Wireless Distributed Architecture for Therapeutic Functional Electrical Stimulation : a technology to design network-based muscle control

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***Abstract*— This paper presents a distributed Functional Electrical Stimulation architecture based on a wireless network, for therapeutic training of disabled patients. On this distributed architecture, a global controller can pilot a set of stimulation and acquisition units and modify dynamically stimulation and acquisition parameters. This solution intend to be a tool for researchers and therapist to develop closed-loop control algorithms and strategies for therapeutic rehabilitation applications with external FES, in a clinical context. In a wireless network-based control, the variable delay introduced by the network must be taken into account to ensure the stability of the closed loop. Thus, in order to characterize the medium on which the control is performed, we carried out accurate measurements of the architecture performances (stack-crossing, round-trip time, etc.).**

I. INTRODUCTION

Transcutaneous (surface) electrical stimulation can generate an artificial contraction of skeletal muscles by applying sequences of electrical pulses to sensory-motor system via electrodes which can be placed on the skin [1]. This is a technique widely applied for physical therapy, sports training and clinical purposes. It can be used for muscle atrophy treatment, muscle force training, endurance training, pain treatment and functional movement therapy [2].

Functional Electrical Stimulation (FES) concerns the restoration of a functional movement in disabled patients. The contractions actuate joints by stimulating one or more muscles that exert torques about the joint. The resulting joint angle can be controlled by modulating the intensity of stimulation delivered to the flexor and extensor muscles, which actuate the joint in opposite directions. FES applied to limbs include foot drop correction, single joint control, cycling, standing up, walking, hand grasping enhancement... A wide range of disabilities are concerned with FES, they include spinal cord injuries, stroke, multiple sclerosis; cerebral palsy and Parkinson's disease [3], as well children as adults. Two distinct objectives may be targeted when using those techniques, depending on the type of disorder: chronic assistance or acute training. FES can be applied for example for walking assistance and training in post-stroke hemiplegic patients, as well as for standing and gait restoration in paraplegic patients [4]. The FES used in the framework of exercise was termed Functional Electrical Therapy (FET).

In this context of FET, stimulators which are used are wire-based and centralized ones. "Fig. 1" shows a classical centralized stimulator architecture. Some electrodes are used for stimulation or recording of the sensory-motor system activity (electromyogram, also called EMG) and each electrode is connected to a central controller by means of wires (the number of wires for each electrode depending on its number of poles). Those wires constitute an important constraint for the patient mobility, and thus for the training. Sometimes, those devices have sensors allowing them to acquire physical measurement (torque, angle), consequently they often become cumbersome and impractical to set up.

As it is mentioned in [5], most commonly used FES systems use either open-loop or finite-state control systems. Open-loop FES systems require continuous or repeated user input, while finite-state FES systems execute a preset stimulation sequence in an open-loop fashion when a specific condition is met. Although they are fluently used to correct foot drop, finite-state controllers typically do not adequate to adapt stimulation patterns to as well the frequent changes of walking speed as muscular fatigue. Commercial closed-loop FES cycling products are available; cycling cadence is kept constant by increasing the amount of stimulation delivered to the muscle when this latter begins to fatigue [5]. Nevertheless, this control remains limited for many applications of FES technology.

Many potential applications of the technologies of FES (balancing during standing, torso control during sitting, and

Fig. 1. Wire-based and centralized architecture of stimulation

walking), [5], for the deficient persons require a closed-loop and real-time control more sophisticated (modulation of the stimulation according to information of sensors) for a more effective FES as far as physiology is concerned. This would allow to compensate for the errors of model and the disturbances, and limit user intervention during FES-assisted tasks. Some works, about foot drop correction, showed the advantage to adjust dynamically the envelope of stimulation to be closer to the natural activity of the tibialis anterior: improvement of the dorsiflexion and the reduction of the electrical load received by the muscle, what could be a solution to decrease the muscular fatigue as well as the electric consumption of the systems [6]. One more time these works were based on cumbersome prototypes of research.

Furthermore, to treat a large number of functional deficiencies by the FES, it is necessary to be technically able to coordinate stimulation and acquisition on various distributed sites, more or less distant on the human body and implied in the current phase of the movement. It thus implies that the stimulators and the sensors are connectable directly or via a network.

However, the wire constraint must be removed to propose realistic solutions for rehabilitation clinical applications to be used by physiotherapist and/or the patient himself. Mobility, cumbersome and practicality are important criteria in FES technology design.

Like Loeb et al. suggested the use of distributed units through the BION system [7], to answer all these needs we mentioned, we designed and developed in partnership with the company Vivaltis[1], a new technological device for distributed FES architecture based on distributed stimulation units (DSU) [8]. The aim being to perform a network-based muscle control, we introduce a Medium Access Control (MAC) method [9] allowing a deterministic exploitation of the medium with groups of units, and favouring the reactivity of the distributed architecture unlike MAC strategies based on competition in Wi-Fi or ZigBee technologies. This deterministic MAC, allows to measure communication delays and define bounds, useful for network-based control.

II. METHODS

A. FES distributed architecture

In our distributed architecture of external FES, a global controller manages, via a wireless link of communication, a set of distributed and autonomous stimulation units (DSUs). It's the same for the distributed units of measure (DMUs), dedicated to the observation.

The concept is to decentralize the command of the muscle in closer of his activator, that is electrodes. In other words, the distributed units (DUs) performs the generation of the electric impulses and the sequence of these impulses of stimulation as well as the acquisition of signals locally.

The evolution of the concept of a centralized stimulation towards a distributed stimulation also requires that units are communicating as shown by "Fig. 2". The communication within the architecture is based on a 3-layer protocol stack in compliance with the reference given by the structure of reduced OSI model. These layers are the Application layer, the MAC layer and the Physical layer. The physical layer ensures the telecommunication over a wireless medium (used in considered hardware design). The MAC layer of this protocol stack ensures a deterministic medium sharing, that is to says that only one unit can speak over the network at a time (no collision) and response time of request acknowledgement is known and bounded. It allows the controller working with either a single (unicast) or a group of DUs (multicast) by using an individual or group identifier [9].

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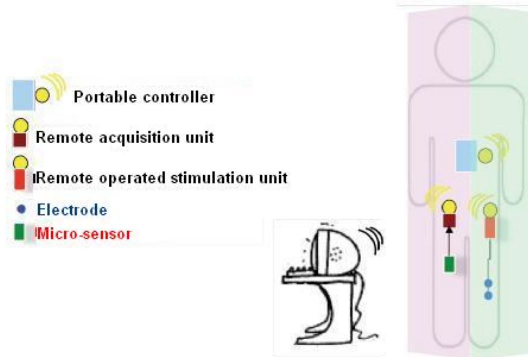

Fig. 2. Wireless-based and distributed architecture of stimulation

The application layer supports configuration, programming and remote operating [10](from start/stop requests to online control) of the remote unit.

So, those DUs have means of treatment and communication as well as a digital and analogue electronics parts for stimulation and/or acquisition. The controller of FES ensures the control of the application and control of the network of DUs. Therefore, our solution meets the needs of mobility (wireless link), coordination of the activity of several sites on the human body (network) and on-line control of stimulation parameters (frequency, amplitude and pulse width) or acquisition parameters (acquisition frequency, gain, range).

B. Hardware design

A new commercial stimulation device, "Fig. 3", based on the wireless architecture of external FES, was developed by Vivaltis company. The DUs was baptized "PODs®" by the manufacturer.

The controller board can be interfaced with a computer via a USB link what allows to get computing power for processing measurement, graphic interface, execution of a law of command. So, the practitioner can define an application of FES and parametrize it according to the patient. The controller can also embed a law of command and execute it. This controller communicates with the PODs® via a $IEEE802.15.4$ 2.4 GHz Radio Frequency (RF) link.

Each POD® consist of two electronic boards, "Fig. 3". A mother board embeds the communication protocol stack and ensures treatment of requests, and a dedicated daughter board executes the request (stimulation or acquisition). Currently, two types of daughter board have been developed : $Stim/Bio$ and $Universal$ boards. The $Stim/Bio$ board has two input/output channels to perform either stimulation or physiological signal acquisition (Electromyography) also called biofeedback. On the regulated-current-based stimulator, the maximal current is 100 mA with a load impedance of 1 kOhms. Besides stimulation frequency, amplitude, pulse width and electrical polarity are modifiable. The biofeedback system measures physiological signals from 10 Hz to 1 kHz and has three selectable measuring ranges ($80\mu V$, $200\mu V$ and $400\mu V$ maximum). The $Universal$ board is a hardware interface board between the mother board and another acquisition board. The acquisition board can be plug on the $Universal$ board and communicate with the mother board by a SPI[2] bus. The integration of new sensors and/or actuators is thus possible for future applications.

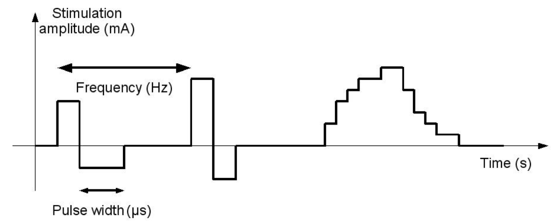

Fig. 4. Examples of generated pulse pattern

III. RESULTS

A. Stimulation and biofeedback sequence

On a $Stim/Bio$ POD®, a stimulation sequence is initiated by a configuration then a start request. The configuration defines:

- pulses frequency and pulse width,
- default current amplitude and the maximal limit (for security),
- pulse pattern (biphasic or monophasic square-wave, exponential ...), see ”Fig. 4”,
- modulation period of current amplitude,
- modulation period of pulses frequency and pulse width (wobble).

During the stimulation sequence, the frequency, the pulse width and the current amplitude can be set dynamically according to constraints defined.

In biofeedback, in order to detect the electromyogram (EMG) envelop, the sampled signal is filtered locally before being returned to the controller. The filter parameters are also set dynamically.

B. Measurements

In case of (wireless) network-based control, the variable delay introduced by the network must be taken into account to ensure the stability of the closed-loop [11], [12]. If the delay is bounded and its variation is predictable, there exists a stable controller. Our approach, which aims at guaranteeing the system stability under some limits, relies on a dynamical adaptation of the muscles control law parameters according to both measurement and prediction of communication delays variation. Thus, in order to characterize the medium on which the control will be performed and the network quality of service (QoS), we carried out precise measurements of the architecture performances (stack-crossing, round-trip time, frame losses rate, etc.). In this part, we focus on round-trip time (RTT) of controller request, frame losses and reception signal level.

There is two main causes of frames losses. On one hand, obstacles presence (human, liquid or metallic body) and increasing distance (beyond 10 meters) between the two communicating units cause important reduction reception signal level. As shown in ”Fig. 5”, lost frame rate is relatively low (4,2 %) even in case of low reception signal level. The frames losses seems to occur when the reception signal level is below a threshold of -90 dBm. In this experimental setup, controller exchanged with the POD®, 5000 frames of 100 Bytes, one every 50 ms. The POD® was placed on a lower limb of a person to 7 meters away from the controller. Each one was situated in two different rooms separated by a steel structure wall.

On the other hand, frames collisions can occur if other wireless technology standards using the 2.4 GHz Radio Frequency band, as Wi-Fi, Bluetooth or Zigbee, are in the same environment and on the same neighbourhood frequency

[2] $http://en.wikipedia.org/wiki/Serial_Peripheral_Interface_Bus$

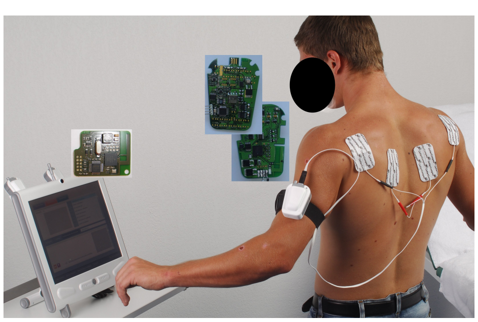

Fig. 3. Vivaltis commercial stimulation unit, also called PODs®

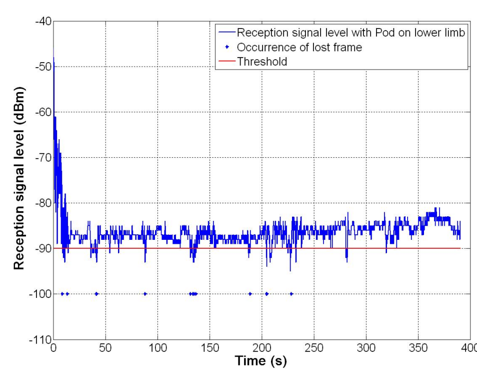

Fig. 5. Frames losses occurrence with a low reception signal level

channel than our communicating FES units. In this new experimental setup, laptop communicated via Wi-Fi on the $IEEE802.15.4$ channel 7, corresponding to the Wi-Fi standard channel 6. We carried out an exchange of 5000 frames between our communicating units over each $IEEE802.15.4$ channel and "Fig. 6" shows a peak of the frames losses rate on a the occupied channel by the Wi-Fi devices.

In order to perform network-based closed-loop control on this architecture by integrating communication and processing delays within the control loop, we carried out experiments for estimating the delay and its variation according to frame payloads and disturbances. Collected datas from the experimental setup gave a first estimation of communication performances (stack-crossing and transmission time) between entities of this distributed architecture. Under these experimental conditions, 100 requests were sent for each stimulation operation, both stack-crossing time and RTT (application request sending + acknowledgement) are relatively constant "Tab. I".

IV. CONCLUSION

In this paper, we presented a new technological device of FES which is today an exploitable tool for the researchers and the therapists to take up the challenges of the development of closed-loop control algorithms and strategies for therapeutic rehabilitation applications with external FES, in a clinical context.

The carried out analysis in term of performance of our FES wireless distributed architecture, gives a first representation of the quality of service. This will allow to verify the adequacy of our architecture performance with the aimed applications because the time delay between stimulation and the variable onset of a muscle contraction in addition to the processing and transmission delays must be taken into consideration for muscle control development [13], [5]. In the future, our work will be to design means and strategies of QoS supervision, in particular the variations of transmission delay and the lost frame rate on the wireless network (application request resending in a lost frame case). Then we will experiment a network-based multi-site closed-loop control on this architecture, according to the dynamic evolution of the QoS.

The aimed application is the muscle control based on a high order sliding mode controller (HOSM) under wireless based distributed FES architecture. The response of stimulated muscle is nonlinear and time varying due to fatigue, nonphysiological recruitment of muscle fibers, and changing muscle composition due to regular FES use. The required robustness regarding parameters variations and external disturbances lead us to adopt a controller relying on the sliding mode theory. We wish to adapt the HOSM control law applied to the FES-based control of the movement of the human knee joint by co-contraction of quadriceps and hamstring muscles [14].

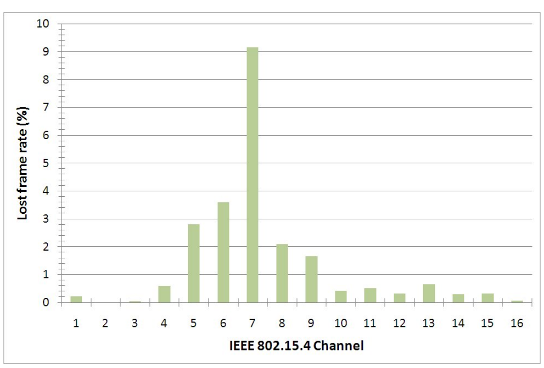

Fig. 6. Frames losses rate according to the IEEE802.15.4 standard channels

TABLE I

ROUND-TRIP TIME OF THE STIMULATION OPERATIONS

Operations	Mean RTT (ms)	Standard deviation (ms)
Configuration stimulation	12.19	0.016
Start stimulation	5.92	0.011
Set stimulation amplitude	6.678	0.018
Stop stimulation	5.999	0.009

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