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DEMAR is a common project with University of Montpellier 2, University of Montpellier 1 and CNRS.

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2. Overall Objectives

2.1. Introduction

Functional Electrical Stimulation (FES) has been used for about 30 years in order to restore movements. At the beginning, only surface stimulation was possible and thus only used in a clinical context due to the low reliability of electrode placements. In the early eighties, implanted FES appeared through well known applications (pacemaker, Brindley bowel control, cochlear implant, and more recently Deep Brain Stimulation). The complexity of the system for movement restoration is such that no commercial application really arise. Even though the original idea of FES is still the same, activating the moto-neurone axons with impulse current generator, the stimulus waveform and its parameters have drastically evolved and the electrode placements became various: epimysial stimulation at the muscle's motor point, neural stimulation on the nerve, Sacral roots stimulation near the spinal cord. These changes came from fundamental research, not yet achieved, in neurophysiology. This knowledge can efficiently be included in the next implanted neuroprosthetic devices allowing a wide variety of features. Moreover, currently, FES is the only way to restore motor function even though biological solutions are studied, because the research are not yet successfully tested on humans. Few teams carry out researches on implanted FES (http://www.ifess.org) and the functional results remain poor. Nevertheless, the technique has proved to be useable and needs enhancements that will be addressed by DEMAR. In particular, complex electrode geometries associated with complex stimulus waveforms provide a way to perform fibre type selectivity and spatial localisation of the stimuli in the nerves. These features are not yet implemented and demand new hardware and software architectures. Some teams in Denmark (SMI U. Aalborg), Germany (IBMT Franhaufer Institute), England (U. College of London), Belgium (U. Catholique de Louvain), United States (Cleveland FES centre), and Canada (Ecole Polytechnique de Montréal), work on multi-polar neural stimulation but mainly on the electrode aspect.
Such a complex system needs advanced control theory tools coupled with a deep understanding of the underlying neurophysiological processes. This major area of research will be also an important part of the DEMAR objectives. Very few teams (for instance ETH in Zurich, Switzerland) work on this topic because it needs a great amount of interactions between completely different disciplines such as neurophysiology, biomechanics, automatic control theory, and advanced signal processing. Besides, animal experiments performed in order to validate and identify models are particularly difficult to manage. Control schemes on such a complex non linear, under-actuated system, not completely observed and perturbed by the voluntary movements of the patient are quite difficult to study due to the lack of precise simulations platforms (for practical evaluation before experimentation) and the lack of theoretical results on such systems.
DEMAR is a joint project between INRIA, CNRS, Universities of Montpellier 1 and 2. DEMAR is located at LIRMM (joint CNRS and University laboratory working on Computer sciences, Micro electronics, and Robotics) in Montpellier. DEMAR works in close relationship with rehabilitation centres among them the Centre Bouffard Vercelli in Cerbère and Propara in Montpellier. International collaborations exist since 2003 with the Sensory Motor Interaction Lab at the University of Aalborg in Denmark (Professors Dejan Popovic, Ken Yoshida). DEMAR research interests are centered on the human sensory motor system, including muscles, sensory feedbacks, and neural motor networks. Indeed, DEMAR focuses on two global axes of research:

- Modeling and controlling the human sensory motor system.
- Interfacing artificial and natural parts through implanted neuroprosthetic devices.

The main applied research fields are then:

- Quantitative characterization of the human sensory motor system firstly for motor disorders diagnosis and objective quantification, and secondly in order to help the design of neuroprosthetic devices.
- Restoring motor and sensitive functions through implanted functional electrical stimulation (FES) and neural signals sensing.
- Improving surface stimulation for therapy (verticalization of paraplegic patients, reduction of tremor, reeducation of hemiplegic post-stroke patients...)

2.2. Highlights

We signed an agreement with Propara clinical center in Montpellier allowing us to perform experiments on humans, under ethical comitee approval, within their buildings. We have now a permanent office in the clinic.

3. Scientific Foundations

3.1. Modeling and controling the human sensory-motor system

Our global approach is based on the theoretical tools of the automatic control theory.

3.1.1. Modeling

Designing efficient control schemes and performing realistic simulations need for modeling. The scientific approach is to develop multi scale models based on the physiological microscopic reality up to a macroscopic behavior of the main parts of the sensory motor system: muscles, natural sensors and neural structures. We also aim at describing multi scale time models to determine impulse synchronized responses that occur in a reflex or with FES, up to a long term fatigue phenomenon. All these models have a control input that allows them to be linked as different blocks of the sensory motor system.
Besides, we have to deal with problems related to the identification protocols. Identification is then based on the observation of signals such as EMG, output forces, and movement kinematics, while medical imaging gives the geometrical parameters and mass distributions. The success of the identification process is highly sensitive to the quality of the experimental protocols on animals and humans.

3.1.2. Synthesis & simulation

Simulation platforms have been largely developed for biped systems, including advanced impact models (using non regular equation, work carried out in collaboration with BIPOP). Given that kinematics and dynamics are described using Denavit-Hartenberg parameters and the Lagrangian formulae, such tools can be used. Nevertheless, important differences rely on the actuators and their associated model. Thus, based on this platform, a new one can be developed including the complex muscle dynamics. In particular, muscle dynamics contain discontinuous switching modes (contraction - relaxation, extension - shortening), strong non linearities, length and shortening speed dependencies that imply complex numerical resolutions.
As regards synthesis, generating a useful and efficient movement means that criteria can be defined and evaluated through an accurate numeric simulation. Optimization methods are then used to process the data in order to obtain stimulation patterns for a given movement. Two problems occur, firstly the complexity of the models may provoke the failure of the optimization process, secondly the criteria that have to be optimized are not always known. For instance, we have to define what is a "normal" gait for a paraplegic patient under FES; are the global energy, the joint torques, the estimated fatigue for each muscle the appropriate criteria ?=0A

3.1.3. Closed loop control

Some tasks cannot be performed using open loop strategies. Keeping standing position with a balance control can be improved as regard the fatigue effect using ankle / knee / hip angle sensors feedback. Muscle's contraction is then controlled to ensure the minimum of fatigue with the maximum stability. Cycling, walking on long distance pathways, need some control to be achieved with a higher level of performance. Modeling and simulation will be used to design control strategies while theoretical studies of performances (robustness, stability, accuracy) will be carried out. The system is highly non linear and not completely observable. New problems arise so that new strategies have to be designed. Finally a compromise between complexity, efficiency, robustness, and easy usage of the system has to be found. Thus, the success of a control strategy design will be evaluated not only through its intrinsic performances but also regarding its ergonomic.

Advanced control strategy such as high order sliding modes for the low level control of the co-contraction will be studied because of its robustness towards model uncertainty. Trajectory free predictive control will be also investigated for a movement phase such as swing phase during gait, because the movement can be described as intuitive constraints such as the center of mass need not to fall. Finally high level hybrid approaches based on continuous control and event triggered commutation of strategies will be studied using a formal representation of the architecture.

3.2. Interfacing artificial and natural parts through neuroprosthetic devices

To overcome the limitations of the present FES centralized architecture, a new FES architecture was proposed according to the SENIS (Stimulation Electrique Neurale dIStribuée) concept: the distribution of i) the stimulation unit with its control near its activator, i.e. its associated neural electrode ii) the implanted sensor with its embedded signal processing.
FES will be thus performed by means of distributed small stimulation units which are driven by an implanted and/or external controller depending on its role (in terms of functionalitites it ensures) in charge of the coordination of stimulation sequences. Each stimulation unit (called DSU, Distributed Stimulation Unit) will be in charge of the execution of the stimulation pattern, applied to the muscle by means of a neural multipolar electrode. A DSU is composed of analogue and digital parts (§6.3.1.2).
The SENIS architecture therefore relies on a set of DSU which communicates with an implanted or an external controller. We therefore studied the communication architecture and defined an adequate protocol, assuming firstly that the communication should be performed on a wireless medium and secondly that this architecture can also contain distributed measurement units (DMU for sensors).
The external supervisory controller will probably be designed and implemented according to a software component based approach, like that we developed for instance for robot controllers

3.2.1. Stimulators

We mainly focus on implanted devices interfaced with neural structures. Both the knowledge about how to accurately activate neural structures (neurophysiology), and technology including both electrode manufacturing and micro electronics will be studied. Complex electrode geometries, complex stimulus waveforms, and the multiplicity of the implantation sites are the subjects we deal with in order to obtain a selective, progressive and flexible activation of neural structures. Our theoretical approaches are based on:

- Design and test in micro electronics with ASIC developments.
- Formal Petri Nets representation and automatic model translation based implementation.
- 3D electrostatic theory to model interactions between electrodes and neural structures.
- Electrophysiology modeling such as Hodgkin-Huxley model.

3.2.2. Sensors

The development of a closed-loop controller implies the use of sensors whose choice and number are highly constrained by practical, psychological and cosmetic considerations: the stimulation system has to be implanted in order to simplify its use by the patient; it is therefore not possible to cover the person with various external apparatuses. An alternative to artificial sensors is the use of natural sensors already present, which are intact and active below the lesion in the spinal cord of the injured patients. DEMAR is then interested in implanted sensors in order to design complete implanted solutions (stimulation and sensing). As regards sensing, two kinds of sensors will be studied:

- Physical sensors such as micro attitude centrales.
- Natural sensors that means interfacing with afferent nerves and ENG recordings. The same theoretical tools and technology as for implanted stimulators could be used.

In both cases, advanced signal processing applied to biosignals is needed to extract relevant pieces of information.

3.2.3. Patient interface

The patient interacts with the system in three ways:

- He decides which movement he wants to achieve and informs the system.
- He performs voluntary movements in a cooperative way, to turn right or left for instance, but he could also disturb the system when a closed loop control is running.
- Passive actions like arm supports through the walker for the paraplegic patient are used to control balance and posture.

It's not trivial to integrate all these events in the system. This field of research can learn from tele-operation and Human Machine Interfaces research fields. The patient needs also to get pieces of information of the current state of the system. Sensory feedback have to be implemented in the system such as screen, sound, tactile vibrations, electrical stimulation, etc... Choosing meaningful pieces of information such as heel contact, and the way to encode it, will be addressed.

3.2.4. Supervision & networking

Activating the system through stimulators, sensors, and analyzing patient behaviors need multiple devices that communicate and demand energy. Interfacing natural and artificial parts imply to address problems such as networking, data transfer, energy storage and transfer through wireless links. On such a complex system, supervision is necessary to ensure security at the different involved levels. Fault tolerance and reflex behavior of the system will be studied to improve system reliability particularly when the patient uses it at home without any medical person support. The theoretical approach is based on Petri Nets to design and then analyse the behavior of the entire distributed system.

4. Application Domains

4.1. Objective quantification and understanding of movement disorders

Modeling based on a physical description of the system lets appear meaningful parameters that, when identified on a person, give objective and quantitative data that characterize the system. Thus, they can be used for diagnosis.
Modeling provides a way to simulate movements for a given patient so that through an identification process it becomes possible to analyse and then understand his pathology. But to describe complex pathology such as spasticity that appears on paraplegic patients, you need not only to model the biomechanics parts - including muscles -, but also parts of the peripheral nervous system - including natural sensors - to assess reflex problems. One important application is then to explore deficiencies globally due to both muscles and peripheral neural nets disorders.

4.2. Palliative solutions for movement deficiencies

Functional electrical stimulation is one possibility to restore or control motor functions in an evolutive and reversible way. Pacemaker, Cochlear implants, Deep Brain Stimulation are successful examples. DEMAR focuses on movement disorder restoration in paraplegic and quadriplegic patients, enhancements in hemiplegic patients, and some other motor disorders such as bladder and bowel control. Nevertheless, since some advances in neuroprosthetic devices can be exploited for next generation of cochlear implants, the team also contributes to technological and scientific improvements in this domain.

The possibility to interface the sensory motor system, both activating neural structure with implanted FES, and sensing through implanted neural signal recordings open a wide application area:

- Restoring motor function such as grasping for quadriplegic patient, standing and walking for paraplegic patient, foot drop for hemiplegic patients. These applications can be firstly used in a clinical environment to provide to physiotherapist a new efficient FES based therapy (using mainly surface electrodes) in the rehabilitation process. Secondly, with a more sophisticated technology such as implanted neuroprostheses, systems can be used at home by the patient himself without a clinical staff.
- Modulating motor function such as tremors in Parkinsonian patient using DBS (Deep Brain Stimulation). Techniques are very similar but for the moment, modeling is not achieved because it implies the central nervous system modeling in which we are not implied.
- Sensing the afferent pathways such as muscle's spindles, will be used to provide a closed loop control of FES through natural sensing and then a complete implanted solution. Sensing the neural system is a necessity in some complex motor controls such as the bladder control. Indeed, antagonist muscle's contractions, and sensory feedbacks interfere with FES when applied directly on the sacral root nerve concerned. Thus, enhanced activation waveforms and sensing feedback or feedforward signals are needed to perform a highly selective stimulation.

In any case, experimentations on animals and humans are necessary so that this research needs a long time to go from theoretical results to applications. This process is a key issue in biomedical research, it needs: i) design of complex experimental setups both for animals and humans, ii) ethical attitude both for humans and animals, with ethical committee approval for human experiments iii) volunteers and selected, both disabled and healthy, persons to perform experiments with the adequate medical staff.

5. Software

5.1. Software

5.1.1. RdP to VHDL tool

Participants: David Andreu, Grégory Angles.

To go further in the modularity and the reusability of sub-parts of complex hardware systems, we defined HILECOP components. An HILECOP component has : a Petri net-based behavior, a set of functions whose execution is controlled by the PN (Petri Net), and a set of variables and signals. Its interface contains places and transitions from which its PN model can be inter-connected as well as signals it exports or imports. The interconnection of those components, from a behavioral point out view, consists in the interconnection of places and/or transitions according to well-defined mechanisms : interconnection by means of oriented arcs or by means of the "merging" operator (existing for both places and transitions). We started, through an INRIA ODL (Opération de Développement Logiciel), the development of an Eclipse-based version of HILECOP with the aim at making it accessible to the academic community.

5.1.2. SENISManager

Participants: David Andreu, Grégory Angles, Robin Passama.

We developed a specific software environment called SENISManager allowing to remotely manage and control a network of DSUs, i.e. the distributed FES architecture. SENISManager performs self-detection of the architecture being deployed (Fig. 1; left). This environment also allows for manipulating micro-programs from their edition to their remote control (Fig. 1; right).

5.1.3. STIMWare

Participants: David Andreu, Robin Passama.

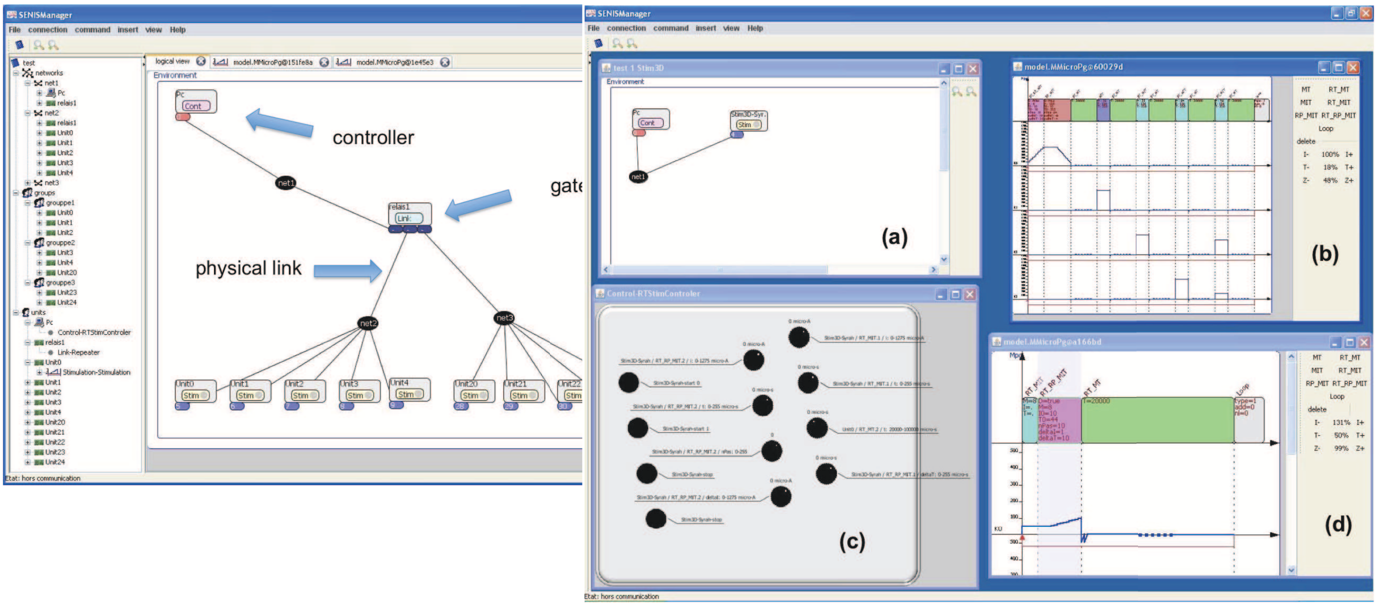

Figure 1. Left) Example of SENIS Architecture managed through SENIS Manager. Right) Some windows available on SENISManager: (a) FES architecture management, (b) graphical editing of micro-programs, (c) console for remote control of the execution of micro-programs of which parameters' values are displayed in real-time (d).

We designed and partially developed a software environment allowing the management and control of a heterogeneous technology based external FES architecture. This software environment eases the configuration and exploitation of the external FES platform since it ensures the interoperability of the heterogeneous entities implied within the platform. It is based on a middleware and a set of modules organized according to two-layer software architecture: the interaction layer and the control layer. The interaction layer directly pilots stimulators and sensors used in the platform, ensuring the communication with these entities according to their specific protocol-stacks. Its middleware contains a scheduler in charge of the scheduling, the activation and the monitoring of the corresponding modules. The control layer supports the development of control strategies, potentially based on a set of heterogeneous entities (stimulators and sensors), like closed-loop controllers and/or supervisory controllers. This software is already tested woth stimulators used on patient under ethical comitee approval.

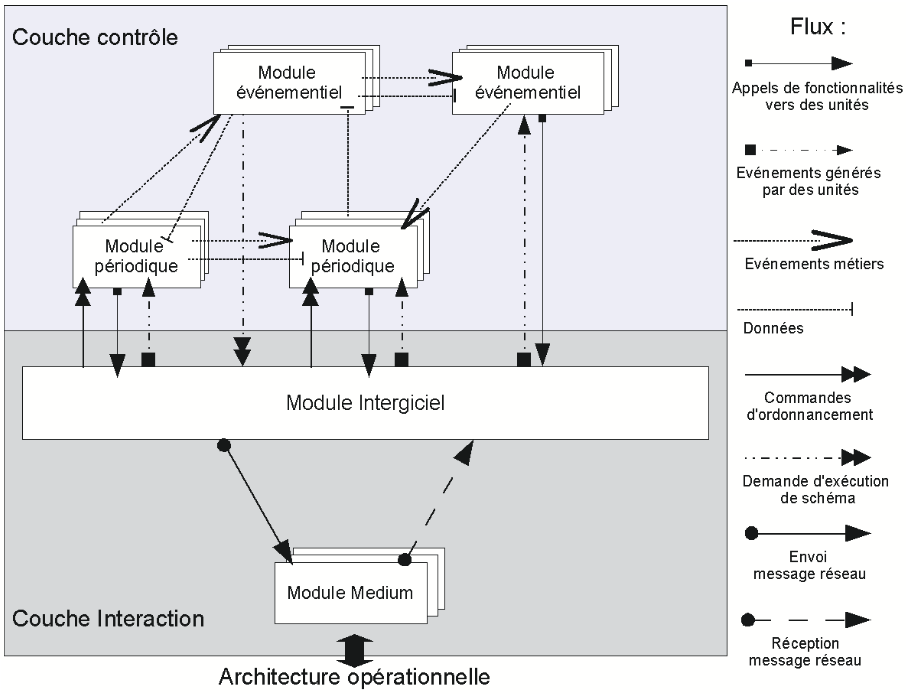

Figure 2. Schematic description of the software environment allowing the deployment of control strategies based on heterogeneous entities

A graphical interface will allow the end-user to manipulate the FES architecture (software entities and their associated hardware), at greater abstraction level.

5.1.4. Simulating the human motion under Functional Electrical Stimulation using the HuMAnS toolbox

Participants: Martine Eckert, Mitsuhiro Hayashibe, David Guiraud, Pierre-Brice Wieber (BIPOP), Philippe Fraisse.

Mathematical models of the skeletal muscle can support the development of neuroprotheses to restore functional movements in individuals with motor deficiencies by the mean of Functional Electrical Stimulation (FES). Since many years, numerous skeletal muscle models have been proposed to express the relationship between muscle activation and generated force. DEMAR model integrates the Hill model and the physiological one based on Huxley work allowing the muscle activation under FES. We propose an improvement of this model by modifying the activation part. These improvements are highlighted through the HuMAnS toolbox using a 3D biomechanical model of human named Human 36 which has 36 DOF. [20] describes this toolbox and the software implementation of the model. Then, we introduce the simulation results of the knee joint actuated by the muscle group (Quadriceps/Harmstring) using FES.

6. New Results

6.1. Modeling and identification

6.1.1. EMG-to-force estimation with physiology based muscle model

Participants: Mitsuhiro Hayashibe, David Guiraud, Philippe Poignet.

EMG-based muscle model for voluntary muscle contraction has a number of applications in human-machine interaction, sports science, and rehabilitation [22] . EMG-based model can account for a subject's individual activation patterns to estimate muscle force. For this purpose, so-called Hill-type model has been used in most of the cases. It already has shown its promising performance, but it is still known as a phenomenological model for microscopic scale in muscle physiology. We have already developed the physiological based muscle model for the use of functional electrical stimulation (FES) which can render the myoelectrical property also in microscopic scale. In this work, we develop EMG-to-force estimation framework based on this full physiological based muscle model in voluntary contraction. In addition to Hill macroscopic representation, a microscopic physiology originally designed by Huxley is integrated. It has significant meaning to realize the same kind of EMG-to-force estimation with a physiological based model not with a phenomenological model, because it brings the understanding of the internal biophysical dynamics and new insights about neuromuscular activations. Using same EMG data set of isometric muscle contraction, the force estimation results are shown by classical approach and new physiological based approach.

Subjects were seated on a chair with their right foot fixed on a Biodex dynamometer (Biodex Medical Systems, Inc., New York, USA). The torque around ankle joint was measured when it is voluntarily generated for the extension position. For EMG measurements, bipolar surface Ag/AgCl-electrodes were placed on the muscle belly of the medial Gastrocnemius (GAS) and Soleus (SOL). Synchronous acquisition of the force and differential EMG signal was done with 2kHz.

Here, isometric moment was estimated only from EMG signals by the classical Hill approach and the proposed physiological based muscle model. The predicted torque was compared with the directly measured torque around the ankle by Biodex system. For this preliminary trial, we make the comparison for the normalized torque against the one of maximum voluntary contraction (MVC). Finally, normalized estimated torque by classical Hill approach was obtained as shown in Fig. 3. The normalized data of measured torque (red), normalized estimated torque of SOL (green), normalized estimated torque of GAS (magenta) and the sum of two muscles (blue) are plotted.

For the new model, the rectified EMG was low-pass filtered with 30Hz cut-off frequency. Then, chemical input $u(t)$ was created by thresholding the extracted EMG signals as shown in Fig. 4. The thresholding can be assumed as muscle cell's all-or-nothing response to Action Potential (AP). The generated input command $u(t)$ was given to the contractile element of physiological model and the active stiffness k_c and the muscle force F_c were computed. Finally, normalized estimated torque by physiological muscle model could be obtained as shown in Fig. 5. In order to confirm the estimation ability both for short-term contraction and long-term contraction, the result which includes two type contractions is presented here. We have developed a method that allows to estimate the muscle force from EMG signal with physiology based model which has the link to underlying microscopic filament dynamics. The proposed method features:

- a novel physiologically detailed model for EMG-to-force estimation instead of a phenomenological Hill-type muscle model,

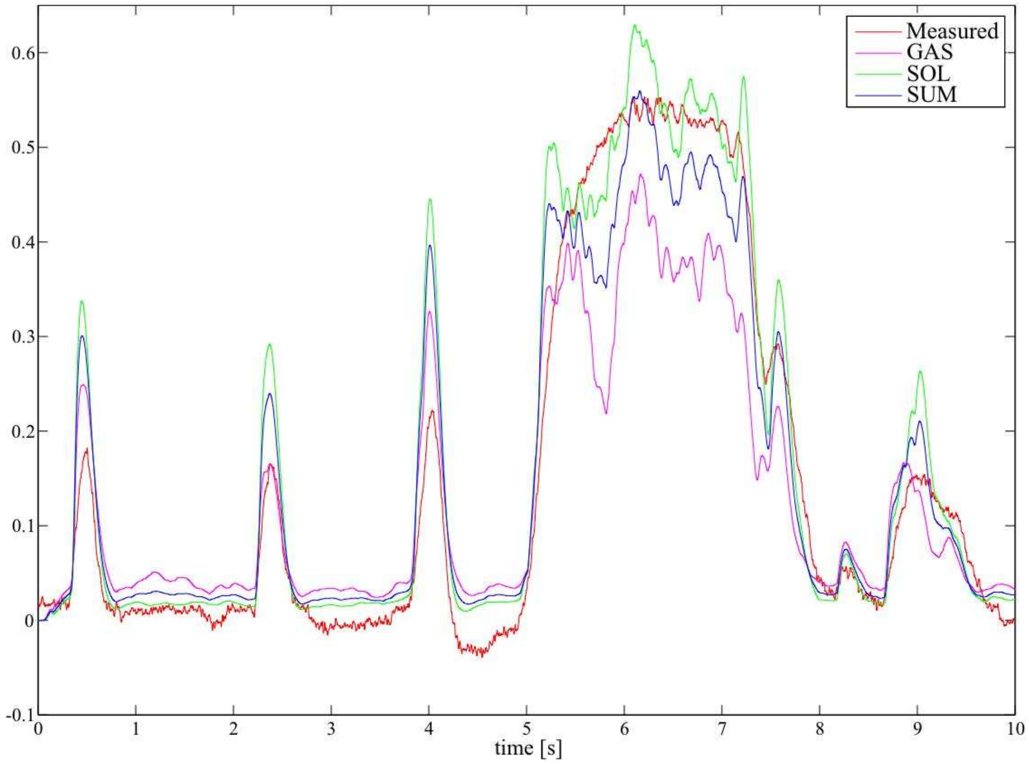

Figure 3. Normalized estimated torques by classical Hill approach and measured torque (red:measured, magenta:GAS, green:SOL, blue:SUM).

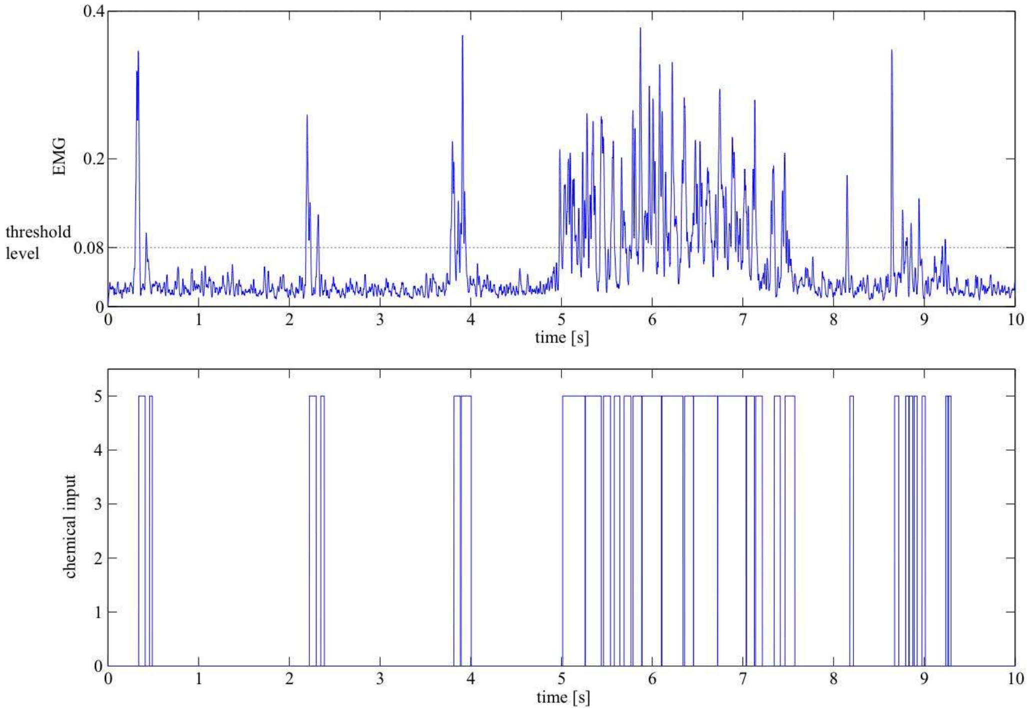

Figure 4. Generation of chemical input. Top: filtered rectified EMG signal, bottom: generated chemical input by thresholding.

- the estimation improvement both for short-term and long-term contraction,
- the consideration of firing rates of motor units in the generation of chemical command input.

Future work will focus on increasing the number of tests and the further interpretation of neuromuscular system both in voluntary and artificial activation in FES.

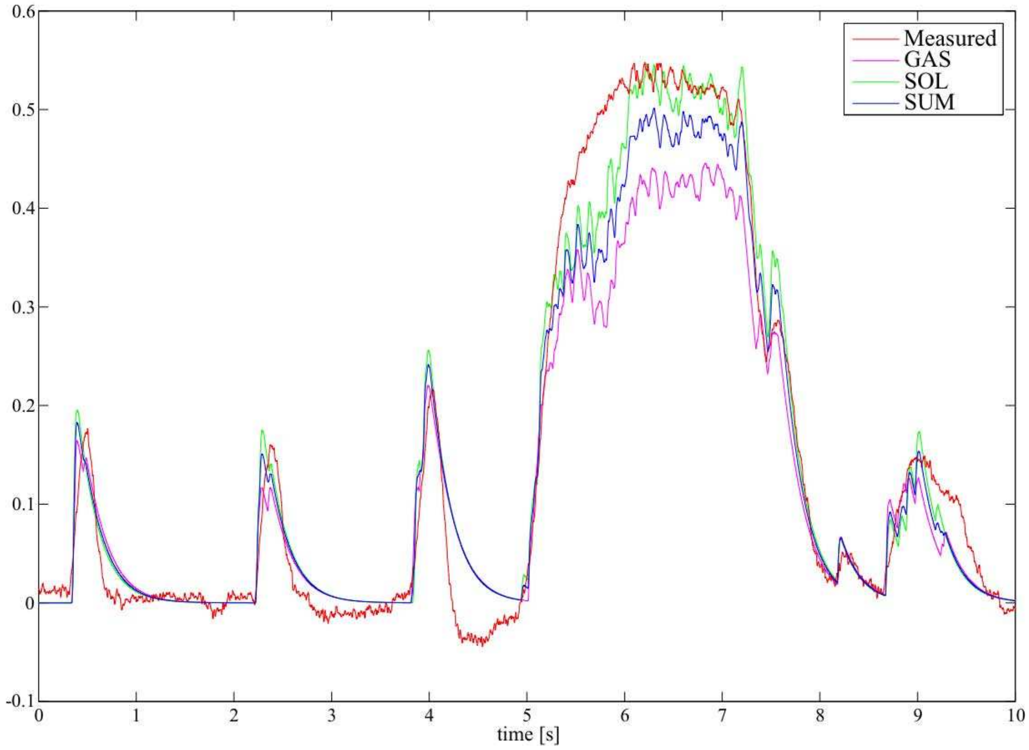

Figure 5. Normalized estimated torque by physiological muscle model and measured torque (red:measured, magenta:GAS, green:SOL, blue:SUM).

6.1.2. Activation modeling of motor response and H-reflex in FES

Participants: Mitsuhiro Hayashibe, David Guiraud, Floor Campfens.

In [17] a model which can improve the estimation of muscle activation under electrical stimulation, is presented. The model includes not only the motor recruitment but also the H-reflex recruitment, which is responsible for a larger exerted force than the estimation only based on motor recruitment at the lower stimulation levels. In the model, it is assumed that recruitment curves are sigmoid shapes, and there is no difference between motor units and no specific recruitment order. An accurate recruitment function is needed in the activation block of a muscle model. At this point, one important phenomenon is not taken into account in the estimation of motor unit recruitment under FES: the presence of the Hoffmann reflex (H-reflex). The H-reflex is present especially at the lower stimulation intensities and produces higher force than what would be expected based on the motor response. It makes the force estimation at low intensities of stimulation inaccurate. This H-reflex also prevents from accurately identifying the recruitment curves. This work proposes a model of motor unit recruitment which takes into account the H-reflex and uses the EMG signal for the identification of both contributions: the H-reflex and the direct motor activation.

The descending part of the H-reflex recruitment curve is complicated. The simplest explanation for the extinction of the H-reflex is that the reflexive activity is blocked by collisions with the antidromic action potentials induced in the motor neurons by the stimulation. This means that a function to describe the reflexive activation of motor units also requires the motor recruitment as an input.

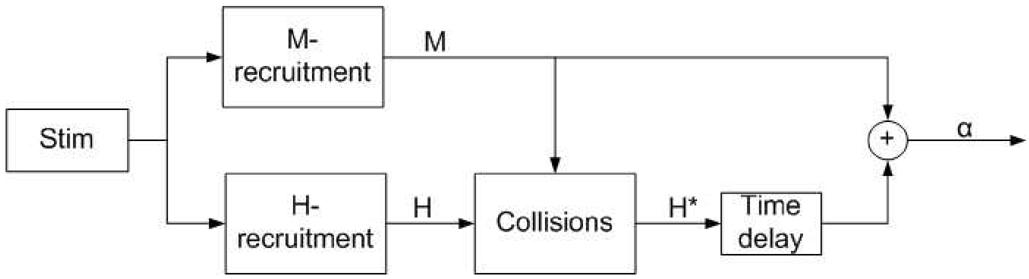

Figure 6. Structure of the activation model with H-reflex.

The form of the model in which the motor recruitment and H-reflex recruitment can be considered is given in Fig. 6. From the stimulation parameters (amplitude or pulse width), the motor response (M) and the H-reflex recruitment (H) are calculated. The portion of H which actually results in muscle contraction is calculated as H*. H* reaches the muscle slightly later than the direct motor response, so a time delay is added. Finally, the both pathways are added to give the muscle activation α. The expected overlap between the motor recruitment and the H-reflex recruitment (COL) is given by: $COL = H \cdot M$. The number of motor units activated in the H-reflex (H*) is then given by: $H^* = H - COL = H(1 - M)$. The formulas to calculate the M and H* can be written using a sigmoid function as below:

$$\begin{aligned} M &= \frac{1}{1 + exp\left[cm_2 \cdot (cm_3 - s)\right]} \\ H^* &= \frac{ch_1}{1 + exp\left[ch_2 \cdot (ch_3 - s)\right]} \cdot (1 - M) \end{aligned} \tag{1}$$

where s is the stimulation input (amplitude or pulse width) and ch_1, ch_2 and ch_3 constants alter the plateau, steepness of the sigmoid and the stimulation needed for 50% recruitment respectively.

Recruitment curves were determined for two subjects to test the proposed activation model. Transcutaneous stimulation was applied to the tibial nerve by bipolar electrodes. A cathode electrode was placed in the popliteal fossa, a $50cm^2$ anode electrode was placed on the knee at the patella. Stimulation was delivered in the form of rectangular monophasic pulses with durations of $200\mu s$ by a constant current stimulator (DS7AH, Digitimer Ltd., Hertfordshire, UK). For EMG measurements bipolar surface Ag/AgCl-electrodes were placed on the muscle belly of the medial gastrocnemius.

From the EMG data, the amplitudes of both M-waves and H-waves were determined as the peak to peak values. All amplitudes were normalized to the maximal M-wave amplitude. Stimulation amplitude was normalized to the minimum stimulation amplitude needed to induce a maximal M-wave. The recruitment functions for M and H* are fitted to the measured M-wave and H-wave amplitudes. As can be seen in the plot, the fitted function follows the measured data closely. This is also visible in the high R^2 values: The average R^2 value for the 4 fitted recruitment curves (motor recruitment and H-reflex recruitment for each subject) is 0.9731. For the motor response recruitment curves the average R^2 is 0.9916. For the H-reflex recruitment curves the average R^2 is 0.9545. It can be concluded that the proposed activation model can be used to give a better estimation of the muscle activation induced by electrical stimulation. Future work includes the testing of the proposed

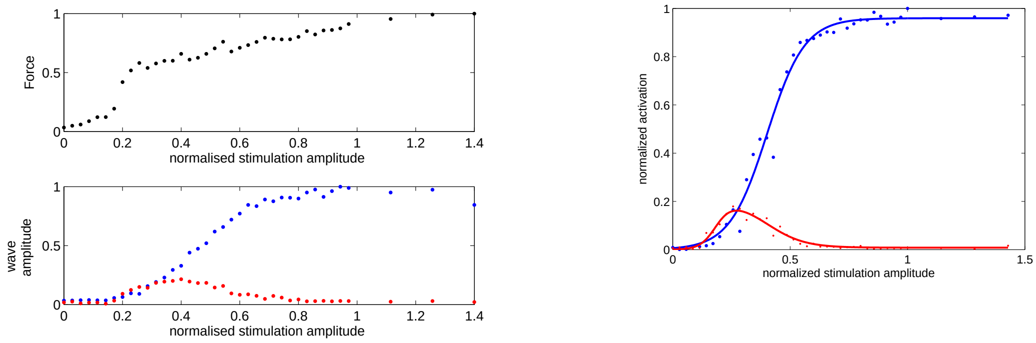

Figure 7. Left) Normalized data of one of the subjects: peak force and wave amplitude. In the lower graph the red dots represent the H-wave. The peak force shows a strong increase just above the motor threshold. Rigth) Recruitment curves fitted to the measured data of one of the subjects.

activation model in combination with the muscle model and comparing the force predictions with the new and the old activation model.

6.1.3. Bladder function modeling

Participants: Jérémy Laforet, David Guiraud, Christine Azevedo-Coste, David Andreu, Maureen Clerc (ODYSSEE).

We present here a major update of our smooth muscle model [24][23]. It aims at enabling the simulation of smooth muscle contraction under functional electrical stimulation (FES). The main addition is a model of calcium dynamics at the smooth muscle cell level. It links the calcium concentration to the electrical potential of the membrane cell, therefore to the neural stimulation. This concentration is used to control the microscopic mechanics of the muscle, leading to contraction. We also refined the mechanical model. As smooth muscle have way slower dynamics than striated ones, the second order terms can be neglected. We chose to simulate a well known example : the behavior of the bladder under the stimulation of a Finetech / Brindley implant. This way we can compare our qualitative results with the experimental data available in the literature. They show good consistency both in shape and time course.

6.1.3.1. Experimental validation

We also developped a setup for animal *in-vivo* experiments which should allow us to perform the identification of the model parameters. We have designed an experimental setup in order to elicit artificial contractions of the detrusor and record intravessical pressure, the goal being to compare recorded data with the model output. Acute experiments were conducted on 2 anaesthetized New Zealand white rabbits. The experimental data gathered are compared to the predictions of the model. In figure 9 we present preliminary experimental results. Stimulation parameters were: frequency=5Hz, amplitude=5mA and Pulse-width=500μs. We can see a quick rise of intravesical pressure on the onset of stimulation and its much slower return to baseline when stimulation is over. The rapid change around 40 seconds was due to bowel movements. They also could be use to identify its parameters using extended Kalman filtering methods.

To obtain quantitative simulation, we need to perform the identification of the model's parameters. A simpler mechanical model helps here, leading to fewer parameters to identify. We now try to identify our bladder model on rabbits before carrying out experiments on humans. Then, FES implant may be optimised and accurately tuned to perform micturition with the best low pressure / high outflow ratio.

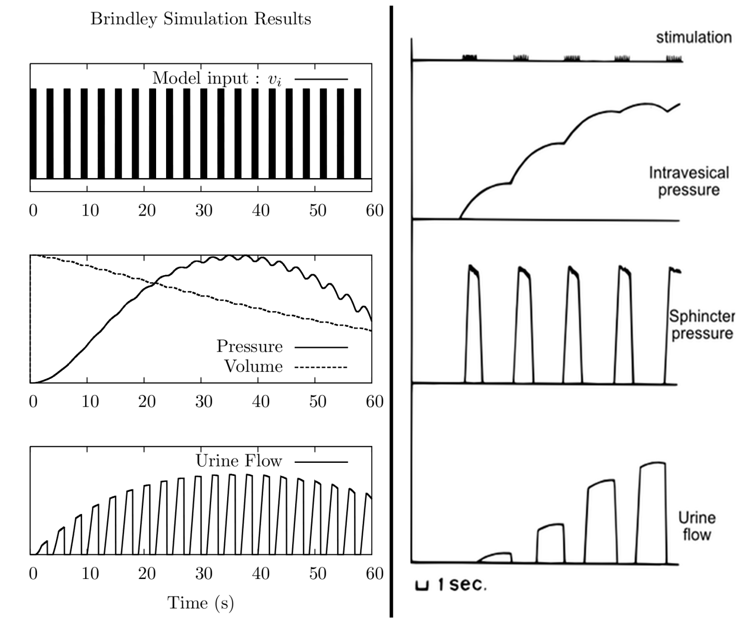

Figure 8. Simulation results vs Brindley's experimental data

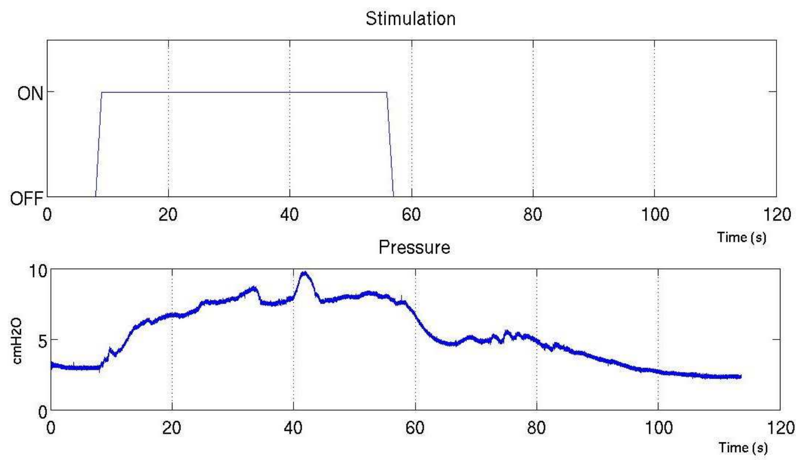

Figure 9. Experimental data: Intravesical pressure response to stimulation

6.1.3.2. Modeling selective stimulation of nerves

We also worked on selective stimulation of nerves using multipolar cuff electrodes (3 rings of 4 contacts each). It is a joint work with project-team Odyssee [18]. Using a specially modified version of the OpenMEEG software it is possible to compute the 3D electrical potential field inside the electrode based on the injected current. Selecting a nerve fiber inside the nerve it is possible to know the applied potential and thus to compute its effect on the fiber. In this aim we developed a model of the nerve fiber using Neuron software. It computes the potential in the fiber given the stimulation potentials. It allows to see if an action potential is generated and how it propagates in the fiber. Using these models we plan to test different approaches to activate fibers selected by their location and/or diameter. This will also be validated on animal experiment. They are planned for the begging of next year.

We focus on restoring bladder control, the function for which there is one of the highest demand from patients suffering from lesion of the central nervous system. For optimal bladder control, one must stimulate, within the same nerve, two muscles whose characteristics and actions are very different (the Detrusor which contracts the bladder is a smooth muscle, and the external sphincter which closes the output contains a striated muscle), and hence, one must perform selective recruitment of the two different types of nerve fibers.

6.1.4. Fatigue studies

Participants: Maria Papaiordanidou, David Guiraud, Alain Varray (University of Montpellier 1).

6.1.4.1. Time course of neuromuscular changes during low-frequency electrical stimulation

The purpose of the study [28] was to examine the time course of neuromuscular fatigue during a low-frequency electrostimulation (ES) session. Methods: Three 17-train stimulation bouts at 30 Hz (duty cycle 40%) were used to electrically induce fatigue in the plantar flexor muscles. Before and after every 17-train bout, torque, electromyographic activity [expressed by Root Mean Square (RMS) and Median Frequency (MF) values] and evoked potentials (M-wave and H-reflex) were recorded, while the twitch interpolation technique was used to assess the level of voluntary activation (LOA). Results: Torque during maximal voluntary contraction decreased significantly from the very first stimulation bout [-6.6 (4.84)%, P<0.001] and throughout the session [-10.32 (7.31)% and -11.53 (5.53)%, P<0.001, for the second and third bouts, respectively]. The LOA and RMS/M values were significantly decreased during the ES session [-2.9 (4.81)%, P<0.05, and -17.5 (26.05)%, P<0.001, respectively, at the end of the protocol], while MF showed no changes. The Hmax/Mmax ratio and M-wave characteristics were not significantly modified during the session. All twitch parameters were significantly potentiated after the first bout and this potentiation was present until the end of the protocol (P<0.001). Conclusion: The maximal torque decrease, evident from the early phase of a low-frequency ES session, appeared without a concomitant inhibition of motoneuron excitability or depression of muscle contractile properties. This is consistent with an early failure of the central drive to the muscle. A possible explanation could be the increased activation of type III and IV afferent fibers, leading to an impairment of the descending command.

6.1.4.2. Neuromuscular fatigue during triceps surae low-frequency electrical stimulation in subjects with different force generating capacity

Fatigue induced by neuromuscular electrical stimulation (NMES) is poorly understood. Although recent research gives evidence for the implication of different sites along the pathway of force production, the muscle factors susceptible to influence the response to electrically induced fatigue are still unknown. The purpose of the study [27] was to identify the time course of neuromuscular changes during NMES according to the muscleÕs capacity to generate force. Twelve healthy subjects with different force generating capacities [7 Strong (S) with mean torque during Maximal Voluntary Contraction (MVC) $88.8 \pm 1.6Nm$ and 5 Weak (W) with mean torque $64.4 \pm 3.2Nm$], participated in an electrostimulation protocol for the triceps surae, composed of 3 series of 17 stimulation trains (4s ON - 6s OFF, pulse duration 450?s, frequency 30Hz, at maximal tolerated intensity). Neuromuscular tests were performed before, during and immediately after the protocol. Torque and EMG activity of the gastrocnemius medialis muscle were continuously recorded. Alterations in muscleÕs characteristics (excitability and contractile properties) were evaluated by analysis of the muscle compound action potential (M-wave) and twitch torque. Motoneuron excitability was assessed by the H reflex, expressed in absolute value and normalized to M-wave maximal amplitude (respectively Hmax and H/M). Changes in the central command were assessed by using the twitch interpolation technique and the root mean square (RMS and RMS/M) obtained during MVC. MVC significantly decreased from the first 17-train bout and throughout the protocol for both groups (from $88.8 \pm 1.6Nm$ vs $64.4 \pm 3.2Nm$ at pre to $78.8 \pm 3.3Nm$ vs 58.2 ± 2.7 at post51, for S and W respectively), giving evidence of precocious neuromuscular impairments. Motoneuron excitability was not affected (no change of Hmax and H/M). Muscle contractile properties were significantly potentiated at post17 and for the rest of the protocol ($42 \pm 14\%$ for S vs $37 \pm 6\%$ for W for Pt values at post51) for both groups. Muscle excitability was significantly altered only in S, as proved by the significant decrease in M-wave amplitude and muscle response to trains of stimulation (-2.9 ± 3.7% and $-13.4 \pm 5.9\%$ for post51 respectively). Level of voluntary activation assessed by the twitch interpolation technique was lower for W and, although RMS/M was significantly decreased for both groups, the decrease in W was significantly more pronounced ($-21.8 \pm 4.5\%$ vs $-14.5 \pm 6.2\%$ for S). Neuromuscular fatigue can be attributed to both central and peripheral mechanisms for the S group, while for W it appears that mainly central mechanisms are involved. These observations should be taken under consideration when seeking to optimise training strategies for people with neurological disorders. The results show that improving muscle strength does not necessarily delay neuromuscular fatigue, but does change its nature.

6.1.5. Towards a model of the auditory system

Participants: Christophe Michel, Jérome Bourien (INM, INSERM CHU Montpellier), Christine Azevedo Coste.

Hearing is a complex sensorial process. Auditory stimulation devices turn sounds into nervous messages directly interpreted by the brain. In order to improve stimulation performances and stimulator calibration it would be of great interest to have a relevant model. The primary objective we were interested in was to study and implement the computational model proposed by Meddis (Sumner et al. 2003a). This model integrates all the hearing process mechanisms, from the reception of a sound to the generation of action potential trains in one or more of the auditory nerve fibers. The model is composed of: i) a band pass filter that models the speed of vibration of the stapes, ii) a linear resonator in parallel with a non-linear resonator model the movement of the basilar membrane, iii) a system of differential equations that model the release by the inner hair cells (see Fig. 1), neurotransmitter (NT) endogenous (glutamate) in the synaptic cleft (Fig.1). We have evaluated this model by comparing the data it generates with those acquired in vivo in guinea pigs by the 'Inner ear team' of the INM. From these results, we have proposed improvements to the initial model according to recent data from literature (Ruel et al., 2008).

Meddis model allows to reproduce faithfully: i) the behaviour of the basilar membrane, ii) the cycle of NT glutamate release, iii) the response of one given auditory nerve fiber to a stimulus. The model gives low importance to the role of auditory neurons. We introduced a model of integrator with threshold neuron type model at the output of the existing Meddis model and show that the obtained output is closer to the in vivo recorded data. This new model can also be used to simulate the functioning of the auditory system device in pathological conditions. As an illustration, we simulated the involvement of another type of glutamate receptor (present at the synapse but inactive in normal condition) and the result is consistent with recorded data (Ruel et al. 2008).

Using a computational model of the auditory system device adapted from that of Meddis offers the possibility of gaining knowledge about normal functioning and simulating pathological conditions. This work is ongoing in the form of a thesis CIFRE funded by the company MXM-Neurelec.

6.2. Movement synthesis & control

6.2.1. Correction of drop-foot

Participants: Christine Azevedo Coste, Roger Pissard-Gibollet (SED INRIA), David Andreu, Bernard Espiau (INRIA RA), Jérôme Froger (Rehab. Centre, Grau du Roi, CHU Nîmes), Rodolphe Héliot (Berkeley Univ.).

Hemiplegia is a condition where one side of the body is paretic or paralyzed; it is usually the consequence of a cerebro-vascular accident. One of the main consequences of hemiplegia is the drop-foot syndrome. Due to lack of controllability of muscles involved in flexing the ankle and toes, the foot drops downward and impede the normal walking motion. Today, there are commercially available assistive systems (e.g., Odstock stimulator) that use surface electrodes to stimulate Tibialis Anterior (TA) muscle and prevent drop-foot. The efficiency of drop-foot stimulators depends on the timing of stimulation and functionality of dorsiflexion motion. Classically, available stimulators use footswitches to detect foot on/off events. Theses discrete events allow only to play with the duration of the stimulation pattern, but does not allow for precise online modification of the pattern itself. We have developed algorithms to monitor the ongoing walking cycle by observing the valid limb movements. In order to ensure legs coordination during walking, the CPG (Central Pattern Generator) concept was introduced, and we proposed a robust phase estimation method based on the observer of a non-linear oscillator.

There is a patent on this work, FR2908293(A1) [38]

Based on these preliminar results we are now working on the validation of the approach. We have started to modify a commercial stimulator, ODSTOCK, in order to be able to trigger it using our own sensors and algorithms (Fig. 10). A COLORS action has been obtained on this topic.

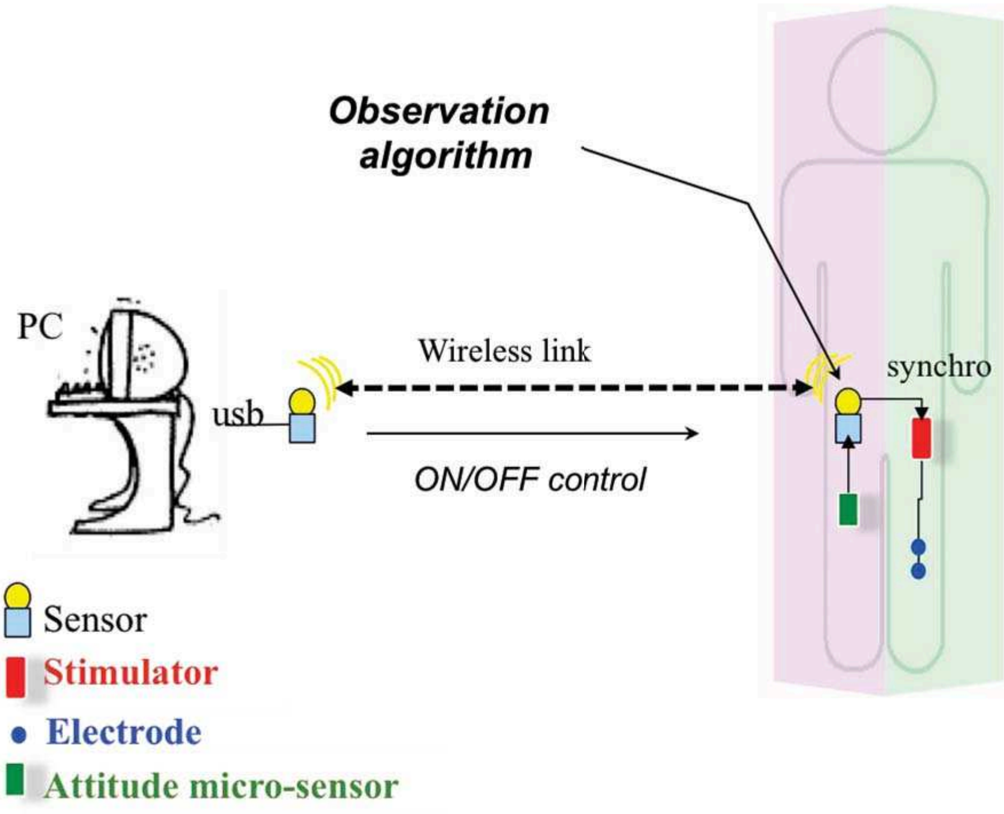

Figure 10. Principle of the system for drop-foot correction using continuous information on gait cycle phase

We have also proposed a new wireless FES architecture dedicated to drop-foot correction, based on communicating stimulator and sensors; it has been labeled from the Orpheme pole. In parallel to this, the approach of TA stimulation is being tested on individuals suffering of Parkinson Disease. This work is carried out with Mirjana Popovic (Univ. Belgrade, Serbia), Geraldine Mann (Odstock).

6.2.2. Online pathological tremor characterization and neuromusculoskeletal modeling

Participants: Antonio Bo, Philippe Poignet, Dingguo Zhang.

Details are in [33], [34], [35], [15].

In the context of the TREMOR project, whose final goal is to evaluate the use of FES to suppress pathological tremor on the upper limbs, the first part of the study was concentrated on the development of algorithms to estimate in real-time the time-varying features of tremor. Different filters and models were compared, as well as tremors related to different pathologies. Also, the methods were developed to identify tremor components from the acquired signal, considering the applied sensors also measure voluntary motion. Figure 11 illustrates the results obtained from a Parkinsonian tremor.

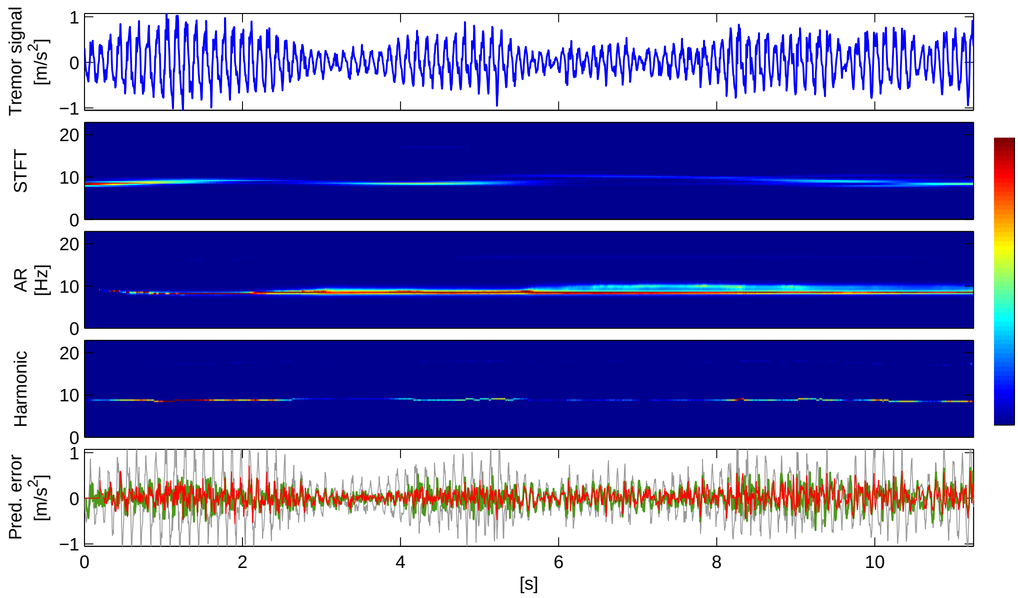

Figure 11. The upper graphics shows the measured acceleration, $\tilde{s}$. Following, the estimated spectrograms with the STFT (Short-time Fourier Transform), with the AR (Auto-Regressive) model and with the harmonic model. The bottom graphics shows the 5-step ahead prediction error of the naive prediction (gray), the AR model (red) and the harmonic model (green).

In our experiments, signals from low cost inertial sensors and electromyography were used to characterize tremor. For that purpose, a multichannel experimental acquisition system was developed and used by our clinical partner at the CHU Montpellier. The techniques developed in the work were also used to provide useful clinical measures to quantify tremor for medical studies and diagnose.

The second part of the work was devoted to the development of a neuromusculoskeletal model of the wrist joint composed by a pair of antagonist muscles. The purpose of this model is to provide a simulation environment where the study of compensation strategies with FES may be conducted before real experiments, but also to provide a mathematical model that can be used in the design of model-based control algorithms. One important feature regarding this model is the incorporation of neural dynamics and reflex loops in the modeling, since electrically stimulated and voluntary activation will occur concurrently.

6.2.3. Synthesis of optimal Functional Electrical Stimulation patterns for knee joint control

Participants: Mourad Benoussaad, Philippe Poignet, David Guiraud.

The synthesis of Functional Electrical Stimulation (FES) patterns allows to generate a desired movements of paralysed limbs of spinal cord injured subjects.

Our approach of synthesis is based on a nonlinear optimization formulation that may encounter physiological and technological constraints. For instance, we consider a biomechanical knee model and its associated agonist/antagonist muscles.

This nonlinear optimization minimize the antagonistic muscular activities in order to reduce the muscular fatigue while achieving a desired movement.

Different tests have been performed and the results compared with regard to the energetic balance. The approach is illustrated in simulation with:

1. sinusoidal desired knee joint trajectory
2. optimal reference knee joint trajectory (see figure 12)
3. without explicit reference knee joint trajectory (see figure 13)

For analyzing the results and comparing the two last methods of synthesis, we use two criteria. The first energetic criterion, at the joint torque level, is:

$$E_j = \frac{1}{t_{end}} \sum_{t=0}^{t_{end}} |(\Gamma_h(t) - \Gamma_q(t)).\dot{\theta}(t)|$$

The second energetic criterion takes into account the antagonistic muscular activities, as follows:

$$E_{mus} = \frac{1}{t_{end}} \sum_{t=0}^{t_{end}} (\alpha_q^2(t) + \alpha_h^2(t))$$

These criteria present the average of energies consumption by unit of time at different levels of actuation; 1) at torque level as classical motors used in robotics and 2) at muscle activities level, which take into account the muscular co-contraction.

We shown [14] that the trajectory tracking presents high energy consumption which demonstrates the inappropriateness of classical robotics methods for musculoskeletal system. Instead, minimization of muscle activation only gives better results with regard to energy consumption, still with a reasonnable trajectory tracking error. However, the energy consumption, in tracking trajectory case, is due to the co-contraction appearance which is very useful for the movement during the walking phases like a ground contact.

The simulations have been performed with model parameters estimated from real subject data. However, in practice, these parameters should be identified for each subject with a noninvasive way.

A global noninvasive identification protocol was applied to identify the physiological model-parameters of SUAW implanted subject. The cross validation was performed using a real knee angles of the subject under Functional Electrical Stimulation of quadriceps. The results of this study were submitted as a publication for 2009 IEEE International Conference on Robotics and Automation (**ICRA'09**)

An ethical authorization was accepted and an experimentations with spinal cord injured subject are in processing in order to validate our approach.

6.2.4. Activity of the trunk at gait initiation and during locomotion

Participants: JeanCharles Ceccato, Christine Azevedo Coste, Jean René Cazalets [UMR 5227, Bordeaux], Mathieu De Sèze [UMR 5227, Bordeaux].

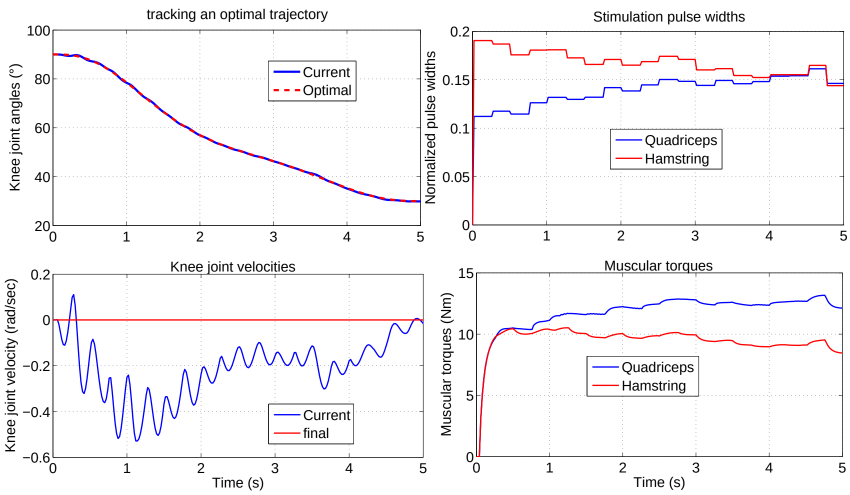

Figure 12. Tracking an optimal reference trajectory.

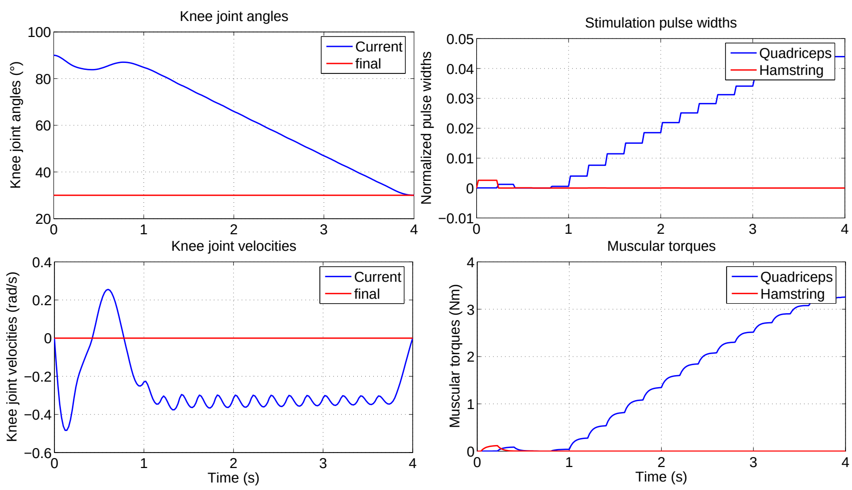

Figure 13. Synthesis without reference trajectory.

The trunk, due to its central position in the body and its mass plays an important role in maintaining balance and thus optimizing energetic cost and propulsion. In collaboration with the 'Movement, Adaptation, Cognition' laboratory, UMR 5227 in Bordeaux, we defined a protocol to record kinematics and EMG (electromyography) activities of trunk during gait initiation and gait. We used one motion capture system (ELITE, BTS, Italy), wireless EMGs (Kine, Iceland) and force plates to add some dynamic information to the data collected. Our results concerning EMG confirm and extend already obtained result at the UMR 5227 about a descendent metachronal (segment by segment) activation wave of the back muscles during gait on the next swing-leg side. This wave was found to be present with lesser amplitude before the first step at gait initiation (fig.14).

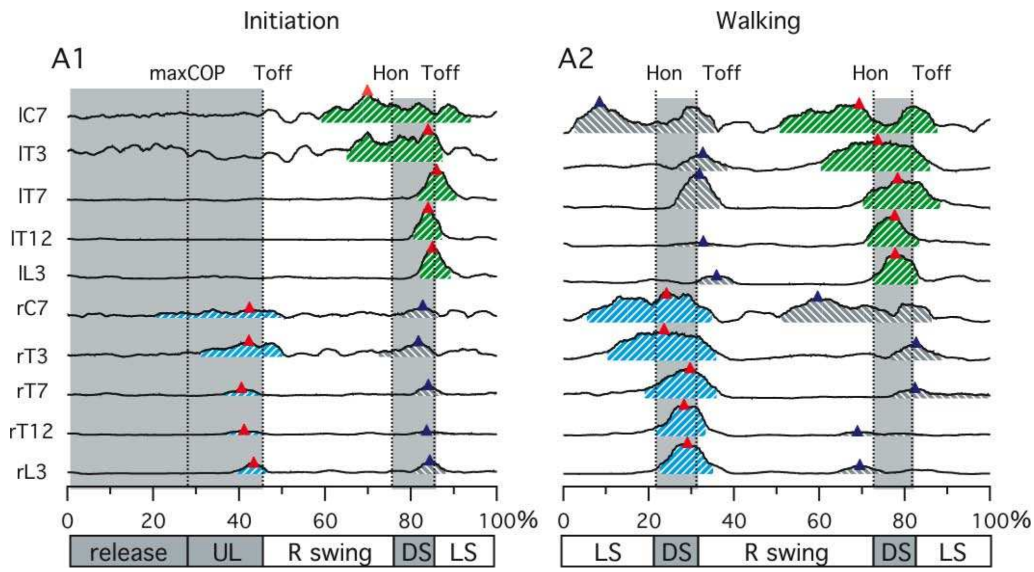

Figure 14. Bilateral EMG of trunk Erector Spinae mucles during gait initiation and walking

Kinematic analysis of the trunk movement was consistent with the description found in literature and brought some new information about gait initiation organization. We were able to locate the main areas of kinematic activity in the trunk during gait initiation and walking (fig.15) [37]).

In the kinematic data we observed that the upper trunk presents an activity before the lower trunk, from the gait initiation. This descendant activity is coherent with the observed muscular activity which is also descendant in the trunk and occurs before kinematic activity. This leads us to assume that the activity of trunk muscles drives trunk movements more than it restricts them during walking. In the hypothesis of a central pattern generator (CPG) which would control trunk activity during walking, the comparison of gait initiation and walking leads us to place the beginning of the CPG 'walking program' before the first step, around maxCOP during gait initiation phase.

To gain further insight in this trunk CPG we actually adapt an oscillator network based on analog work made on salamander (Ijspeert and Cabelguen) that models trunk activity in human waling. And to extend the capability of this model we are now doing similar measurements on walking but for different locomotion modes like running, cycling or hopping. Then, we could be able to have a model of CPG that can reproduce human trunk segmental synchronizations for different mode of locomotion. We are planning to apply these results to control electrical stimulation of lower limb in hemiplegic post-stroke patients.

6.2.5. Safe path planning (application to humanoid robots)

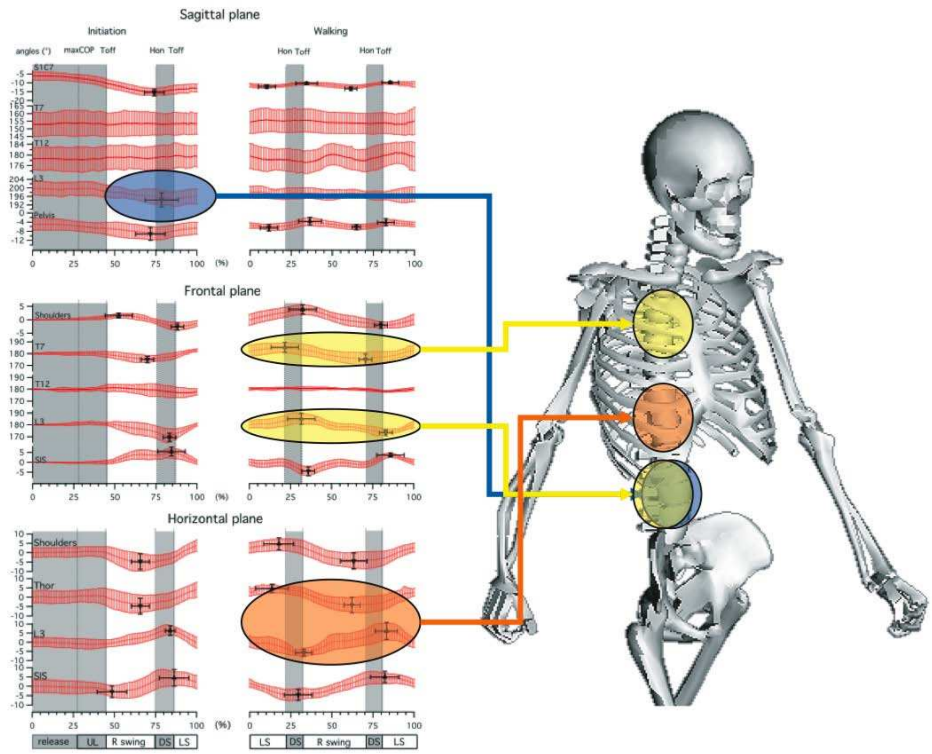

Figure 15. Kinematic of the trunk during gait initiation and walking

Participants: Sébastien Lengagne, Nacim Ramdani, Philippe Fraisse.

6.2.5.1. Objective

The main objective of this work is to allow a global navigation for humanoid robots, which ensures their integrity and balance. We decompose this objective into three different topics :

- To develop a method generating safe motions for humanoid robots. These motions are stored in a database and used online for global navigation
- To implement a state-of-the-art algorithm that creates sequences of motions wich makes it possbile to reach a goal position in an environment.
- For online adaptation of the motion to produce a new motion needed if the robot cannot reach the goal considering the ones available in the database.

6.2.5.2. Safe path planning

First we focused on the generation of safe motions and defined the motion planning problem as a Semi-Infinite Programming (SIP) problem. A SIP problem is an optimization problem with a finite number of variables and an infinite number of constraints . It consists in finding the parameter vector **X** that:

$$\begin{array}{rl} minimizes & F(\mathbf{X},t) \\ subjectto & \forall i, \forall t \in [0,T] \quad g_i(\mathbf{X},t) \leq 0 \\ and & \forall j \quad h_j(\mathbf{X}) = 0 \end{array} \tag{2}$$

where F denotes the cost (or objective) function, g_i the set of inequality constraint functions, h_j the set of equality constraint functions.

The set of the inequality constraints $g_i(\mathbf{X})$ translates the physical limits of the system. Hence, the integrity and the balance of the robot rely on the validity of these constraints. These inequality constraints must be satisfied over the whole motion duration: $\forall t \in [0,T]$. However, classical optimization algorithms use a finite number of discrete constraints. Thus the inequality constraints must be discretized.

Time-point discretization

The discretization concerns the process of transferring continuous models and equations into discrete counterparts. The usual way for discretization is to pin out a number k of discrete values during the motion (Cf. Fig:16). The equation 2 is replaced by :

$$\begin{array}{rl} minimizes & F(\mathbf{X},t) \\ subjectto & \forall i, \forall t_k \in \{0, t_1, ..., t_n-1, t_n = T\} \quad g_i(\mathbf{X},t_k) \leq 0 \\ and & \forall j \quad h_j(\mathbf{X}) = 0 \end{array} \tag{3}$$

Once the optimization is finished with optimal results, the motion satisfies all the constraints for the discrete instants, but this does not ensure that the constraints are satisfied between two time-points (Cf. Fig:16).

Time-interval discretization

The main idea of the time-interval discretization is to decompose the interval motion duration $[0,T]$ into a set of smaller intervals : $[0,t_1] \cup ... \cup [t_n-1,t_n]$ and to bound the functions $g_{j(t)}$ with a minimum and maximum value during each time interval as shown in Fig:17 and resumed in equationeq:sip2. In this case the optimization algorithm will detect the constraints violations.

$$[t_0,t_N] = [t_0,t_1] \cup ... \cup [t_{N-1},t_N] \tag{4}$$

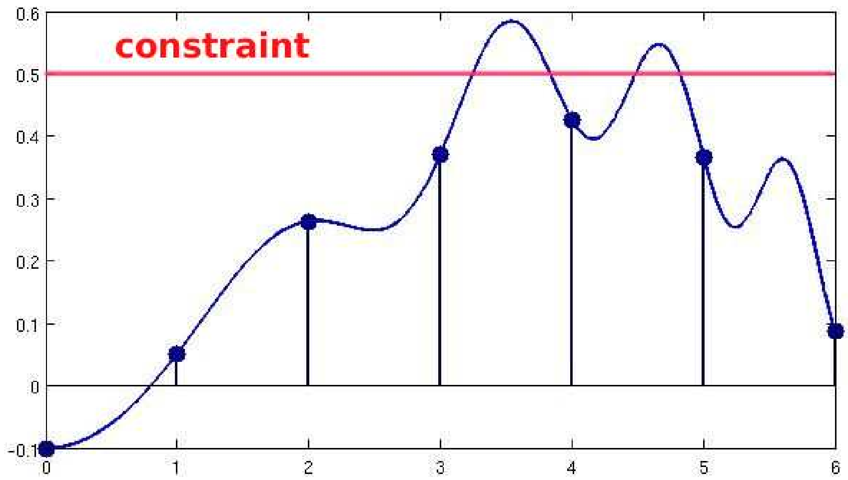

Figure 16. illustration of a time-point discretization

$$\begin{array}{rcl} minimizes & & F(\mathbf{X}, t) \\ subjectto & \forall i, \forall j \in \{1, 2, ..N\} & \max\limits_{\forall \tau \in [t_{j-1}, t_j]} g_i(\mathbf{X}, \tau) \leq 0 \\ \text{and} & & \forall j \quad h_j(\mathbf{X}) = 0 \end{array} \tag{5}$$

This method was applied successfully to the path planning of the HOAP-3 Humanoid robot considering a 2D model [26], [36], and with a more complex 3D model.

6.2.6. Modeling human postural coordination to improve the control of balance

Participants: Vincent Bonnet, Philippe Fraisse, Nacim Ramdani, Julien Lagarde [EDM-UM1], Denis Mottet [EDM-UM1], Benoit Bardy [EDM-UM1].

The task studied is based on an experimental paradigm that consisted in tracking a target motion with the head [16]. The contribution of the intrinsic human joint stiffness is investigated, and permit to improve the reproduction of the human tracking task. This postural coordination modeling offers many opportunities for better comprehension of neuromuscular movement control. And could be employed to detect human postural pathologies or to implement efficient and simple balance control principles in humanoids. The modeling we have developed, provides very realistic predictions of postural sway movements during head tracking task, including temporal phenomenon. Many of the differences between experimental and simulated results can be reduced by the use of a muscle model allowing a joint varying active stiffness. These results show that flexible joints on humanoid robot should improve the performances of the movement control in term of efficiency by minimizing the energy needed. By now, we believe that our model is promising in capturing behavioral invariants observed in human and could be employed to characterized human postural pathologies for disabled or elderly people.

6.3. Implanted neuroprosthesis design

6.3.1. Activating the natural parts through neuroprosthetic devices

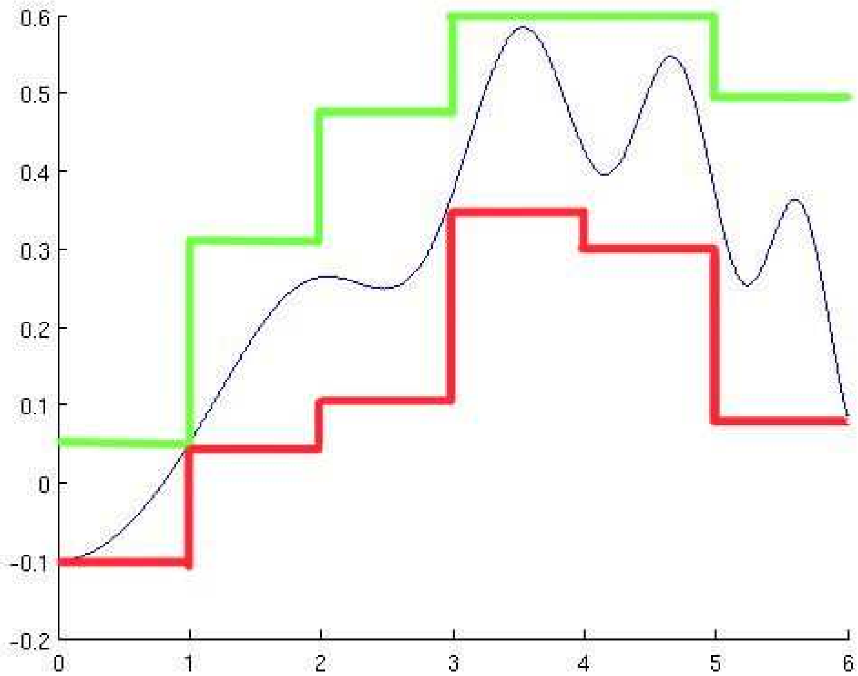

Figure 17. illustration of a time-interval discretization

6.3.1.1. Analog design of new generation of microstimulator

Participants: Guy Cathébras, Jean-Baptiste Lerat (MXM), Fabien Soulier, Serge Bernard.

In the context of different projects (Neurocom, Time), we develop a new generation of microstimator for FES. One of the main objectives is to be as generic as possible. We plan to address the stimulation of cochlea, and peripheral nerves. The general architecture consists of a Digital-to-Analog Converter (DAC), a high voltage generator and an output stage. The design schedule starts with the definition of expected specifications, then the architecture and schematics are proposed and tested thanks to simulation tools. Finally, after fabrication we have to characterize the real circuit and to check the correlation with initial specifications. For this task, there are several challenges:

- How can generic specifications be defined without resulting in a too complex system, unsuitable for body implantation.
- To optimize the global power consumption and avoid as often as posible the need of an external energy supply.
- To study the overall dependability of the system in order to guarantee the safety of the patient while maintaining the functionality in any possible conditions of application (power supply variation, impedance drift) or taking into account the process variations.

The DAC and the output stage are designed in the same ASIC (Application Specific Integrated Circuit). The third version of the DAC has been developed during this year. It is based on an original architecture allowing a very high insensitivity to process variation and thus achieves a very good linearity. Moreover, the proposed architecture guarantees the monotony of the DAC response and reduces the possible current glitch. While designing this new version, we are in the process of characterizing the preceding version. Concerning the output stage, we have proposed a novel architecture allowing a extended field of configurations. The proposed output stage can manage up to 24 poles of a single electrode. For each of these poles, which can be switched as anode or cathode (current source or sink) with no disconnection, we manage the current stimulation phase but also guarantee a complete nerve-electrode interface discharge. Finally, concerning the high voltage generator we have used a commercial component but we have proposed (and validated) a digital-like control of the circuit for possible integration into the final implant as a System-in-Package.

6.3.1.2. Distributed Stimulation Unit (DSU)

Participants: Guillaume Souquet, David Andreu, David Guiraud.

Since the beginning of the project we dealt with complementary aspects of an implanted Functional Electrical Stimulation control architecture: distributed stimulation units, deterministic implanted network, architecture implantation (electronic devices). This distributed FES architecture allows deploying over the network, closed-loop control and supervisory control [31]. This control can be supported by an implanted and/or external controller (Figure 18), in the latter case the implanted controller will act as a 'gateway' (i.e. it will ensure the communication between implanted and external FES worlds through an inductive link). Network-based closed-loop control can be performed using remote stimulation modulation allowed by DSU. Indeed a DSU executes micro-programs that can contain RT-instructions, i.e. instruction of which parameters can be remotely modified ('on-line' modulation, ie while executing the micro-program as shown on figure 19). This distributed FES architecture can be considered as an open control architecture.

6.3.1.3. Communicating between units

Participants: Guillaume Souquet, David Andreu, David Guiraud.

A DSU embeds a protocol-stack based on three layers: Application, MAC and Physical layers [7]. The logical behaviour of the MAC protocol, called STIMAP (Sliding Time Interval based Medium Access Protocol), has been formally validated. We presently work on improvments of the MAC protocol to improve DSU time-intervals synchronization (on our asynchronous network), since our last formal analysis shown that intervals overlapping could occur in a particular case. Indeed, if the selected sliding rule is 'sliding if previous node emitted' and the network performances are strongly asymmetric and the interval duration (D) is too close to the RTT (Round Trip Time), then two intervals overlapping could occur. We are performing deeper analysis

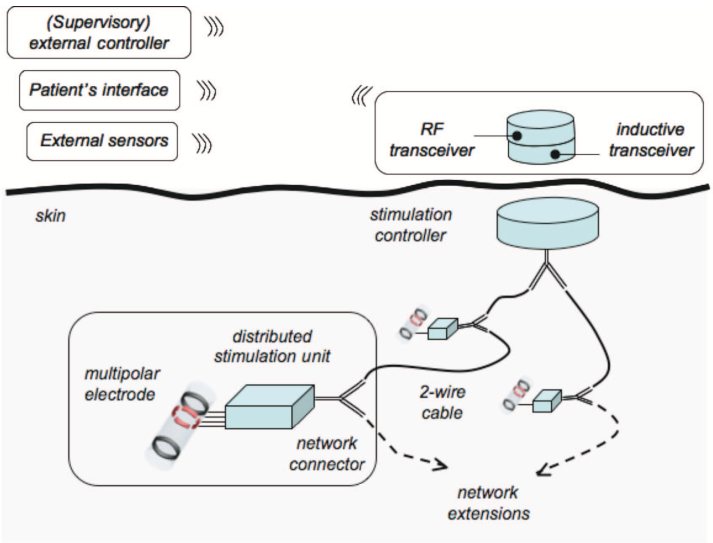

Figure 18. The global architecture may support external and implanted devices respectively communicating through RF link and 2-wire medium, these two 'worlds' being connected through an inductive link.

to precisely express the relation (i.e. the limit value of D), and studying mechanisms based on the listening of the traffic to improve time-intervals synchronization.

The MAC protocol is used in other domains than FES, like that of wireless test of SiP we mentioned in the previous report. This protocol will also be used to develop a deterministic personal area network dedicated to sport performance analysis within the ESPAD project (Embedded Sport Performance Analysis Data), leaded by TraceEdge. This FEDER project has been labelled by two poles of competitiveness: SporAltec and Minalogic.

6.3.1.4. Design and prototyping methodology

Participants: David Andreu, Guillaume Souquet.

The design and the prototyping of the DSU's digital part is based on HILECOP components (cf. section 5.1.1). We improved the design flow according to a more rigorous component-based approach combined with model transformation principle [12]. We also defined another operating mode for basic-components (elementary VHDL components), based on the basic-component's activity control approach [13]. This approach, combining PN, components and automatic model translation, has several advantages:

- it is possible to analyse and/or simulate the system's behaviour at different levels, from the PN level of abstraction to the VHDL one.
- it separates model optimization and implementation optimization, so for a given model different implementations can be tested by only modifying the elementary VHDL components according to which the model translation is performed. This allows taking into account the size and the power consumption of the resulting implementation. For instance, whatever is the ASIC fabrication technology, static power consumption is taken into account by reducing the number of gates (the circuit size) and dynamic power consumption is reduced by means of our componentÕs activity control approach [13]. This approach is a kind of clock gating based on the structure of the high-level model (PN). Since PN is an oriented graph, it is easy to determine which are the ‘pastÕ state, the ‘currentÕ state (marked places) and the 'next' one (places to be marked once the output transitions

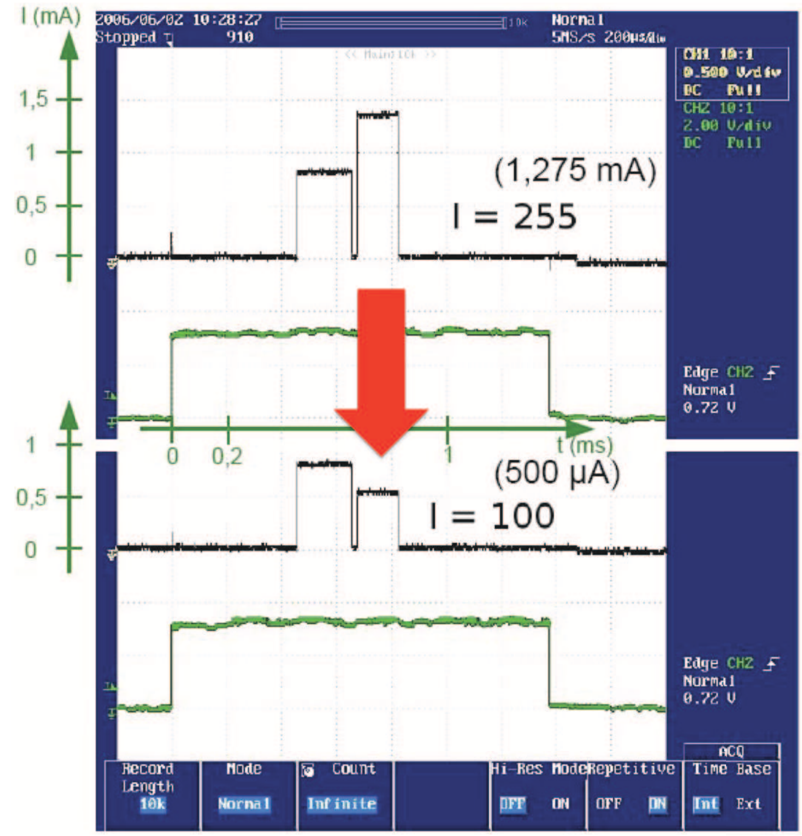

Figure 19. Example of network based stimulus amplitude modulation (observed at the output of the stimulator)

of the currently marked places will be fired). Since our VHDL description relies on a component-based approach (place and transition elementary VHDL components), it favours integrating in each component the local control of its own activity. According to simple logic rules, a basic component autonomously determines and controls it activity by enabling or not its clock (kind of clock gating). This approach of basic-componentÕs activity control is indeed interesting since on a PN there are only a few states (and a few state changing) that are simultaneously actives. Doing so we can 'switch-off' many VHDL components corresponding to the PN implementation, reducing the dynamic power consumption.

- it favours Globally Asynchronous Locally Synchronous (GALS) paradigm-based realization of complex systems since each 'macroscopic' component (e.g. a module of the DSU) has its own clock and can export signals required for its asynchronous interactions (hand-shake between macroscopic components for instance). This also contributes to power consumption reduction since it avoids overclocking of some modules, that could occur using the same frequency for the whole digital part of the DSU, i.e. the highest clock frequency of the required ones. Moreover, it avoids the potential stability and functional losses, even if not permanent incorrectness, resulting from too overclocked speed operating modules.

6.3.2. Recording afferent signals through neuroprosthetic devices

6.3.2.1. Intrafascicular electro-neurography

Participants: Milan Djilas, Christine Azevedo Coste, Ken Yoshida [IUPIU, Indianapolis], Guy Cathébras.

For further background information on this study, the reader is referred to previous activity reports and [1].

A model of aggregate firing activity in intrafascicular neural signals can be used to track muscle state only under the condition that the muscle stretch velocity is sufficiently low to have constant coefficients in the equation linking afferent nerve firing rate and muscle length. The ideal case would be to have a linear relationship for the whole range of motion of interest. Unfortunately, type Ia sensory fibers, that predominately encode information about the rate of change of muscle length, introduce a component in nerve response that makes the relationship non-linear and velocity-dependant.

The neuroelectric activity recorded with the 8-channels tfLIFE (thin film longitudinal intrafascicular electrode) is a mixture of signals from several adjacent neurons and noise. The experimental protocol in our animal experiments was designed in such a way to have activity from only afferent muscle spindle nerve fibers, these being type Ia and type II sensory nerve fibers. If this mixture could be decomposed into activities of these different sources, isolation of the activity of type II sensory fibers would allow for a developed model to be used to track muscle length variations with an a priori unknown stretch profile.

We have developed a methodology for spike detection and classification that is an expansion of idea of using complex wavelets, so it covers a range of temporal scales. Wavelet-based detection performance on the synthetic signals is compared to the amplitude thresholding method, being the most common used today. Detection is evaluated on a range of thresholds, starting from background noise level up to the maximal magnitude found in the transformed signal. The multi-scale complex CWT has an advantage that it also offers a framework for classifying the detected neural spikes. Action potentials differ in their shape and amplitude and it was necessary to choose a feature set and a distance metric with which they would be distinguishable. Classification performance is evaluated on both the synthetic ENG by comparing the wavelet-based classification to methods based on template matching and principle components analysis. These two methods, used as references in the comparison, are the most commonly used because of their relative simplicity which enables fast, real-time implementation.

The spike sorting technique was applied on experimentally recorded rabbit muscle spindle afferent nerve activity. Only flexion periods of ankle joint motion (stretch periods of the MG muscle) were analyzed. The detection threshold was chosen to be seven times the standard deviations of the background noise level (in wavelet space). The detected units were classified into 10 clusters. This matches approximately the number of units being picked up by one recording site of the tfLIFE when the muscle is maximally stretched. Results from the clustering shows that up to 2 or 3 spike classes per rabbit show a linear relationship between their computed neural firing rate and instantaneous muscle length. Since this relationship is not linear when using the aggregate afferent firing rate, the result is an indication that the algorithm is capable of isolating activity of units less sensitive to muscle stretch velocity. Results from one rabbit are shown on Fig. 20. The left plot shows the relationship between the aggregate firing rate of all detected spikes and muscle length. The relationship is clearly not linear in the region where the muscle stretch velocity slows down rapidly (region where normalized muscle length is close to 1). The right plot shows the same relationship, but this time using only the activity of the fibers insensitive to the velocity of muscle stretch. A linear regression analysis performed on both shows that the fit on the right plot is better [19] .

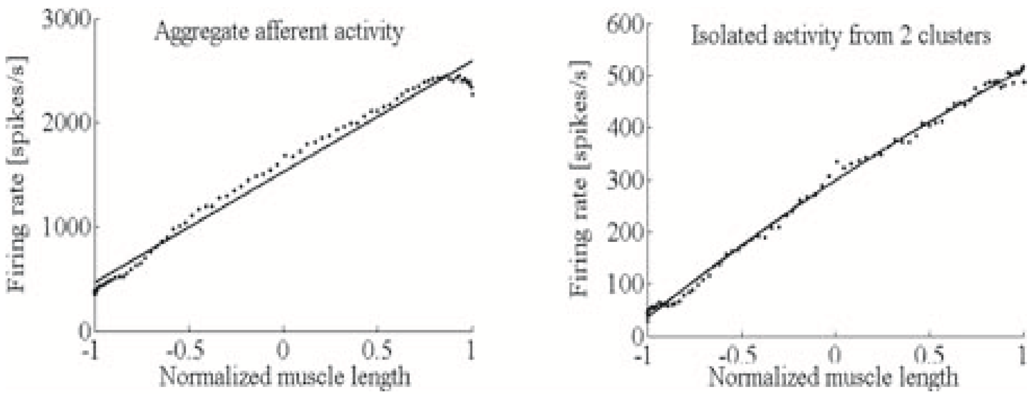

Figure 20. Afferent neural firing rate vs muscle length. The left plot shows the aggregate activity of all detected spikes. On the right, only activity from 2 clusters, having a good linear fit to the data, are used to compute the firing rate. Linear regression analysis was performed on both (full lines). On both plots muscle length was normalized by 4 mm.

6.3.2.2. Multipolar-cuff recording

Participants: Lionel Gouyet, Guy Cathébras, Fabien Soulier, Serge Bernard.

Concerning signal acquisition from the neural system, our main objective is the development of an electroneurogram (ENG) recording device. We designed an electrode and the associated micro-electronic system to improve the sensitivity and the selectivity compared to the available devices, together with a better rejection of parasitic signals such as electro-myogram (EMG).

In order to provide these recordings, we define a hexagonal multipolar cuff electrode with a large number of contacts on the nerve. Numerical simulations have shown that this specific geometry provides an efficient attenuation of muscular signals (EMG) that are parasitic signals in our context [21], [30]. This attenuation is a major concern in our research because of the level of the neural signals (μV) compared to those of parasitic signals (mV).

A specific low-noise amplifier was designed in $0.35\,\mu$m technology aimed to amplify the targeted signals (ENG) while attenuating the EMG. The EMG rejection requires an averaging of signal obtained from the different poles. Thus, the first stage of our amplifier was developed to calculate the potential difference between a pole and the average of the six surrounding poles. In order to reduce both the area overhead and the noise induced by the circuit, we have developed an architecture using as few transistors as possible.

Our perspectives for the year 2008 were several in vivo experiments to evaluate the improvement of our recording chain compared to the existing devices, but because of a problem in a part of our amplifier (supply sources), which oscillate, we couldn't reach our objectives. Moreover, a new preamplifier has been designed with better characteristics than the first one, and will be manufactured the next year; the in vivo experiments will be held with this new preamplifier. With a clever disposition of the pad ring, we should use the same materials designed for the test of the first amplifier.

Nevertheless, the results obtained with this first circuit are encouraging. Indeed, we obtained a rejection of parasitic signals about 35 dB (the circuit was designed for a 62 dB rejection but, by considering the oscillations of the supply sources, we can consider this result as good ; the new preamplifier showed in figure 21 will be designed for a 111 dB rejection of the parasitic signals).

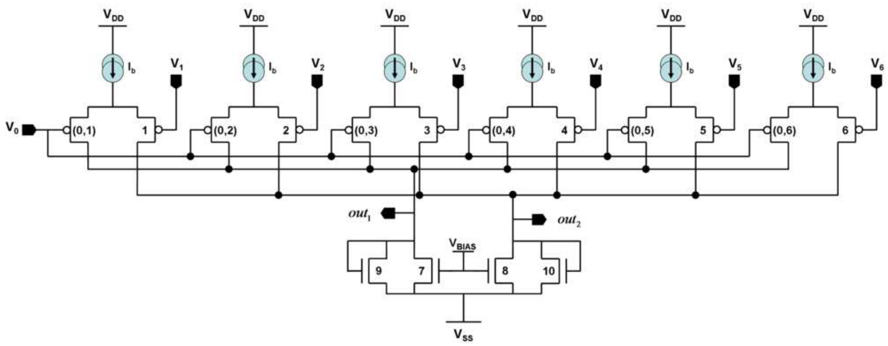

Figure 21. Structure of our new preamplifer.

Also, our perspectives for the next year are the manufacturing of our new circuit and the same experiments planned. There will be three different experimental validations:

- to evaluate the sensibility of our cuff electrode compared with a classical tripolar one (ex vivo),
- to evaluate the selectivity (ex vivo),
- to evaluate the EMG rejection and will be the one inevitably in vivo.

6.3.3. Dependability for medical implants

Participants: Fanny Le Floch, Guy Cathébras, Fabien Soulier, Serge Bernard.

Our research focuses on improving the dependability of implanted medical devices [25]. Indeed, implants have to be reliable during their all life in the human body. First, we have done a bibliography on the attributes, means and impairments of dependability (see figure 22). Then, we have identified the most sensitive points in terms of dependability for the implant itself and its environment . The next objective will be to estimate the dependability of the system and each subsystem by means of appropriate tools. As base of the study, we plan to work with MXM in a real case study of the next generation of cochlear implant. Based on the study, we will be able to propose general design guidelines for high level dependability of FES systems.

6.3.4. On-chip measurement of nerve-electrode interface impedance

Participants: Loic Bourguine, Guy Cathébras, Jean-Baptiste Lerat (MXM), Fabien Soulier, Serge Bernard.

The objective of this work is to determine impedance between the nerve and the Electrode to guarantee the safety of the patient., This measurement is actually used:

- to check the status of electrodes,
- to follow their evolution,
- to optimize consumption by monitoring the saturation of transistors.

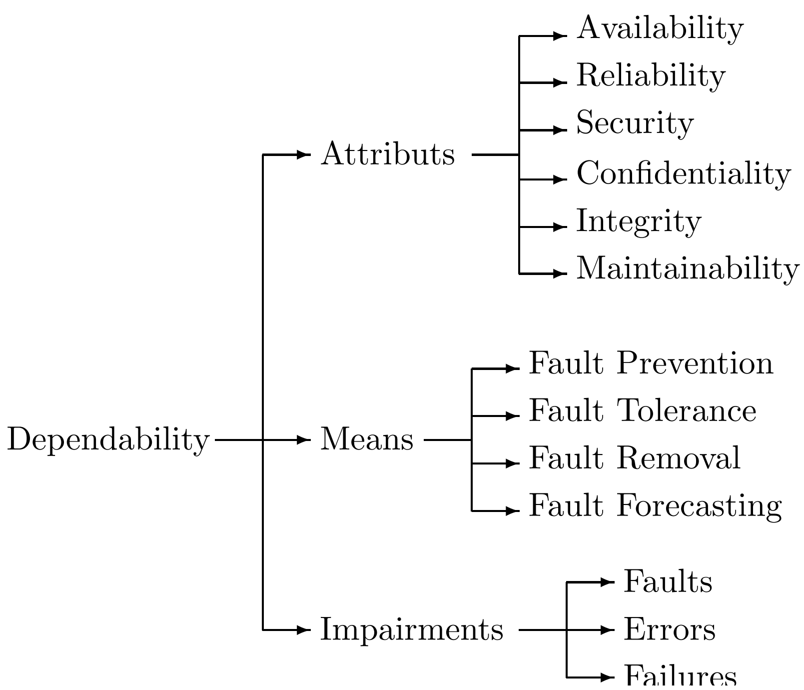

Figure 22. The dependability tree.

Previous works has determined the model of this interface. It consists of resistors and capacitors (fig. 23). R_1 mainly consists in the resistance of the wire between the implant and the electrode, C_1 is the safety output capacitor and $R_2//C_2$ stand for the physiological environment at the interface.

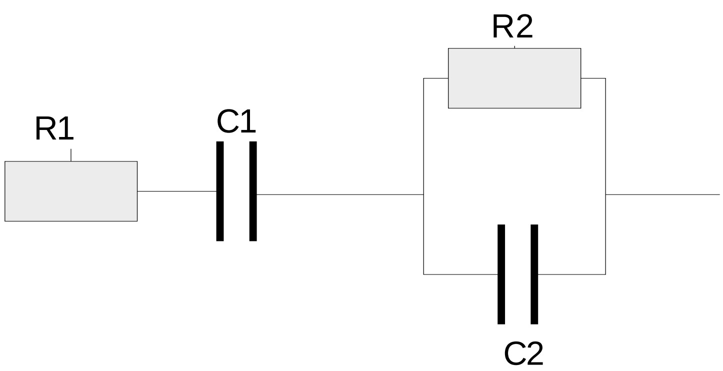

Figure 23. Approximate electric model of the nerve-electrode impedance.

The impedance measurement thus relies on finding the 4 model parameters. To be able to extract these parameters, we have developed dedicated Analog-to-Digital Converter. The main issues for this design are not the resolution or the operating frequency but the dynamic of the measured signals. Indeed, the typical values of the impedance parameters ($R_1 = 2\,k\Omega$ and $R_2 = 200\,\Omega$) and the associated current imply that the voltage across R1-R2 might be higher than 10 volts.. In the context of integrated circuit this level of signal amplitude is considered as high voltage and required very specific technology and complex design techniques. The final circuit have been design in H35 (0.35μm) *high voltage* technology from Austria Micro System company.

We have developed two ADC devices. The first one is a 5-bit flash converter based on folding and interpolation architecture. The main advantages of this ADC are the possible real time operating mode and the low silicon area overhead required.

The schematic of the converter is given in the fig. 24. This converter is very fast,the conversion time is one clock cycle, but the resolution is relatively low (5bits).

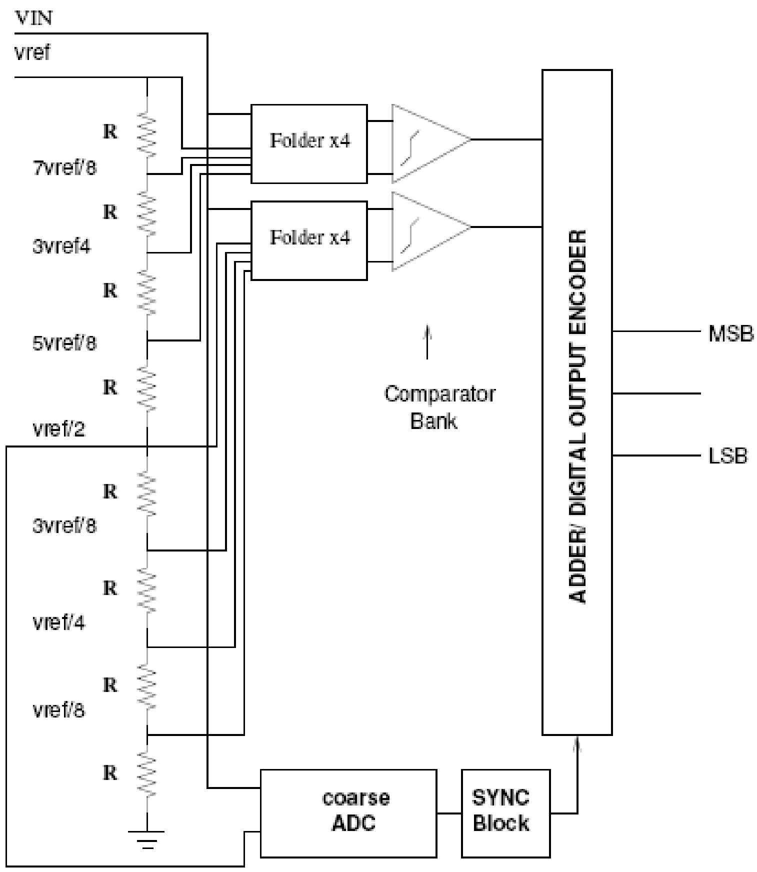

Figure 24. Interpolated flash converter structure.

The second converter is a dual slope converter. The schematic of the converter is presented in the fig. 25. With this structure we obtain a high resolution but the conversion time is quite slow if the output signal is a 10 bit signal, 2^{10} clock cycles are required for each sample conversion. After a first study of the specifications according to the initial constraints of our applications, wehave decided to focus on the Folding and interpolation architecture.

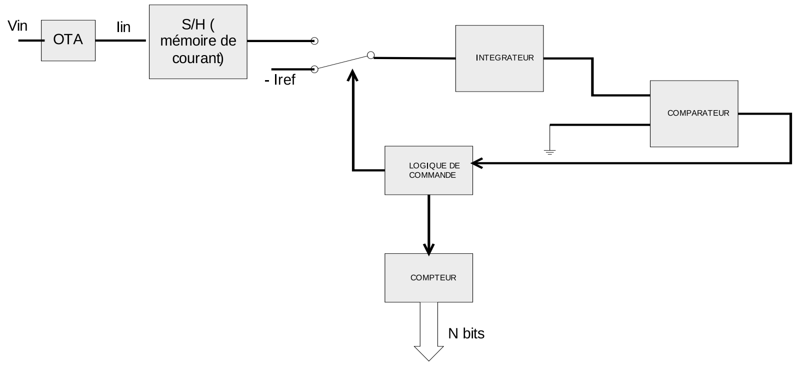

Figure 25. Dual slope converter structure.

6.4. Experimental Campaigns

We would like to emphasize a transversal part of the work carried out by our team: the experimental campaigns.

6.4.1. Human experiments with paraplegics

Participants: Maria Papaiordanidou, Mourad Benoussaad, Alain Varray, Charles Fattal, David Guiraud, Mitsuhiro Hayashibe, Patrick Benoit (physiotherapist, Propara).

We have been running several experiments on human subjects along the year.Ethical considerations and security are the most important things to keep in mind and to pay attention with. We have, for each experiment, to write down a detailed proposal that is submitted to the local ethical committee to be allowed to perform experiment. It could take more than one year from the idea to the process of the data, including protocol design, ethical committee approval, and data pre processing. All these experiments must be performed under clinician supervision in a medical center. **Complete paraplegic patients at PROPARA center (Montpellier):** in 2008, 5 patients have accepted to participate in a study concerning muscle model identification, fatigue, and movement synthesis.

6.4.2. Low frequency subcutaneous Electrical Stimulation on patients presenting Spinal Muscular Amyotrophy

Participants: Christine Azevedo-Coste, Louis Viollet (INSERM U781 -Hôpital Necker.

Spinal Muscular atrophy (SMA) is a common genetic disorder (incidence: 1/6000 to 10000 births) characterized by progressive degeneration of lower motor neurons with trunk and limb paralyses. Pathophysiology of the disease remains unclear but recent experimental data suggest a functionnal dysruption of the motoneuron-nerve-muscle communication, causing muscle weakness, denervation, amyotrophy and muscle contractures. Based on previous published data on neuromuscular disorders, we hypothetized that subcutaneous low frequency electrical stimulation of atrophic muscles in SMA could-at least partially-restore muscle volume and function and prevent muscle contracture.The objectives of this pre-study was to test this hypothesis and to evaluate whether the technique was usable or not on this type of population. We equiped one 12 years old girl presenting SMA type III, with one 2-channels electrical stimulator. Quadriceps and biceps femori of the left thigh were stimulated 90mn per day at low frequency. The patient was able to place electrodes and use the stimulator on her own and on a daily basis. After 4 months, we observed a slight increase of muscle strength and a decrease of knee contractures in the stimulated limb, measured by manual testing and orthopedic examination, compared to the controlateral limb. No pain nor fatigue effect was reported. This preliminary observation suggests that superficial muscles of the thigh are positively influenced by the electrical stimultation and that this technique is well tolerated. In a future project, we plan to evaluate more precisely, using anthropometric tools and validated motor strength test devices, the evolution of muscular and functional performances of the stimulated muscles within time and in comparison with non-stimulated muscles. We also plan to include new patients in the study.

7. Contracts and Grants with Industry

7.1. Contracts and Grants with Industry

An industrial technological transfer contract is ongoing with the MXM company that develops cochlear implant and artificial lens implant. MXM can perform also Ethylene Oxyde sterilization necessary for all our experimental setups used during surgery. A DSU prototype (named Stim-3D) and programming environment (MedStim) has been developed within this frame; it allows to graphically describe and to directly download stimulation sequences (pattern of stimulation) into the DSU component (FPGA).
A contract has been signed with Vivaltis company that is specialized in the development of external stimulation. We commonly aim at new advanced external FES system dedicated to clinical rehabilitation ([32]).

8. Other Grants and Activities

8.1. International grants

- France-Stanford Center for Interdisciplinary Studies: prospective project (Visits and exchanges) in collaboration with Professor Oussama Khatib from robotics lab (Stanford University) focused on Artificial Walking. 3 months stay for P. Fraisse in Stanford.
- National Medical Reasearch Council - Nayang Technological University. Project on Pathological tremor. LIRMM scientific leader: P. Poignet. Funding for exchange (Oct. 2006 - Oct. 2009)
- European project Time, (2008-2012). 375keuros, "*Transverse, Intrafascicular Multichannel Electrode system for induction of sensation and treatment of phantom limb pain in amputees*". Partners : AAU (Aalborg, Denmark), MXM (Vallauris, France), SSSA (Pisa, Italy), IMTEK (Freiburg, Germany), UAB (Barcelona, Spain), UCBM (Roma, Italy), IUPUI (Indianapolis, USA).
- LIRMM scientific leader for Merlion program with NTU, Singapore. Junior and senior Researcher exchange (2009-2010).
- ECONET EGIDE collaboration with Serbia and Slovenia.

8.2. National grants

- PsiRob ANR Project TREMOR on pathological tremor compensation using FES, 243 kE. Partners: MXM, Propara, CHU Montpellier (Oct. 2006 - Oct. 2009). This project is jointly conducted with the DEXTER team at LIRMM.
- EADS contract phd thesis grant support of M. Djilas (2005-2008) '*Natural sensor feedback on-line interpretation for skeletal muscle artificial control*'. 105keuros
- DGE Neurocom, (2007-2010). 475keuros, '*Implant Cochléaire 'tout implanté' pour la restauration des surdités sévères et profondes*'. Partners : MXM-Neurelec, ELA-Sorin group, APHM Hopitaux de Marseille, CHU Montpellier.

9. Dissemination

9.1. Services to scientific community

- C. Azevedo-Coste and Bernard Espiau organized a special Session at IROS conference 'Assistive robotics for functional therapy applications' in september 2008.
- D. Andreu:
  1. member of the Program Comity of 'First International Conference on Embedded Systems & Critical Applications' (ICESCA'08), Tunis, Tunisia, May 15-16, 2008 (http://www.icesca08.com).
  2. member of the Scientific Comity of '2nd IEEE International Conference on Wireless Communications in Underground and Confined Areas' (ICWCUCA'08), Val d'Or Quebec, Canada, August 25-27, 2008 (http://www.icwcuca.ca).
  3. chairman of the 'Networks' session at ICWCUCA'08, Val d'Or Quebec, Canada, August 25-27, 2008 (www.icwcuca.ca).
  4. with P. Fraisse are co-organizer and co-chairman of the 'Networks & Control' session at 17th IFAC World Congress (IFAC'08), Seoul, Korea, July 6-11, 2008 (http://www.ifac2008.org).
  5. member of the Program Comity of 'Control Architecture of Robots' workshop (CAR'08), Bourges, France, May 29-30, 2008 (http://www.bourges.univ-orleans.fr/CAR08/index.htm).
- P. Poignet:
  1. Co-organizer with J. Gangloff of one session on Medical Robotics at the French National Meeting on Robotics Research 2008 (GdR Robotique 2008)
  2. Co-organizer of the french working group on Medical Robotics of the GdR Robotique (http://www.lirmm.fr/GDRRob/gt1.htm)
  3. Member of the IFAC T2.3 technical committee on Nonlinear Control Systems
  4. Member of the CNU 61 (2008-2012)
  5. Member of the 2008 comittee evaluating the PEDR
  6. Member of the evaluation committees of the ISIR lab (Paris 6), Ecole des Mines de Douai, LGI2A (Univ. Artois)
  7. Responsible for the 'Spécialité Doctorale' in Micro-electronics and Control System (about 80 PhD students) - (http://www.edi2s.univ-montp2.fr/)
  8. Responsible for the Licence Professionnelle par Apprentissage - Métiers de la Mesure et de l'Instrumention
  9. Member of the scientific comittee of the LIRMM
  10. Associate editor of JESA

9.2. Teaching

- Guy Cathébras, Professor at Polytech'Montpellier (Electronics, Robotics and Industrial Informatics (ERII) Department), teaches: Mathematics and Signal theory for 3rd year ERII students; Analog integrated circuits: "An introduction to electronics: designing with Bipolar transistors", for 3rd year ERII students; "CMOS Analog integrated circuits design" CAD practical works for 4th year ERII students; "CMOS standard cells design" CAD practical works for 4th year ERII students. item Philippe Poignet Professor at IUT Montpellier Applied Physics teaching automatic control and signal processing.
- Philippe Fraisse, Professor at Polytech'Montpellier (ERII) teaching automatic control and networks.
- Fabien Soulier, assistant professor at Polytech'Montpellier (ERII) teaching electronics and signal processing.

9.3. Organization of seminars

- D. Andreu presented an invited communication [12] at ISABEL'08 conference.
- D. Guiraud and D. Andreu presented an invited plenary communication [7] at ICWCUCA'08 conference.

9.4. Participation in seminars and workshops

- Christine Azevedo-Coste has been invited to give a talk, "Robotics for handicap" by "Association femmes 3000" in Sophia Antipolis, January 2008
- Christine Azevedo-Coste has been invited to give a talk, "Robotics for handicap" during Handicap congress by H2A in Sophia Antipolis, May 2008, http://www.h2a-agora.com/

9.5. Theses and Internships

9.5.1. Thesis Defenses

1. **Milan Djilas**, *Natural sensor feedback on-line interpretation for skeletal muscle artificial control*, Thesis INRIA/EADS, 2005-2008.
2. **Lionel Gouyet**, *Conception d'un système pour l'acquisition sélective de signaux neurophysiologiques*, Thesis LIRMM MENRT, 2005-2008.
3. **David Guiraud**, *Modélisation du système sensori-moteur humain en vue de l'étude de ses déficiences. Développement de solutions palliatives à l'aide de neuroprothèses*, Habilitation à Diriger des Recherches, juin 2008.

9.5.2. Ongoing theses

1. D. Andreu and P. Fraisse co-supervise Mickael Toussaint, "Conception et réalisation d'une architecture de stimulation musculaire externe distribuée et sans-fil : Application au contrôle de mouvement d'une articulation", Thesis CIFRE VIVALTIS, 2008-2011.
2. Guy Cathébras and Christine Azevedo-Coste co-supervise **Milan Djilas**, "*Natural sensor feedback on-line interpretation for skeletal muscle artificial control*", Thesis INRIA/EADS, 2005-2008.
3. Guy Cathébras and Serge Bernard co-supervise **Lionel Gouyet**, "*Conception d'un système pour l'acquisition sélective de signaux neurophysiologiques*", Thesis LIRMM MENRT, 2005-2008.
4. Christine Azevedo-Coste and J.-R. Cazalets (UMR 5543-Bordeaux), co-supervise **Jean-Charles Ceccato**, "*Étude des systèmes posturaux dynamiques.*", Thesis BDI DGA-CNRS, 2006-2009 (Bordeaux/Montpellier).
5. David Guiraud and David Andreu co-supervise **Guillaume Souquet**, "*Conception et réalisation d'une architecture de stimulation électro-fonctionnelle neurale implantable pour le contrôle de la vessie*", Thesis CIFRE MXM, 2006-2009.
6. David Guiraud, David Andreu and Christine Azevedo-Coste co-supervise **Jérémy Laforêt**, "*Modélisation du recrutement sélectif en neurostimulation multipolaire multiphasique, application à la stimulation neuromotrice sélective*", Thesis LIRMM MENRT, 2006-2009.
7. Philippe Poignet and David Guiraud co-supervise **Mourad Benoussaad**, "*Synthèse de séquences de stimulation optimales pour la déambulation de patients paraplégiques.*", Thesis BDI INRIA / Région LR, 2006-2009.
8. Philippe Fraisse and Nacim Ramdani co-supervise **Sébastien Langagne**, "*Génération de mouvement adaptative sous contraintes pour la déambulation d'un patient paraplégique par la prise en compte des mouvements volontaires de ses membres supérieurs.*", Thesis BDI INRIA / Région LR, 2006-2009.

9. Philippe Poignet supervises **Antônio Bo**, “*Compensation active du tremblement pathologique du membre supérieur via la stimulation électrique fonctionnelle.*"
10. David Guiraud and Alain Varray supervise **Maria Papaiordanidou**,"*Nature périphérique et centrale de la fatigue musculaire.*"

9.5.3. Starting theses

1. Serge Bernard, Guy Cathébras co-supervise Fanny Le Floch, *sûreté de fonctionnement des circuits implantables dans le corps humain.*, MENRT.
2. Guy Cathébras Fabien Soulier co-supervise Olivier Rossel, *Circuits intégrés de recueil et d'interprétation des signaux nerveux*, Axa fundation.
3. Jérôme Bourien (INM, Montpellier) and Christine Azevedo-Coste, co-supervise **Christophe Michel**, '*Modélisation de l'efférence latérale du système auditif périphérique*', CIFRE MXM.

9.5.4. Internships

- **2007-2008**
  1. Jérôme Bourien (INM, Montpellier) and Christine Azevedo-Coste, co-supervise **Christophe Michel**, MASTER, “*Modélisation système auditif périphérique : de l'onde sonore à l'émission du potentiel d'action*"
  2. David Andreu supervises Grégory Angles. "SENIS Manager: an environnent for configuration, deployment and exploitation of a distributed neural FES architecture", 6 month computer engineer contract (XMX). This project is carried out within a technological transfer frame with MXM lab. Company.
  3. David Andreu supervises Robin Passama. "Conception et réalisation d'un environnement de développement d'applications de contrôle basées sur la stimulation électro-fonctionnelle", 12 month computer engineer contract. This project is carried out within the NEUROCOM project.
  4. David Andreu supervises Steve Coustenoble, "Passerelle pour Architecture de Stimulation Electro-Fonctionnelle Per-Opératoire" Projet Industriel de Fin d'Etudes (MEA engineer final year), from September 2007 to January 2008.
  5. David Andreu supervises Roland Tiraboschi and Guillaume Roquet, "Contribution au Test de Production des SiP : Architecture et Protocoles de Communication Embarqués au sein du SiP". Projet Industriel de Fin d'Etudes (MEA engineer final year), from September 2007 to January 2008.
  6. David Andreu supervised Benjamin Braga, “Prototypage et validation d'un exécuteur embarqué dédié à la stimulation implantée de la cochlée", Engineer final internship, from February 2008 to June 2008.
- **2008-2009**
  1. David Andreu supervises Olivier Blanc, “Contrôle et ordonnancement de stimulation électro-fonctionnelle implantée", Summer engineer internship and Projet Industriel de Fin d'Etudes (ERII engineer final year), from July 2007 to January 2008.
  2. David Andreu supervises Jérôme Barbaras, “Conception et réalisation d'un ASIC numérique d'une unité de stimulation électro-fonctionnelle répartie", Summer engineer internship and Projet Industriel de Fin d'Etudes (ERII engineer final year), from July 2007 to January 2008.
  3. David Andreu supervises Nicolas Glachant and Jonathan Presti, “Datation relative de données et recalage temporel dynamique sur une architecture de stimulation électro-fonctionnelle implantée", Summer engineer internship and Projet Industriel de Fin d'Etudes (ERII engineer final year), from July 2007 to January 2008.

4. David Guiraud and Mitsuhiro Hayashibe co-supervise Floor Campfens, "Design of an activation block for the DEMAR muscle model" International internship for master degree at Department of Biomedical Engineering (Prof. Peter Veltink), University of Twente, from March 2008 to June 2008.
5. Fabien Soulier supervises Gabriel Confais, Circuit de génération haute tension, Universitary diploma of technology internship.
6. Fabien Soulier supervises Antoine Escallon, caractérisation du Convertisseur Analogique-Numérique d'un stimulateur implantable dédié à la stimulation électrique fonctionnelle, Universitary diploma of technology internship.
7. Fabien Soulier supervises Jia Zhai, Mesures d'impédances électrochimiques pour la caractérisation de lÕinterface électrode-milieu physiologique, Master internship.

9.5.5. Contract Engineers

- David Andreu supervised Robin Passama. “Conception et développement d'une infrastructure logicielle de contrôle et d'exploitation d'une architecture de FES externe", Computer Science Engineer (1 year contract, DEMAR financial support).
- David Andreu supervises Grégory Angles. “Conception et réalisation d'un environnement logiciel, basé sur Eclipse, pour le prototypage rapide sur composants électroniques programmables (HILECOP)". Computer Science Engineer, INRIA ODL contract (2 years contract, INRIA financial support).
- David Andreu supervises Robin Passama. “Développement des aspects logiciels de contrôle et d'exploitation des neuroprothèses", Computer Science Engineer (1 year contract, DEMAR financial support).
- Serge Bernard and Fabien Soulier co-supervise Loic Bourguine 'Micorelectronics design'. (2 year DEMAR financial support)

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[2] L. GOUYET. *Conception d'un système pour l'acquisition sélective de signaux neurophysiologiques*, Ph. D. Thesis, Université Montpellier II - Sciences et Techniques du Languedoc, December 2008.

[3] D. GUIRAUD. *Modélisation du système sensori-moteur humain en vue de l'étude de ses déficiences, Développement de solutions palliatives à l'aide de neuroprothèses*, Habilitation à Diriger des Recherches, Université Montpellier II - Sciences et Techniques du Languedoc, may 2008.

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[8] C. AZEVEDO COSTE. *Concevoir les technologies de l'information pour la société de demain : quel rôle pour les femmes ?*, in "Rencontre débat Femmes 3000 Côte d'Azur, France", 01 2008, http://hal-lirmm.ccsd.cnrs.fr/lirmm-00331881/en/.

[9] C. AZEVEDO COSTE. *La robotique au service du handicap moteur : Tour d'horizon des travaux de recherche dans le domaine*, in "Salon du handicap, de l'accessibilité et de l'autonomie, France", 05 2008, http://hal-lirmm.ccsd.cnrs.fr/lirmm-00331882/en/.

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[26] S. Lengagne, N. Ramdani, P. Fraisse. *A New Method for Generating Safe Motions for Humanoid Robots*, in "IEEE International Conference on Humanoid Robots", 12 2008, http://hal-lirmm.ccsd.cnrs.fr/lirmm-00326044/en/.

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