people. In this way, an increasing number of assistive devices, such as hand exoskeletons or wearable soft robotics, have been developed to increase the effectiveness of the therapy and, at least, to assist the impaired ones to perform their ADLs.[2–5] However, while exoskeletons have some advantages in hand rehabilitation by supporting and stabilizing the joints, they have some drawbacks since they do not prevent muscle atrophy, primarily if the impaired does not try to perform the exercises himself.[6]

FES utilization in the rehabilitation process has been gaining importance as an intervention to improve rehabilitation outcomes,[7–9] mainly when used in conjunction with therapy.[10,11] The use of this technology presents difficulties in achieving repeatability and precise movements and can also be painful for the patient[12]; however, it brings satisfactory results, increasing the range of motion of the impaired ones, reducing muscle spasm, and retarding disuse atrophy.[10,13,14] However, the extended use of FES in the rehabilitation process has limitations due to muscle fatigue. This reaction occurs because this technique induces unnatural motor unit recruitment order, imposing muscular activation, that causes the degradation of the force over time, preventing the completion of the rehabilitation process.[14]

Several works present fatigue estimators mainly based on EMG for torque estimation techniques to feedback this parameter to close the control loop.[15–19] However, its main drawback relies on the complex hardware and software needed to filter and process the signal to collect samples from a muscle contraction and then estimate the fatigue correctly—this isa disadvantage because electrical stimuli artifacts from the FES system interfere with the contraction acquired through the electrodes. In this way, methods based on near-infrared spectroscopy (NIRS) have been proposed to monitor muscle fatigue through oxygenation of the muscles. However, the complexity of the system is still one of his drawbacks.[20,21] Other works are based on Mechanomyography (MMG) since they only reflect the mechanical behavior of a group of muscles and do not be affected by FES artifacts.[22,23] For these reasons, we propose a fatigue estimator that relies on optical-fiber-based FMG transducers. These sensors also monitor the mechanical behavior of the muscles without compromising the integrity of the force measurements from each contraction. In this way, the energy of each contraction can be calculated on real-time and analyzed in order to determine that the force exerted by the muscles decreases over time due to muscle fatigue produced by the artifacts of electrical stimulation. This work aims to validate the utilization of this type of transducer to estimate muscle fatigue in a portable module, under real-time scenarios, and during FES-based rehabilitation processes.

The rest of this article is structured as follows: Section II elaborates on the system architecture and the hardware employed to design the fatigue estimator module. Methods

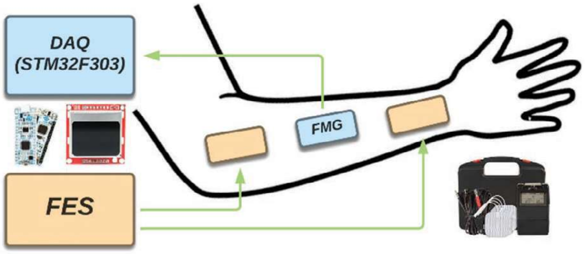

FIGURE 1 Roscoe Twin Stim Unit with two electrodes and one FMG transducer connected to the Data Acquisition Module

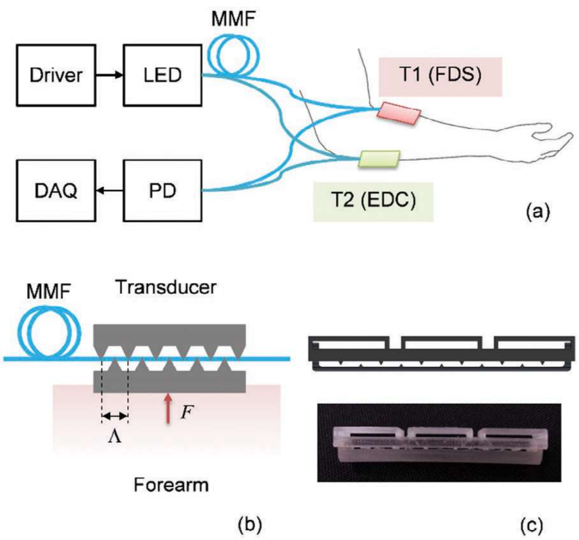

FIGURE 2 (A) Schematic of the FMG sensor: light emitted by a LED source is launched into multimode fibers (MMF). Radial forces exerted by forearm muscles excite the transducers (T1 & T2) and modulate the light intensity measured by a photodetector (PD). Microbending device: (B) schematic and (C) photograph

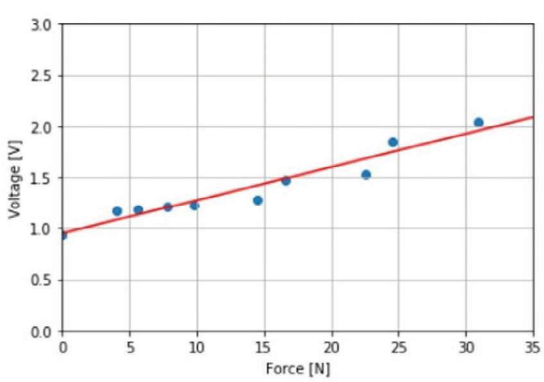

FIGURE 3 Optical fiber FMG sensor characterization curve. The data points are the average of 10 measurements, whereas the solid line is a linear curve fitting

to process and analyze the FMG signal are described in Section III. Finally, experimental results and conclusions are presented in Sections IV.

System architecture: The data acquisition module of the system was implemented with an affordable and high-performance microcontroller unit (MCU) based on the ARM Cortex-M4F architecture (ST Microelectronics STM32F303K8), a lithium coin cell of 3 V and Nokia 5110 LCD to display the energy measurements of each contraction. In addition, the electrostimulation induced each contraction through two surface electrodes of (EMS/TENS) placed on the flexor digitorum superficialis, as shown in Figure 1.

Optical-fiber-based force myography sensor: Force myography is a mechanical technique in which muscles contractions involved, particularly in hand movements, are retrieved from radial pressures exerted by the forearm muscles. Thus, FMG transducers require a more straightforward and lower-cost measurement setup compared to EMG, since in several cases, it is quite helpful to acquire information in terms of the force exerted by the muscles (e.g., muscle fatigue estimation).[24] The optical fiber sensors present lightweight, high sensitivity, low hysteresis, and immunity to electromagnetic interference, making these devices suitable for rehabilitation and human-robot interaction systems.[25–27]

The optical fiber sensor is shown in Figure 2A, is composed of an 820-nm LED source (HFBR-0400, Agilent Technologies), whose light is launched into multimode silica fibers of ~2 m long and 62.5/125 core/clad diameters. The optical signal is modulated by the optomechanical transducer (Figure 2B). The output intensity is measured by a photodetector (HFBR-24X6, Agilent Technologies), which signals amplified filtered by a signal conditioning stage and subsequently processed in the MCU.[28] The transducer (Figure 2B) consists of a microbending device of $L = 60$ mm length and with a $\Lambda = 10$ mm of periodicity. As the corrugated structure of the device mechanically deforms the waveguide of the light, the core-guided light modes are coupled to radiation modes yielding optical losses; therefore, the output light intensity can be correlated to the input force or displacement produced by a muscle contraction.[29] The microbending approach demands a more straightforward setup, with a dynamic range and stability.[30] The mechanical part of the transducer was fabricated using 3D printing technologies with acrylonitrile butadiene styrene (ABS) filament, providing an ergonomic device that can be placed on specific group of muscles using of Velcro straps.

The characterization curve of the sensor (Figure 3) was obtained by applying controlled displacements with

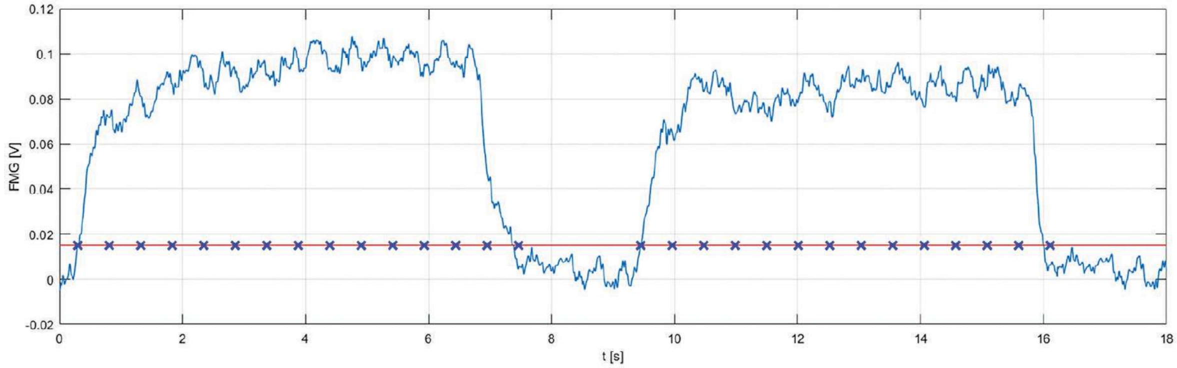

FIGURE 4 Two consecutive muscle contractions were induced using FES and acquired by the FMG transducer. The solid red line represents the predefined thresholds Thon and Th_{off} to detect a contraction, whereas blue crosses describe the beginning of each 512-sample ping-pong buffer

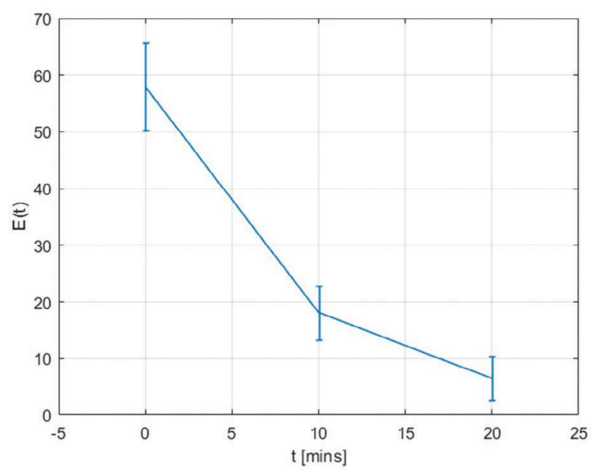

FIGURE 5 The means and standard deviation of the energy of the FMG signal measured in three 10-min intervals for three different subjects

a micrometric stage, yielding ~32.499 mV/N sensitivity within 0 to 31 N range and a correlation of 0.93. In this manner, the transducer provides an easy way to measure the force exerted by a specific group of muscles without extensive signal processing techniques. In addition, the low sensitivity for small forces can be compensated by applying a preload to shift the sensor characteristics to the linear range.

Signal processing and analysis: Two channels of FMG signals can be collected using the on-chip ADC with a sample rate of 1 kHz. Thus, a time window of 32 samples of the signal is processed through a single-threshold method to detect the On and Off timing of the muscles. In this way, one can determine an individual muscle contraction by comparing the RMS value with predefined on (Th_{on}) and off (Th_{off}) thresholds whose values depend on the mean power of the background noise of each channel. Once the muscle contraction is detected, ping-pong buffers of 512 samples (as shown in Figure 4) are used to determine the signal's partial energy using the following expression.

$$E_{i,n} = \sum_{k=0}^{N-1} |x_{k,n}|^2 \tag{1}$$

where $x_{k,n}$ is the sample k of the channel n, N is the size of the collected window and $E_{i,n}$ is the energy measured from the ith ping-pong buffer collected from a muscle contraction. Therefore, the total energy of a contraction measured by the channel n can be determined using the following expression.

$$E_n = \sum_{i=1}^{M} E_{i,n} \tag{2}$$

with M as the total number of ping-pong buffers that comprise a large muscle contraction, whose on and off timings were determined by the double threshold method. In this way, the total energy of each contraction is displayed on the Nokia 5110 to present visual feedback to physical therapy staff. Furthermore, for these trials, the EMS was configured in synchronous mode, with a T_{on} of 8 s, a T_{off} of 1 s, T_{ramp} of 500 ms, a pulse width of 300 μs, pulse frequency of 60 Hz, and a current intensity of 10 mA stimulating the flexor digitorum superficialis (only index, middle and ring fingers flexion) of five unimpaired arms from male subjects between 20 and 30 years old for a period of 20 min in order to reach controlled muscle soreness in each patient.

RESULTS AND CONCLUSIONS: The results of these trials are presented in Figure 5. During the tests, energy measurements were taken at the beginning, 10 and 20 min after. Energy measurements were taken only of the first five consecutive muscle contractions at the three time intervals (0, 10, and 20 min), obtaining 57.9578, 18.0914, and 6.4367with a standard deviation of 7.7438, 4.7547 and 3.9518, respectively. Moreover, the muscle fatigue began to manifest itself between 18 and 20 min after starting the tests for the three subjects, showing a presumably exponential relationship between fatigue and the decay of energy measured in each contraction. Therefore, it has been shown that the proposed module can provide meaningful feedback to rehabilitation systems based on FES, especially for patients who suffered a stroke, not only for therapists but also to close the control loop on complex systems. Finally, the FMG-based module provided promising performance in fatigue estimation over time; however, it needs to be validated in real FES-based physiotherapeutic scenarios or tandem with a robotic assistive device, such as presented in and compared with traditional EMG-based methods.

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Motimove: Multi-purpose transcutaneous functional electrical stimulator

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Abstract: Commercial transcutaneous functional electrical (tFES) stimulators nowadays are most often an integral part of the training/orthotic device. There is only one commercial portable tFES stimulator that provides limited flexibility in different FES applications and supports research work. MotiMove is a new versatile tool for research because of its open architecture that allows real-time control of stimulation parameters on eight stimulation channels in open or closed-loop control from a multitude of sensors connected to the stimulator's inputs. MotiMove is a portable, battery-powered stimulator that supports mechanical systems (e.g., FES cycling, rowing, and walking machines) and target-oriented movements such as reaching and grasping. The incorporated Bluetooth module allows a PC/tablet/smartphone communication for the remote setup of the stimulation mode and parameters and acquiring sensor data. Different powering options allow prolonged operation time. MotiMove has a CE mark for the Class IIa medical device and is already used in healthy and persons with paralysis.

Keywords: cycling, FES, rowing, stimulator

INTRODUCTION: Commercial portable functional electrical stimulators (FES) nowadays are dedicated devices for a single purpose application (e.g., cycling,[1,2] reaching and grasping,[3,4] drop-foot[5,6]). One exception is the RehaMove stimulator,[7] produced by Hasomed GmbH, Germany. Until recently, RehaMove was the only commercial stimulator that could support researchers in developing their stimulation and feedback control protocols by using PC processing power between the stimulator and sensors. Now, MotiMove[8] can provide the same support with more features. MotiMove functions were selected following a more than 40-year research of the group at the University of Belgrade. Therefore, MotiMove was made to support as much as possible the research and development requirements. MotiMove is already used in several end-user products for cycling and rowing. Several randomized controlled trials were conducted in applications in professional sports, muscle preservation, the elderly, and persons with motor disabilities. All of these tests resulted in a high level of satisfaction for the users. There are ongoing clinical studies in persons with spinal cord injury (SCI) rowing and post-Covid rehabilitation.

METHODS

Hardware: MotiMove has eight independent currents sources (8 channels) that can be controlled individually. The shapes of the currents at the output are compensated biphasic pulses with the exponential discharge (Figure 1). Each output stage (channel) has a serial capacitor of 220 pF to act as an additional safety measure to prevent any DC current leakage into the skin. In parallel with this capacitor is a solid-state relay that is activated (to bypass capacitor by short circuit) 450 μs before the positive pulse and turned off after a period that is four-time longer than the duration of pulse from the beginning of the pulse (blacked arrows in Figure 1).

Control is implemented using the STM32 family-based microcontroller with a 12-bit A/D convertor. A Bluetooth module allows communication with a PC/tablet/smartphone to receive the stimulation parameters and send information from sensors and stimulation states. There are three special purpose ports (Figure 2) for up to six analog inputs from sensors (AI), eight digital inputs/outputs, and 2xI2C busses (S1 and S2). One serial communication port (COMM) allows communication with a PC via an isolated USB converter. Alternatively, COMM port can connect to a second MotiMove stimulator for having at disposal 16 channels for stimulation.

The REMOTE port is used to connect the remote controller, allowing gradual stimulation intensity control (throttle), and includes a Start button and an emergency stop button (Figure 3).

MotiMove is powered from an internal Li-Ion battery of 7.4 V. Internal converter boosts the voltage reserve to