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Towards a Co-contraction Muscle Control Strategy For Paraplegics

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Abstract— **Functional electrical stimulation (FES) can help in regaining limited locomotor activities in humans with paraplegia through electrical stimulation of the lower extremity muscles, mainly the quadriceps and hamstrings. Closed loop position control of a flexion-extension knee under (FES) based on a high order sliding mode techniques has been presented. Stimulation currents amplitude were assessed via a variable structure control systems (VSCS). Special attention is given to a 2-sliding (Prescribed convergence law) control algorithm. The controller was used to control shank movement and has shown a robustness against force perturbation as well as high capability of tracking a pre-defined reference trajectory. Co-contraction of the antagonistic muscles, basically quadriceps and hamstrings, may yield an increasing joint stiffness and a stable movement. The redundancy of two muscles spanning the knee joint in co-contraction, is solved by a linear minimization of the sum of the stress in the antagonistic muscles.**

I. INTRODUCTION

Although, open loop control strategies do not account for any changes in the muscles performance such as fatigue or load changes, they are widely used in clinics due to their relative simple implantation [1]. As a response to any external disturbances, closed loop controllers use sensor feedback to update the stimulation levels (intensity and pulse width) [2], [3], [4]. Some authors use a simple PID controller [5], Knee Extension Controller KEC [6], a combination of feedback and feedforward control or an adaptive approach [7]. The so-called "subject centered" strategies [2],[4] introduce the voluntary contribution of the upper body of the patient as an essential part of the control diagram. This later is not yet adopted in clinical use because of the relative high number of parameters to be identified. A better understanding of the muscle element as well as finding a compromise between a complex control strategy and a satisfactory one, should be taken into account. In the current study a mathematical muscle model representing the dynamic phenomenon has been used. The number of recruited motor units increases as a function of intensity stimulation I. This phenomenon is modeled by an activation model (representing the ratio of recruited fibers α and the chemical control input u_{ch}) and a mechanical model (muscle contraction). The term antagonist will be used for muscles, whose moment in a two-dimensional system about a joint is in the opposite direction as the resultant joint moment [8]. The nonlinearities of the muscle model, and the required robustness regarding parameter variations and external disturbances lead us to adopt a controller relying on the sliding mode theory. Few studies have applied this technique to a musculoskeletal system [9]. The High Order Sliding Mode (HOSM) generalizes the basic sliding mode idea by acting on the higher order time derivatives of the sliding variable instead of influencing the first time derivative as it happens in the standard sliding mode control or first order sliding mode (FOSM). This technique has the potential to provide greater accuracy. A 2-sliding mode control may provide up to second order of sliding precision with respect to measurement interval [10]. In the current study a state space model of the knee with two antagonist muscles was derived. The antagonistic function of a muscle is not necessarily restricted to oppose motion but may give stability and stiffness to a joint. An unknown perturbations were added to the muscle forces generated in order to study the accuracy and robustness of the controller. In the next section, the system modeling is presented, it includes model of the knee-muscles and its state space formulation. In the third section, the theory of a high order sliding mode controller is presented. The co-contraction results will be detailed in the last section.

II. SYSTEM MODELLING

A. Biomechanical model

The biomechanical model consists of two segments representing respectively the shank and thigh connected to each other by a revolute joint with one degree of freedom (*fig.1*). The thigh is supposed fixed with respect to the patient laying supine. Two agonist/antagonist muscles act on the knee: the quadriceps acts as an extensor muscle while the hamstrings is the flexor muscle. As a result two forces F_q and F_h cause respectively the extension and flexion of the knee. F_q and F_h are the inputs of the biomechanical model while the angle Θ is the corresponding output. (Θ = 0 corresponds to full extension of the knee, $\Theta = 150°$ corresponds to the maximum flexion and $\Theta = 90°$ represents the resting position).

- L_0 = Thigh length,
- L_1 = Shank length,
- O = Center of rotation (Knee),
- L_{iq} = Distance between O and the insertion point of the quadriceps on the shank,
- L_{ii} = Distance between O and the insertion point of hamstrings on the shank,

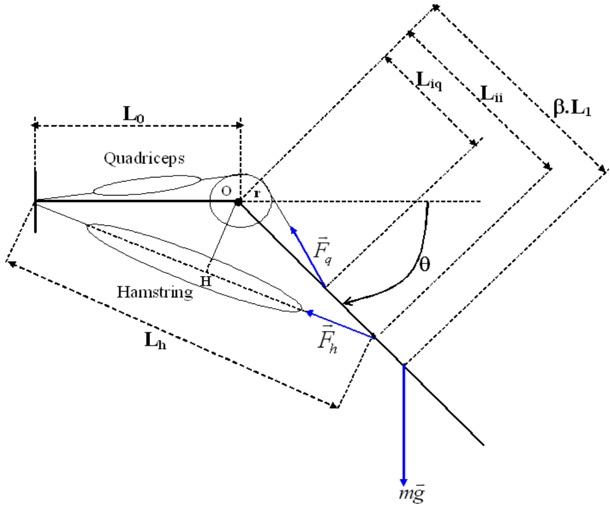

Fig. 1. Biomechanical model of the knee with two muscles: Quadriceps and Hamstrings

- L_q = Length of the quadriceps,
- L_h = Length of the hamstrings,
- r = Pulley radius,
- H = Orthogonal projection of O on Li,
- F_q = Extension force generated by the quadriceps,
- F_h = Flexion force generated by the hamstrings,
- G = Gravity force vector,
- m = Mass of the shank.

The geometric constraints allow us to evaluate quadriceps length L_q depending on the knee angle variable Θ:

$$L_q(\Theta) = \sqrt{L_0^2 - r^2} + r\Theta + \sqrt{L_{iq}^2 - r^2}$$

and the hamstrings length $L_h(\Theta)$:

$$L_h(\Theta) = \sqrt{L_0^2 + L_{ii}^2 + 2L_0 L_{ii} cos(\Theta)}$$

Moment arm of the quadriceps is supposed to be constant and equal to the pulley radius while the moment arm of the hamstrings depends on the variable angle Θ.

$$\overline{OH} = \frac{L_0 L_{ii} sin(\Theta)}{\sqrt{L_0^2 + L_{ii}^2 + 2L_0 L_{ii} cos(\Theta)}}$$

Parameters of the above model were taken for an average person: $L_0 = 50\,cm, L_{iq} = 4\,cm, L_{ii} = 5\,cm, r = 1\,cm, F_v = 0.5\,N.m.s, I = 0.0476\,N.m.s^2, g = 9.8\,N.m^{-2}, m = 3.5\,Kg$. β corresponds to the position of the center of gravity of the shank (< 1), F_v corresponds to coefficient of viscous friction. Figure 2 shows the muscle model used [11], composed from a parallel element E_p and two elements in series E_s (elastic element) and E_c (contractile element). This model is controlled by two variables: u_{ch} a chemical control input and α, the ratio of the recruited fibers. We have described this model by two differential equations where the outputs are K_c and F_c representing respectively the stiffness and the force generated by the contractile element. K_0 and F_0 are the maximum values of K_c and F_c.

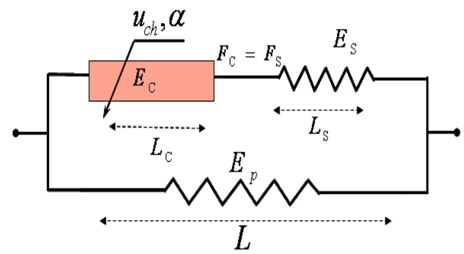

Fig. 2. Muscle model and particularization of EC

TABLE I
PARAMETERS OF BOTH MUSCLES QUADRICEPS AND HAMSTRINGS

Muscle model parameters	Variable	Numeric value (quadriceps -hamstring)	Unit
stiffness of E_s	K_s	1.10^4	N/m
Contractile element length E_c	L_{c0}	$41.10^{-2}, 38.10^{-2}$	m
Elastic element length E_s	L_{s0}	$8.10^{-2}, 10.10^{-2}$	m
Maximum isometric muscle force	F_{max}	3100, 1295	N

$$\left\{ \begin{array}{lcl} \dot{K}_c & = & (s_0 \alpha K_0 - s_u K_c + s_v q \frac{s_0 \alpha F_0 K_c - s_u F_c K_c}{1 + pK_c - s_v q F_c}) u \\ & & - \frac{s_v a K_c}{1 + pK_c - s_v q F_c} \\ \dot{F}_c & = & (\frac{s_0 \alpha F_0 - s_u F_c}{1 + pK_c - s_v q F_c}) u + (\frac{bK_c - s_v a F_c}{1 + pK_c - s_v q F_c}) \dot{\varepsilon} \end{array} \right.$$

$$s_u = sign(u) = \left\{ \begin{array}{lll} -1 & if & u < 0 \\ +1 & if & u > 0 \end{array} \right.$$

$$s_v = sign(\dot{\varepsilon}_c) = \left\{ \begin{array}{lll} +1 & if & \dot{\varepsilon}_c > 0 \\ -1 & if & \dot{\varepsilon}_c < 0 \end{array} \right.$$

$$s_0 = \frac{1 + s_u}{2} \quad a = \frac{L_0}{L_{c0}} \quad b = L_0 \quad p = \frac{1}{K_s} \quad q = \frac{1}{L_{c0} K_s}$$

$$\varepsilon_c = \frac{L_c - L_{c0}}{L_{c0}} \quad \varepsilon_s = \frac{L_s - L_{s0}}{L_{s0}} \quad \varepsilon = \frac{L - L_0}{L_0} \quad L = L_c + L_s$$

where s_u, s_0 and s_v are the signs of the control and the velocities of the contractile element, L_c and L_s represent respectively the length of the contractile and the elastic elements. The ratio of recruited fibers α is considered as a global scale factor which gives the percentage of the maximal possible force which can be generated by the muscle. The parameters of the muscles as shown in (*Table1*) were taken from [12].

B. Muscle-Knee: State space Model

The model of the muscles and knee joint can be rewritten as a non-linear state space function:

$$\dot{\mathbf{X}} = \mathbf{f}(\mathbf{x}, \mathbf{t}, \mathbf{U})$$

Where $\mathbf{X}$=$[X_1 ... X_6]^T = [K_q \quad K_h \quad F_q \quad F_h \quad \Theta \quad \dot{\Theta}]^T$ is the state vector and $\mathbf{U} = [u_q \quad \alpha_q \quad u_h \quad \alpha_h]^T$ the control vector. The variable Θ represents the joint knee angle. The state variables K_q, F_q, u_q, α_q and K_h, F_h, u_h, α_h are respectively the state variables of the quadriceps and hamstrings. The state space model of the knee joint can be expressed as:

$$\dot{X}_1 = (s_{0q}\alpha_q K_{0q} + s_{vq}q_q \tfrac{s_{0q}\alpha_q F_{0q}X_1 - s_{uq}X_3 X1}{1+p_q X_1 - s_{vq}q_q X_3})u_q$$

$$-\tfrac{s_{vq}\alpha_q X_1 r X_6}{L_{0q}+p_q X_1 - s_{vq}q_q X_3} - s_{uq}X_1 u_q$$

$$\dot{X}_2 = (s_{0h}\alpha_h K_{0h} + s_{vh}q_h \tfrac{s_{0h}\alpha_h F_{0h}X_2 - s_{uh}X_4 X_2}{1+p_h X_2 - s_{vh}q_2 X_4})u_h$$

$$-\frac{s_{vh}a_h X_2 L_0 L_{ii} sin(X_5)}{L_{0h}\sqrt{L_0^2+L_{ii}^2+2L_0 L_{ii}cos(X_5)}(1+p_h X_2 - s_{vh}q_h X_4)}$$

$$-s_{uh}X_2 u_h$$

$$\dot{X}_3 = \tfrac{s_{0q}\alpha_q F_{0q} - s_{uq}X_3}{1+p_q X_1 - s_{vq}q_q X_3}u_q - \tfrac{b_q X_1 - s_{vq}a_q X_3 r X_6}{L_{0q}(1+p_q X_1 - s_{vq}q_q X_3)}$$

$$\dot{X}_4 = -\frac{b_h X_2 - s_{vh}a_h X_4 L_0 L_{ii} sin(X_5)}{L_{0h}\sqrt{L_0^2+L_{ii}^2+2L_0 L_{ii}cos(X_5)}(1+p_h X_2 - s_{vh}q_h X_4)}$$

$$+\tfrac{s_{0h}\alpha_h F_{0h} - s_{uh}X_4}{1+p_h X_2 - s_{vh}q_h X_4}u_h$$

$$\dot{X}_5 = X_6$$

$$\dot{X}_6 = \tfrac{1}{I}(X_3 r - X_4 \tfrac{L_0 L_{ii} sin(X_5)}{\sqrt{L_0^2+L_{ii}^2+2L_0 L_{ii}cos(\Theta)}} - F_v X_6$$
$$-mgcosX_5 \beta L_1)$$

III. High Order Sliding Mode

The sliding mode control, has become recently widely used due to its high accuracy and robustness with respect to parameters uncertainty disturbances. The control task is to keep a constraint, given by equality of a smooth function called sliding surface, to zero. The dynamic smoothness in the vicinity of the sliding mode represents the sliding order of the system [10]. As a generalization of the classical sliding mode, this notion has been extended to the high order sliding mode. In this case, the control acts on the higher order time derivatives of the sliding variable instead of acting on its first time derivative. Thus the discontinuity of the control vector does not appear in the first $(r-1)^{th}$ total time derivative.

$$\frac{\partial s^{(i)}}{\partial u} = 0, (i = 1, 2, ..., r-1), \frac{\partial s^{(r)}}{\partial u} \neq 0 \quad (1)$$

s, r represent respectively the sliding surface and the relative degree. u is the resulting control vector. Consequently we have:

$$s = \dot{s} = \ddot{s} = ... = s^{r-1} = 0 \quad (2)$$

The sliding surface used to constraint the dynamic behavior of the knee joint is a first order differential equation chosen as:

$$s = (\dot{\Theta}_d - \dot{\Theta}) + \lambda(\Theta_d - \Theta) \quad (3)$$

Where $\dot{\Theta}_d$, Θ_d are respectively the desired velocity and desired position, λ is a positive coefficient. Higher values of λ, lead to a faster convergence along the sliding surface to the zero point of the phase-plane. Let us consider the sliding surface equation *(3)* in order to determine the relative order of the controlled system. We obtain the following result:

$$\frac{\partial \dot{s}}{\partial u} = 0, \frac{\partial \ddot{s}}{\partial u} \neq 0 \quad (4)$$

Therefore, the relative degree of the sliding mode control is $r = 2$. Considering the step response case ($\ddot{\Theta}_d = \dot{\Theta}_d = 0$), the second time order derivative of the sliding surface can be written as:

$$\ddot{s} = -\ddot{X}_6 - \lambda \dot{X}_6 \quad (5)$$

Inserting the expressions of $\dot{X}_6$ and $\ddot{X}_6$ within equation *(5)* allows writing the second time derivative of s as:

$$\ddot{s} = \varphi(x,t) + \gamma(t,x)u \quad (6)$$

It is assumed that $\Phi > 0$, $|\varphi| \leq \Phi$,$0 < \Gamma_m \leq \gamma \leq \Gamma_M$ [10], where s_0, $u_0 < 1$, Γ_m, Γ_M and Φ are positive constants.We express the equation *(6)* as:

$$\begin{cases} \dot{y}_1 &= y_2 \\ \dot{y}_2 &= \varphi(x,t) + \gamma(t,x)u \end{cases}$$

Where $y_1 = s$. In that case, the problem is equivalent to the finite time stabilization problem for the uncertain second-order system.

The main drawback in HOSM implementation is the increasing state information demand. For a r-sliding controller ($s^r = 0$), $s, \dot{s}, ..., s^{r-1}$ need to be available. In the current study we have implemented the prescribed convergence law algorithm. This choice has been made based on criteria of robustness, finite time convergence, reduction of chattering and relative low information of the state variables [13] As shown in *Fig.3*, the trajectories converge in finite time to the zero of the phase plot. The general formulation of such a class of a sliding mode control algorithm is:

$$\dot{u} = \begin{cases} -u & if \quad |u| > 1 \\ -V_M sign(y_2 - g_c(y_1)) & if \quad |u| \leq 1 \end{cases} \quad (7)$$

Where V_M is a positive constant and g_c a continuous function (*Fig.3*). Moreover, this function must verify some specific conditions (see [13]).

$$g_c(y_1) = -\lambda_1 |y_1|^{\rho} sign(y_1),\ \lambda_1 > 0,\ 0.5 \leq \rho < 1 \quad (8)$$

The sufficient condition for the finite time convergence to the sliding manifold is defined by the following inequality:

$$V_M > \frac{\Phi + sup[\dot{g_c}(y_1) g_c(y_1)]}{\Gamma_m} \quad (9)$$

Larger values of λ_1 accelerate the convergence to reach the sliding surface and provide better robustness and stability.

The control law need the sliding variable s and it first time derivative $\dot{s}$. No explicit knowledge of other system parameters were needed. This not only reduces the computational

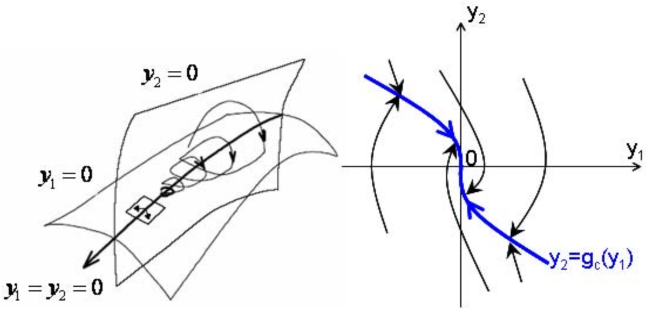

Fig. 3. Phase plot of the prescribed convergence law algorithm

burden of the controller but also make it easy to be tuned via only three parameters ρ, λ_1 and V_M [17], [18].

IV. Co-Contraction of Flexor-Extensor

Muscle co-contraction can be defined as the simultaneous activation of agonist and antagonist muscle groups crossing the same joint and acting in the same plane [19]. The opposite muscles, quadriceps and hamstrings in this case, act simultaneously and thereby may increase the stiffness at the knee joint. According to the state-space of the muscle-knee model, a co-contraction effect would increase the state variable derivatives $\dot{X}_3$ and $\dot{X}_4$ (II.B), which will ensure a safer movement of the shank. The co-contraction phenomenon was represented by a simultaneous contraction of the muscle and its antagonist muscle via a static factor weighting ξ_q, ξ_h for quadriceps and hamstrings respectively. The amount of co-contraction was evaluated, based on a static linear constraints optimization of the muscle forces acting on the knee.

$$S_{ob}(F_i) = \sum_{i=q,h} \left(\frac{F_i}{F_{max,i}} - \xi_i\right)^2 \qquad (10)$$

This objective function is subject to the following constraints:

$$\begin{cases} 0 \le F_i \le F_{max,i} \\ \sum_i \; r_i F_i = M \end{cases} \qquad (i = q, h) \qquad (11)$$

$r_i, F_i, F_{max,i}$ and M represent respectively the moment arm of the muscles as calculated in the above formulations, the actual and the maximum force generated by each muscle and the resultant moment force at the knee joint (*Fig.4*). It has been shown in the case of a planar one degree of freedom (knee joint) [20] that the analytical solution of such a problem can be formulated as:

$$F_i = \xi_i \; F_{max,i} + r_i F_{max,i}^2 \left(\frac{M - \xi_i \sum_i (r_i F_{max,i})}{\sum_i (r_i F_{max,i})^2}\right) \quad (i = q, h) \quad (12)$$

Based on this solution we were able to compute the relation between the amount of co-contraction, relative to the maximum force of both quadriceps and hamstring via the weight factors ξ_q and ξ_h respectively.
At the output of the *HOSM*, the control vector u of the prescribed convergence law [10] is defined as:

$$u = -\int V_M sign(y_2 - g_c(y_1))dt \qquad (if \; |u| \le 1) \qquad (13)$$

We propose a method to define the contribution of the control vector u stemming from the 2-sliding controller to calculate the needed electrical current stimulation values. According to the sign of the resulting control variable (u) at the output of the *HOSM* controller and the value of ξ_i, we have chosen to stimulate whether the quadriceps, the hamstrings or both.

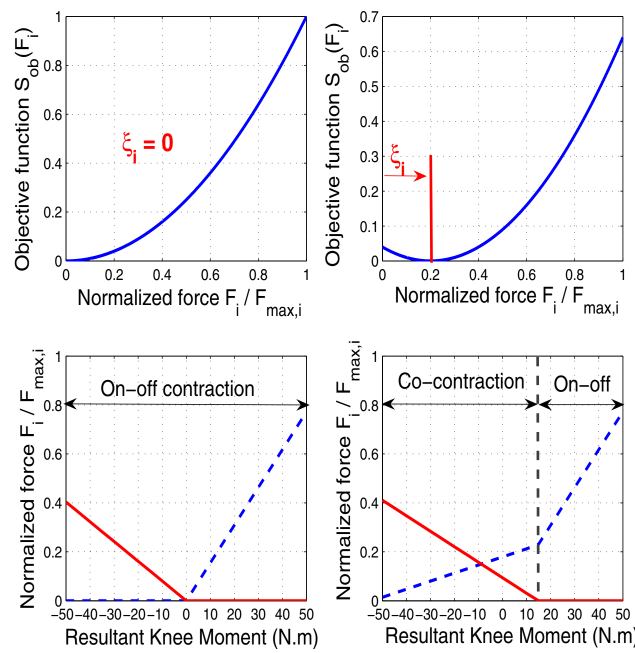

Fig. 4. Objective functions for an amount of co-contraction ($\xi_i = 0$, $\xi_i = 0.2$) and the normalized force against the resultant knee moment, (- for quadriceps and - - for hamstrings)

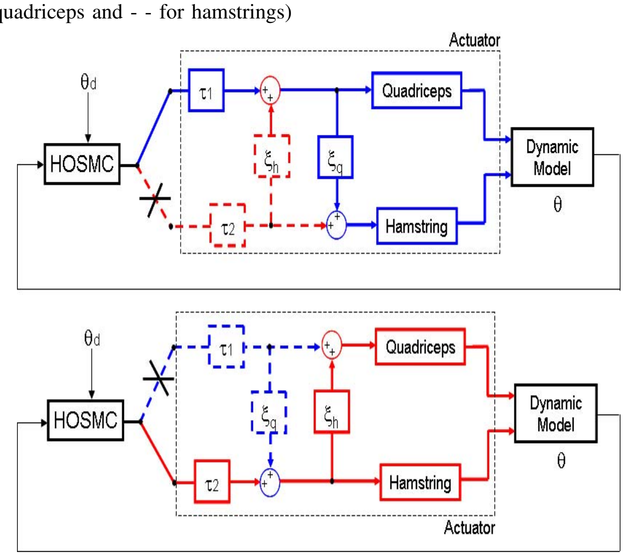

Fig. 5. Co-contraction of both muscles

$$If(u>0) \Rightarrow \begin{cases} I_q & = & \frac{u}{u_{nom}} I_{Max} \\ I_h & = & \xi_q I_q \end{cases} \quad (14)$$

$$If(u<0) \Rightarrow \begin{cases} I_q & = & \xi_h I_h \\ I_h & = & \frac{u}{u_{nom}} I_{Max} \end{cases} \quad (15)$$

u_{nom} and I_{Max} correspond respectively to nominal value of the control u and the maximal value authorized to stimulate a muscle (around 200 mA). The stimulation current values for quadriceps I_q and hamstrings I_h and the Pulse Width magnitude, PW_q and PW_h respectively, enable us to evaluate the required ratios of fibers to be recruited (α_q, α_h). The chemical inputs u_q and u_h of the muscles are automatically activated when the electrical currents are respectively superior to zero. We have implemented this algorithm on a simulator built with Simulink^{TM} software. We applied two different desired positions, starting from the rest position, $\Theta_d = 90°$ as:

$$\begin{cases} 1) & 1s < t < 4s: & \Theta_d & = & 130° \\ 2) & 6s < t < 9s: & \Theta_d & = & 50° \\ 3) & Otherwise: & \Theta_d & = & 90° \end{cases}$$

The coefficient of the 2-sliding controller were chosen to verify the condition equations (*9*). The following values have been used: $\lambda = 10$, $\lambda_1 = 20$, $\rho = 0.7$, $V_M = 1$. The simulation sampling period was set to 10^{-3} sec. $Fig.6(a)$ shows the step response for different desired angles. We notice in $Fig.6(b)$ the finite time convergence of the sliding surface about *1sec* in knee flexion and extension. Desired and current angle curves match when sliding surface reaches zero. We show the resulting simultaneous stimulation currents for quadriceps and hamstrings I_q and I_h in $Fig.7(a)$,. The control vector u computed by the equation (*13*) is presented in $Fig.7(b)$. If the resulting control variable u is positive (respectively negative) the quadriceps (respectively hamstrings) is stimulated and controlled in a closed loop as shown in ($Fig.5$), while its antagonist muscle is controlled in an open loop control scaled by ξ_q (respectively ξ_h). The resulting control variable u was scaled, according to its sign by τ_1 (respectively τ_2) to the normalized amplitude of the current stimulation. As it is shown in $Fig.8(b)$ we have added, for a short time, a double force perturbation, representing an external disturbances which may occurred during any task performed by the lower extremities of the paraplegic patient. The amplitude of these perturbations were for the hamstrings $10N$ at $2.8s$ and $7N$ at $4.4s$ representing respectively 23% and 16% of the maximal force. For the quadriceps the perturbations were $50N$ at $7.8s$ and $30N$ at $9.4s$ representing respectively 25% and 15% of the maximal force. In the presence of these relative high perturbations, we can notice a high capability of position tracking.

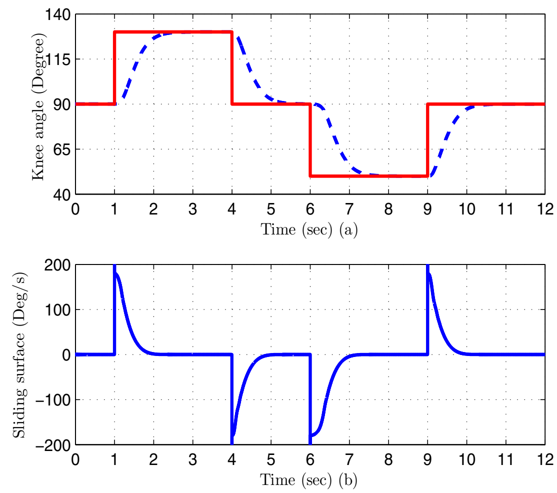

Fig. 6. a) Desired step and actual knee angle variation, b) stabilization of the sliding surface

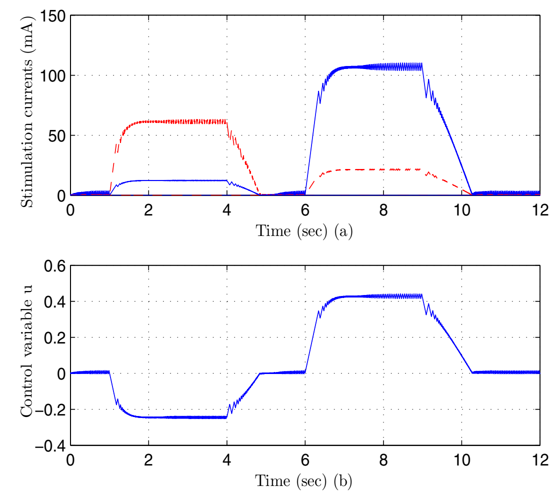

Fig. 7. a) Stimulation current (.. I_h, - I_q) $\xi_q = \xi_h$, b) The resulted control vector u

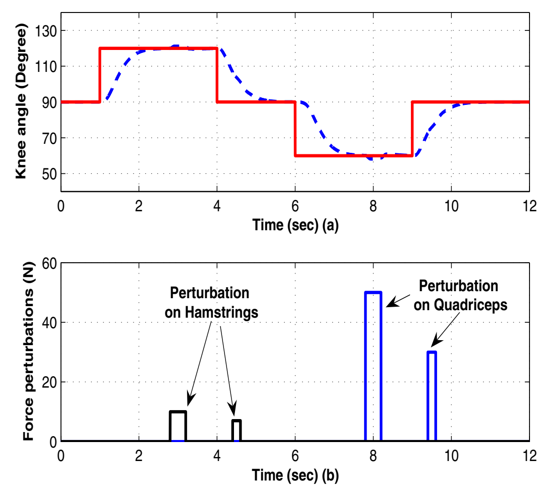

Fig. 8. a) Desired step and actual knee angle variation, b) Force perturbation added to the generated muscle forces

V. CONCLUSIONS AND FUTURE WORKS

A. Conclusions

Few studies have treated the human muscle as an entire physiological element in a closed loop system. Known by their robustness against unknown perturbation and their accuracy, we used the sliding mode control. Because of the nonlinearity and the presence of a 2 relative degree order system, we adopted in the current study, a high order sliding mode controller *HOSM*, which seems necessary to ensure a robust control and a safer movement of the lower extremities. This later was applied to a new multi-scale model developed within the *DEMAR* project. The muscle model is based on internal physiological characteristics assembling two levels: the microscopic one, involving the sliding actin-myosin filaments and the macroscopic part represented by a contractile element and an elastic element. We were able to control two antagonist muscles quadriceps and hamstrings alternatively and simultaneously (the so called co-contraction effect) with the same control vector, increasing the joint stiffness and forcing dynamically the system to behave as a first order response. The co-contraction was controlled based on a static optimization of the sum of the stress in the antagonistic muscles. Satisfactory stability and tracking error were achieved after a finite time delay. The performance of the closed loop system has been assessed in the presence of an external force perturbations. The controller has shown a great accuracy and robustness against these perturbations.

B. Future Works

In order to validate simulation results further work will be carried out with paraplegic patients to evaluate the accuracy and the robustness of the high order sliding controller. Experiments are ongoing to validate the 2-sliding controller by using a multi-moment platform on a paraplegic patient. Further study of the amount of co-contraction predicted will be detailed not only in isometric conditions but also through a range of movement and will be identified through *EMG* Electromyography measurements. Other control strategies will be implemented as well to analyze the relevance of the muscle model in much complex situations.

VI. ACKNOWLEDGMENTS

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