2010 IEEE International Conference on Robotics and Automation
Anchorage Convention District
May 3-8, 2010, Anchorage, Alaska, USA

Tremor Attenuation Using FES-based Joint Stiffness Control

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Abstract— **In this paper, a strategy to attenuate tremor based on co-contraction of antagonist muscles using Functional Electrical Stimulation (FES) is fully presented. Both methods to track tremor features in real-time, while filtering voluntary motion, and to identify a suitable joint model are described. Using this information, the stimulation controller modulates joint stiffness based on tremor intensity, while preventing the generation of undesirable joint torque. An experimental evaluation of the system, which confirmed the effectiveness of the approach, is also presented.**

I. Introduction

Tremor is the most common movement disorder found in human pathology [1]. It is not a life-threatening pathology, but it often decreases considerably the person's quality of life. Patients with tremor, defined as an involuntary, approximately rhythmic and roughly sinusoidal movement, present reduced ability to perform simple daily tasks, such as drinking a glass of water or opening a door.

An absolutely effective treatment for pathological tremor is not yet available, since current pharmacological and surgical alternatives still present limitations with respect to cost, risks, and effectiveness. A different approach is the use of assistive technologies, such as robotic devices [2], upper limb exoskeletons [3], and the use of Functional Electrical Stimulation (FES) [4].

A robotic system built to attenuate the effects of tremor must present several features. It must be able to distinguish between voluntary and pathological motion and to react to changes in the trembling motion, since tremor often presents highly time-varying dynamics. Also, the system must attenuate the effects of tremor while minimizing the induced fatigue, pain, and discomfort. Based on these requisites, the present work was conducted with the long term goal of evaluating the use of surface FES for the pathological tremor compensation on the upper limb.

In his pioneer work, [4], [5] proposed a closed-loop FES-based tremor compensation system in which a pair of antagonist muscles were stimulated in order to contract out of phase with respect to tremor. The controller was designed in such a way that the closed-loop response of the system was maximized for the tremor frequency. A simple model of joint motion due to FES was used, but the possibility to provide long-term tremor suppression using electrical stimulation was demonstrated.

In this work, however, we are not especially interested in controlling joint motion to counteract tremor. Instead, we are mainly interested in exploring other features of FES-controlled muscles for this purpose. Particularly, we evaluate the use of electrical stimulation to modulate joint stiffness by co-contracting a pair of antagonist muscles, while producing minimum joint displacement. This additional joint stiffness may not only reduce tremor amplitude, but also provide an extra stability to support the person's intended motion.

Such a system may be designed with different levels of complexity. For instance, the stimulation levels that provide a suitable stiffness for that particular tremor may be manually set. An improved closed-loop solution would require the addition of other features. Firstly, since tremor often presents time-varying dynamics, online tracking of tremor severity may be used to estimate the required additional joint stiffness. Also, in order to allow accurate modulation of joint stiffness, a proper identification of the joint dynamics model is needed.

In this paper, both problems are addressed. In the following section, an online tremor characterization algorithm, which also filters voluntary motion, is described. The algorithm, originally presented in [6], is based on the Extended Kalman Filter (EKF) and concurrently estimates both tremor and voluntary motions, as well as the tremor parameters, represented by a nonstationary harmonic model. The tremor attenuation strategy, about which a prospective simulation study has been presented in [7], is described next. Section III presents the joint model, particularly suitable for the current application, as well as the identification procedure. Next, the implementation of a simple controller to attenuate tremor based on these concepts is described. The wrist joint is chosen due to the high incidence of tremor and the functional importance of that joint. Section V presents the experimental setup, represented on Fig. 1, and the performance of this compensation strategy in subjects with no neurological impairment, but under the effects of a FES-induced tremor. Final remarks are presented in the last section.

The concepts and techniques presented here are not exclusive to tremor compensation. They may be applied in the field of artificial muscle control and within problems of human-robot interaction, such as rehabilitation robotics.

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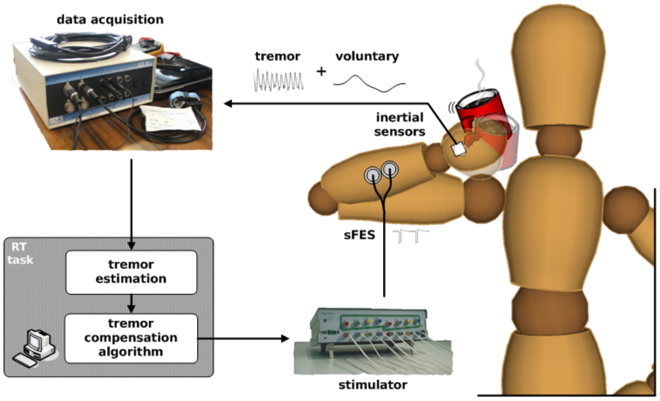


Fig. 1. Experimental setup used to evaluate different pathological tremor attenuation strategies. Each represented element is described throughout the paper.

II. Online Tremor Tracking

This section concerns the online characterization of tremor, but considering that the sensor used for that purpose also measures the intentional motion performed by the person, i.e.,

$$s = s_t + s_v + \nu_s, \tag{1}$$

where s_t is the tremor component, and s_v is the voluntary motion. s is measured by a motion sensor, which may be an inertial sensor, an optical tracking system, digitizing tablets or similar devices. ν_s is an additive white Gaussian noise, $\nu_s \sim N(0, \sigma_s^2)$, that represents sensor error.

Different solutions [8], [3] have already been proposed to accomplish the online estimation of both tremor and voluntary motion components from the measurements of a noisy sensor. Here, we briefly present a method which was originally presented in our previous work [6], where a further presentation of this problem may be found.

In our approach, both tremor and voluntary motions are modeled as nonstationary signals. Since tremor may be seen as a quasi-periodic motion, an harmonic model has been chosen to represent it:

$$s_t = \sum_{h=1}^{H} \left[a_h \sin\left(h \int \omega d\tau \right) + b_h \cos\left(h \int \omega d\tau \right) \right] + \nu_{s_t}, \tag{2}$$

where ω is the time-varying fundamental frequency, a_h and b_h are the coefficients and H is the number of harmonics, the model order. ν_{st}, an additive white Gaussian noise, $\nu_{s_t} \sim N(0, \sigma_{s_t}^2)$, represents modeling errors. ω, a_h, and b_h are time-varying parameters.

Regarding voluntary motion, although it is a slower movement, it does not present the regular features of the tremor motion. Hence, it was modeled as low-pass filtered white noise, with fixed filter parameters tuned to represent the low frequency behavior assumed for voluntary motion.

To estimate concurrently both tremor and voluntary motions and the tremor model parameters, a Kalman filter (KF) is used. Since the problem is nonlinear, one alternative is to use a modification of the KF for nonlinear systems, the EKF, where Kalman equations are applied to the first-order linearization of the nonlinear system around the current state estimation [9].

The filter states are composed by those states related to tremor, $\mathbf{x}_t$, which are composed by the estimated motion and the tremor model recursively identified parameters,

$$\begin{bmatrix} s_t(k) & a_1(k) & \cdots & a_H(k) & b_1(k) & \cdots & b_H(k) & \omega(k) \end{bmatrix}^T$$

and the states related to voluntary motion, $\mathbf{x}_v$, which are organized accordingly,

$$\begin{bmatrix} s_v(k) & \cdots & s_v(k-F+1) & \nu_{s_v}(k) & \cdots & \nu_{s_v}(k-F+1) \end{bmatrix}^T$$

where F is the order of the voluntary motion filter and k is a multiple of T, the sampling period.

Within our EKF framework, uncertainties from the model, the parameters (modeled as random walks) and the measurement are explicitly considered, increasing the estimation robustness. All the parameters and initial estimates used to configure the proposed recursive algorithm are available in [6].

Once the tremor model parameters are estimated for every time-instant, tremor power or intensity may be computed directly from the coefficients of the harmonic model:

$$P = \sum_{h=1}^{H} \frac{\|b_h - i a_h\|^2}{4}. \tag{3}$$

III. FES-controlled Joint Stiffness

A general model to describe a single joint with one degree of freedom actuated by a pair of FES-controlled antagonist muscles, disregarding external forces, is:

$$J\ddot{\theta} = -M_p(\theta, \dot{\theta}) + M_g(\theta) + M_f(u_f, \theta, \dot{\theta}) - M_e(u_e, \theta, \dot{\theta}), \tag{4}$$

where θ is the joint angle, J is the inertia for flexion/extension motion, M_p is the sum of internal passive moments on the joint which are independent from muscle action, M_g is the moment due to gravity, M_e is the moment due to external forces, and M_f and M_e are the moments resulting from flexor and extensor muscles actions, respectively. u_f and u_e are the normalized stimulation inputs to the muscles.

A. A simplified model

In our previous work [7], a detailed version of (4) was applied to demonstrate in simulation the modulation of joint stiffness during co-contraction of FES-controlled antagonist muscles. The model included several nonlinear effects related to muscle contribution to joint dynamics, such as Hill-based muscle dynamics, proprioceptive feedback, nonlinear moment arms, and others.

In the current work, the interest is to apply this effect to experimentally demonstrate tremor attenuation. In this scenario, different inter-subject and intra-subject complexities are added due to the use of surface FES in practice. For instance, small differences in the electrodes positions highly affect the response obtained. It is even a greater issue if we consider electrical stimulation diffusion to other muscles. Together, these effects prevent the application of same model parameters on different experimental sections. Other time-varying effects may interfere also within a single experiment, like changes in skin-electrode interface, muscle fatigue induced by FES, and others.

Furthermore, detailed models of musculoskeletal dynamics often require elaborate identification protocols. However, the goal of this project is a technology that will be used by tremor patients in a regular basis. Hence, the use of simpler models, which allow faster identification procedures with fewer and simpler sensors is preferred. Moreover, our interest is not in the validation of a FES-controlled joint model, but to use a model that may be applied for tremor attenuation strategies, which also justifies the choice.

Based on these aspects, the following simplified model was chosen to describe joint dynamics:

$$\begin{aligned} J\ddot{\theta} = -B_p\dot{\theta} - K_p\theta + K_g\cos(\theta) + \\ M_f u_f - M_e u_e - K_a(u_f + u_e)\theta, \end{aligned} \tag{5}$$

where B_p is the passive damping, K_p is the passive stiffness, M_f and M_e are equivalent to the maximum provided moment, and K_a is the active joint stiffness provided by the pair of antagonist muscles. K_g is a single parameter that represents the maximum moment due to gravity acceleration.

This model is particularly suitable for our application, since here we are mainly interested in estimating steady-state effects of FES in joint dynamics. Of particular interest are the set of stimulation pairs that produce no torque on the joint, $M_f u_f = M_e u_e$, and the contribution of FES-controlled co-contraction of antagonist muscles to joint impedance. In this model, only the active stiffness, $K_a(u_f + u_e)\theta$, is considered, which further simplifies the identification procedure. A similar model was proposed in [10], a classic work in the domain.

B. System identification

Procedures for identification of FES-controlled muscle models are often divided in two distinct phases [11], [12]. In a first moment, the passive parameters of the model are identified. Then, the different parameters related to FES activation are estimated. In this work, we have applied a similar procedure.

For both identification steps described in this Section, if the procedure is applied to a tremor patient, the direct use of the measurements from the motion sensor may deteriorate the results. In that case, the voluntary motion s_v estimated by the algorithm presented in the last section may be employed.

1) Passive identification: The first phase in the proposed identification procedure is the estimation of the passive parameters in Eq. (5), i.e., J, B_p, K_p, and K_g. Those are estimated with the passive pendulum test, in which the limb is completely relaxed and the resultant motion due to gravity is used to identify the respective parameters.

As in Section II, an EKF is used as the algorithm to estimate the model parameters. The state vector is:

$$\begin{bmatrix} \dot{\theta}_{(k)} & \theta_{(k)} & J_{(k)} & B_{p(k)} & K_{p(k)} & K_{g(k)} \end{bmatrix}^T.$$

Since the model states ($\dot{\theta}$ and θ) are also estimated within the filter, a relatively similar framework, if compared to the one used on Section II, may be used here. Considering that some of the same sensors used in the previous section may be applied to measure or estimate the joint angle, part of the parameters to configure the filter may also be the same. Regarding the algorithm initialization, the performance of the identification may be highly influenced by the initial parameters estimate. We have employed the parameters from [13] as initial estimates.

2) Active identification: The major challenge in the identification procedure, especially during active parameters estimation, is the disturbance caused by involuntary contractions performed by the subject. Most of the literature concerning FES-controlled muscle model identification do not share this problem with the present work, since it is concerned with motor restoration of spinal cord injured patients. In our case, however, the resultant motion is highly affected by any muscle action not caused by the stimulation, particularly for short FES-induced joint motion. Furthermore, these effects are hardly absent both for subjects who are already familiar with FES and those who have not yet experienced it (i.e., it is difficult to relax completely and do not react to the stimulation).

In this scenario, which also persuaded us to choose a model that neglects active damping, the active parameters identification was designed to minimize the disturbances discussed above. The procedure is based on the application of a pseudorandom sequence of stimulation pairs. Since it is hard to capture completely the dynamic properties of such a system, only the steady-state joint angle is applied for the parameters estimation.

In this identification procedure, the joint axis was positioned parallel to gravity, and hence the steady-state response of the system represented by Eq. (5) is:

$$-K_p\theta + M_f u_f - M_e u_e - K_a(u_f + u_e)\theta = 0, \tag{6}$$

where K_p have been already estimated in the passive identification procedure. To estimate the remaining parameters in the equation above, we applied the Gauss-Newton method alternately to estimate the linear parameters (M_f and M_e) and K_a. The alternate estimation and the initial parameters were chosen in order to prevent the algorithm convergence to parameters with no physical meaning ($M_f, M_e, K_a < 0$).

IV. Tremor Attenuation

The information obtained so far may be used to design a tremor attenuation system. In this work, a simple controller was implemented in order to validate the concept. Still, some important aspects that greatly interfere with the success of the compensation strategy must be detailed.

The first aspect that must be pointed out is that the controller closely interacts with the subject. The final goal is not to completely suppress tremor (the case in which maximum joint active stiffness would always be the best control action), but to provide the greater functional benefit, while minimizing total discomfort. Quantifying the functional assistance is a difficult task, but at least it may be argued that the tradeoff lies between tremor amplitude reduction and the supply of additional joint stability.

The controller evaluated in this paper may be seen as a regulator designed to reject an estimated disturbance (the tremor). Particularly, we have designed a simple PI controller with anti-windup, while this last feature is due to the actuator saturation with respect to physiological limits and subject comfort. The controller error is P_t, the tremor severity estimated by the online tremor tracking algorithm. The control law is implemented for a particular controlled muscle and the correspondent antagonist muscle input is given by

$$u_e = \frac{M_f}{M_e} u_f, \tag{7}$$

to ensure no residual torque will be applied.

In this context, such a controller must be able to handle the following situation: since the controller error is always positive (there is no $P_t < 0$), a pure PI regulator will wrongly provide additional joint stiffness if tremor decreases sufficiently. For instance, it is preferable that stimulation ceases if tremor stops. However, it is not what happens with a simple PI, since there is no negative error to minimize the integral action.

The solution implemented in this paper to avoid this problem is to suspend the stimulation if the tremor severity drops below a predefined threshold. If a measurement of the trembling muscle activity was available (from surface electromyography, for instance), a separate estimate of tremor intensity could be computed and such routine would be unnecessary.

V. Experimental Evaluation

This section presents the experimental setup applied to evaluate the methods proposed in this paper, as well as the results and a brief discussion.

Since our effort to conceive a tremor compensation system using FES is still in its early stage, the experiments are performed on subjects with no neurological impairment, where tremor on the target joint was induced by an independent electrical stimulator. This is a new approach, even if tremor compensation methods were already validated on subjects performing voluntary tremor around pathological tremor frequencies [5].

A. Setup

The experimental setup may be represented by the Fig. 1. The main components are the stimulating, sensing and processing units. The stimulator is an 8-channel stimulator, the Prostim, designed jointly by the LIRMM and Neuromedics. It allows independent control of each channel through a serial interface. The main sensor used in the control loop is the IDG-300, an angular rate sensor from Invensense, whose measurements are acquired by a National Instruments acquisition card. Both tremor tracking and tremor attenuation algorithms are executed in a 50 Hz-loop running in a Linux system. The whole system is electrically isolated to ensure the subject's safety.

As additional hardware, a commercial stimulator from CEFAR, the Physio 4, was used as the stimulator to induce tremor. It produces biphasic square pulses, opposed to the biphasic pulses with capacitive discharge generated by Prostim.

B. Subjects and FES normalization

Two male subjects participated in the experiment. In both cases, the target joint was the wrist, particularly the dorsi/palmar flexion. Two antagonist muscles were chosen to compose the pair of antagonist muscles to be controlled, the *Flexor Carpi Ulnaris* (FCU) and the *Extensor Carpi Ulnaris* (ECU). Even though these two muscles also produce residual ulnar deviation, they were selected due to the clearness and strength of the response (obtained with round 3.2 cm electrodes). FES-induced tremor was produced by stimulating with larger electrodes (round 5 cm electrodes) both the *Flexor Carpi Radialis* (FCR) and the *Palmaris Longus* (PL).

Due to inter and intra-subjects variations concerning electrically controlled muscles, every experimental session was preceded by a procedure to identify the appropriate stimulation parameters for each muscle. Considering that the stimulator applied in this work allows online update of all three traditional FES parameters, the following rules were applied:

- Frequency was fixed for every experiment (30 Hz).
- Amplitude was also fixed, but its value was chosen in an initialization procedure. Typical values ranged from 15 to 25 mA.
- General stimulation level was controlled by the stimulation pulse-width (PW). Minimum (PW_{min}) and maximum (PW_{max}) values obtained according to the person's subjective evaluation were used to normalize FES control:

$$PW = (PW_{max} - PW_{min})u + PW_{min}, \tag{8}$$

where u, $0 < u < 1$, is the control variable.

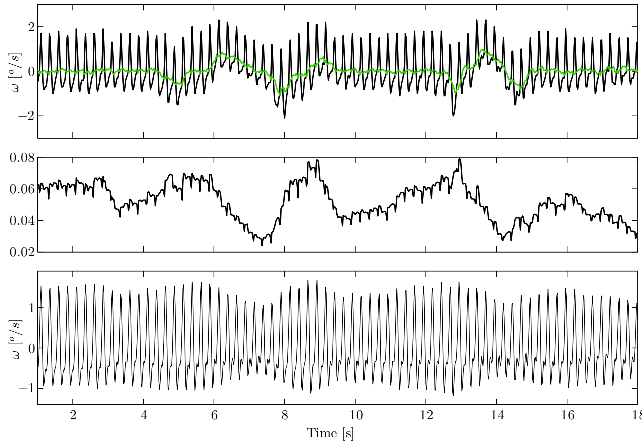


Fig. 2. Estimated tremor (s_t, bottom) and voluntary motion (s_v, top), as well as tremor intensity (P_t, middle), from FES-induced tremor recorded with angular rate sensor.

TABLE I

Wrist parameters estimated with the proposed identification procedure.

J	B_p	K_p	K_g	M_f	M_e	K_a
0.0022	0.020	0.107	0.238	0.282	0.101	0.357

C. Results

In this subsection, results from the methods proposed in this work are presented.

The performance of the tremor tracking algorithm presented in Section II is illustrated in Fig. 2. FES-induced tremor was produced using the CEFAR stimulator and the wrist angular velocity was measured as in Fig. 1 with an angular rate sensor. With a constant stimulation signal (4 Hz frequency, 15 mA amplitude, and 250 μs pulse-width), light voluntary movements were performed to demonstrate the algorithm ability to discern between intentional and artificial motion. Its performance on pathological tremor patients has been presented in [6].

Regarding the system identification, the passive phase was performed in only one subject, while the active part was performed at each experimental session. Using the passive parameters from one subject for another subject was an attempt to verify if the identification procedure could be further simplified. Table I illustrates the parameters from Eq. (5) estimated for the subject A.

The tremor compensation algorithm was evaluated in series of trials, where the subjects also executed light voluntary movements. To allow brief comparison of performance on different situations, tremor of distinct intensities were induced on the two subjects. FES-induced tremor frequency was constant during the tests. Tremor attenuation based on the acquired data was estimated using the tremor intensity ratio between the trials where the stimulation was active or not. Two different methods to estimate tremor intensity were used. One approach was to apply the tremor intensity P_t estimated with our algorithm and the other was adapted from [14], where a method based on the power spectrum of the acceleration of movement around the tremor frequency is proposed. The obtained results are shown on Table II. Figure 3 illustrates two samples of the system performance.

TABLE II

Reduction on tremor intensity computed with two distinct techniques.

Subject	**Tremor Amplitude**	**Trials**	**Attenuation [%]**	
			[6]	[14]
A	High	3	85.79	70.87
B	Low	3	49.75	73.86

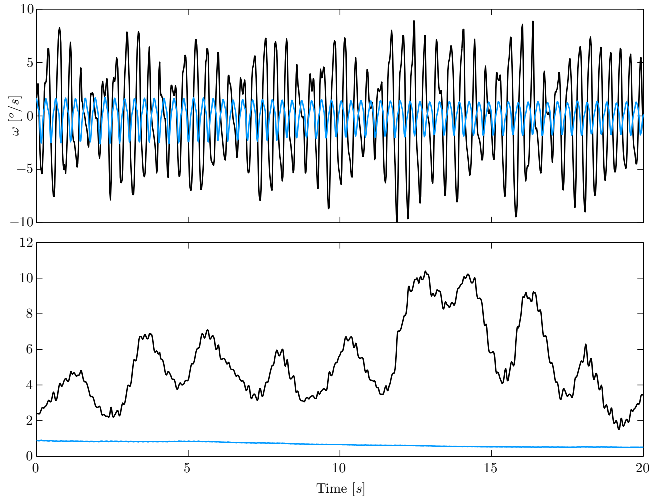
The results presented on Fig. 3 also allow indirect evaluation of the other proposed methods. Tremor tracking algorithm is demonstrated by the tremor intensity estimation, while the identification procedure is validated by the absence of residual torque produced by the stimulation, which would generate undesirable joint motion.

D. Discussion

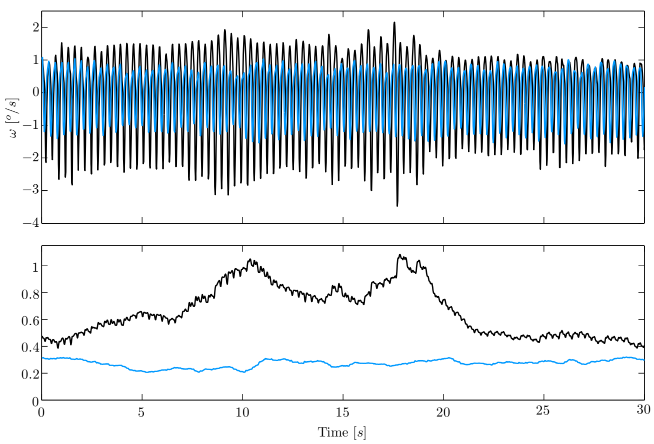
The results presented in the last subsection illustrate the performance of the full system with respect to the proposed task. The presented data have demonstrated the feasibility of using FES-controlled co-contraction of antagonist muscles in order to attenuate the effects of tremor. The performance evaluation of the system, in order to verify which parts need further improvement, is a harder task.

The main measure to analyze the system performance is the reduction of tremor intensity, shown on Table II. However, a satisfactory performance within this parameter requires that all parts of the solution work properly. Some of those individual methods have been evaluated independently, like the tremor characterization algorithm [6]. Regarding the identification and control approaches, only brief evaluations were conducted within this paper. Particularly concerning the controller, a systematic procedure to optimize its performance was not performed, since the main goal of the paper was to demonstrate the feasibility of the proposed system.

It was also not in this scope of this paper a comparison with other strategies of tremor compensation using FES. The absence of clear protocols and methods to quantify tremor severity also contributed in this direction. Nevertheless, some general comments about the proposed compensation strategy may be discriminated. First, no objection concerning pain caused by the stimulation was made by the subjects that took part within the research protocol. Furthermore, while our identification procedure provided the FES contribution to joint stiffness, we could not confirm if other aspects of active impedance were not affect by the stimulation (like damping, which could play an important role in tremor attenuation). Lastly, two benefits of the proposed approach may be pointed out:



(a)



(b)

Fig. 3. Measured wrist angular velocity (ω, top) and estimated tremor intensity (P_t, bottom). In black, data of FES-induced tremor with no compensation, and in blue the attenuated data. Data from subjects A (a) and B (b) is shown.

the additional stability provided to the joint and a less varying stimulation signal, which causes less discomfort to the subject.

VI. Conclusions and Future Works

Attenuating the effects of pathological tremor using electrical stimulation is a complicated task. Tremor often presents time-varying dynamics and muscle command using FES is complex, particularly with surface electrodes. More important, the system must be designed to provide functional support for the patient, with unconditional safety and reasonable comfort.

In this context, this paper presented the complete design of a system capable of providing tremor attenuation using FES-controlled co-contraction in order to increase joint stiffness. Such a system is currently able to estimate tremor features in real-time, while filtering the components from voluntary motion. The identification of a suitable joint model actuated by a pair of antagonist muscles has also been accomplished. Finally, a controller that modulates the active joint stiffness provided to the joint with no residual torque was implemented and validated experimentally.

Our future efforts include further development of selected methods in our current system, such as an expansion of the identification procedure and the design of more sophisticated controllers. Also, evaluation of the current strategy among tremor patients is scheduled.

VII. Acknowledgments

The authors would like to thank the subjects that participated in the research and also the French National Research Agency.

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