CEAI, Vol. 8, No. 3, pp. 15-22, 2006 Printed in Romania

A ROBUST ADAPTIVE AUTO-TUNER FOR ARTIFICIALLY STIMULATED MUSCLES

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Abstract: *Natural biological control involves the normal functioning of the living organism (i.e. human body) to regulate its parameters such that the vital functions are kept within the normal operating range. When this natural control fails, the biological feedback becomes unstable, operating under non-optimal conditions of the subject's vital capacity. In this context, ensuring the subject's nominal surviving capacity requires artificial control of the vital functions. Nowadays technology enables the development of artificial closed-loop devices to correct and provide the normal functions of the organism, thus replacing the damaged parts or helping to recover their natural properties. These procedures are called rehabilitation techniques. By applying electrical impulses to the appropriate muscles, control of paralyzed limb muscles can be achieved. A robust self-adaptive auto-tuning control strategy is proposed for correction and rehabilitation of paralyzed muscles. Re-tuning of the controller parameters with respect to changes in muscle parameters preserves the closed-loop stability.*

Keywords: *adaptive control, biomedical systems, simulation, PI controller, robustness*

1. INTRODUCTION

The field of biomedical engineering is relatively young compared to that of control and automation and is one of the most remarkable in terms of interdisciplinarity. It has been known for many years that the biological world contains many feedback mechanisms and structures [8,9]. Consequently, a manifold of applications of classic and advanced control strategies to biological systems have been defined. Scientists applied their knowledge of mathematical modelling and system analysis to various areas of bio-medicine and identified those physiological functions not yet described in mathematical terms. This knowledge-information created a *playground* to control engineers. However, only nowadays has their work become meaningful, offering practical solutions to modern control problems.

The goal of this contribution is to give a practice-oriented overview of one of the *hot-spot* in biomedical control applications: artificial limbs. Such systems are usually nonlinear, constrained and their parameters vary in time. They are also highly dependent on developments in instrumentation and have a significant impact on the patient's life and consequently in society. Performance analysis of these systems is provided by control engineers as well as by physicians.

The proposed application is functional electrical stimulation control of limb muscles in paralyzed patients. The general principles of artificial electrical stimulation of neural tissue and available models are briefly introduced in section 2. Further-on, an auto-tuning technique is described in section 3 by a direct adaptive control (DIRAC) strategy implemented in a discrete manner to simulate real-life practical implementation aspects. The simulation is made on a nominal 2^{nd} order plus time-delay model of the skeletal muscle and the controller performance is analysed in section 4. Muscle parameter variations are also discussed. Finally, some conclusions summarise this investigation.

2. ELECTRICAL STIMULATION

Artificial electrical stimulation of neural tissue can be used as a neuroprosthetic technique to replace lost functions of the body. It is often referred to as functional electrical stimulation or, in particular, when used to stimulate neuromuscular tissue, as functional neuromuscular stimulation. Good overviews on this topic can be found in the work of Winters and Woo [13] and Stein et al. [12].

In order to apply artificial electrical stimulation to muscle, it is essential to take some basic physiological properties of the neuromuscular system into account; an introduction to this can be found in standard physiology textbooks [5]. In skeletal muscle, extrafusal muscle fibres are the primary unit of contraction. They are activated by axons of α-motoneurons, which originate in the spinal cord. In the neural system, transmission of information takes place in the form of impulse trains; the information is encoded in the pulse frequency. One motoneuron activates 5-1000 muscle fibres simultaneously. All fibres activated by the same motoneuron can be distributed over the entire muscle and form a motor unit, which represents the *unit* of muscle force in a normally innervated muscle.

Artificial muscle activation (functional electrical stimulation FES) can take place by applying electrical impulses to the motoneurons, thus generating action potentials which are transmitted to the corresponding muscle fibres. Thus, with FES, muscle activation can be varied by (a) the energy of the electric pulse, which defines the number of recruited motor units, and (b) the pulse frequency, or the inter-pulse interval (IPI), which determines the contraction of the recruited muscle fibres [11].

2.1 Muscle Models

The range of models extends from biophysical models, which are based on the structure and mechanisms of actual muscle, through analogue models, to purely mathematical descriptions. The most widely used biophysical model is the cross-bridge model, the basic principles of which were developed by Huxley [7]. This type of model is in principle useful to describe all characteristics of the muscle. In biomechanics, analogue models remain more popular. An example is Hill's model [6], which comprises a contractile element (the force generator), in parallel with a spring (muscle elasticity), all in series with a second spring (passive tissue).

For use in implantable FES devices, other model characteristics become important. The controller should be *adaptive, taking into account changes in muscle parameters occurring during rehabilitation.* It should also be computationally suitable for implementation as a microcontroller. Therefore, most real-world implementations are based on simple linear muscle models.

The contraction is assumed to be isometric, which gives a single-input (stimulation) single-output (force) system. Experiments have shown that a second-order dynamic linear model with delay can be applied [4,14] and make the topic of our investigation.

2.2 FES Technology.

There are a manifold of application considering FES for limb muscles, from hand grasping to drop foot correction and general locomotion inability. Developing methods for implementing FES is a highly interdisciplinary topic, involving experts in neuroscience, biomechanics, muscle physiology and control engineering. To capture the information transmitted by healthy neurons within the nerves of interest and to code it in been developed by specialized research units and they represent a key-component in the artificial control-loop of limb muscles.

Once the connection with the nervous system has been successfully established, a neural prosthesis can be implemented. An example of such a control-loop is given in Figure 1. Suppose that the muscles to be controlled are intact, but only the nervous activation link has been broken by injury. A connector with its electrode will be surgically inserted in the hand, at the location of the nerve responsible for controlling the desired muscle area. The connector also plays the role of the implanted receiver/stimulator of electrical impulses. At the end of this connector, there are flexible silicon leads. To stimulate the nerve very short pulses of electricity are passed through each electrode (each electrode can stimulate one muscle). The nerve impulses are traveling down the nerve to the muscle in the same way as the naturally occurring nerve impulses. Power to run the stimulator is passed through the skin using radio waves from a small control box strapped to the outside of the hand or on the shoulder of the patient. Feedback information is ensured using force gauges on the tip of the finger. The user (doctor) only has to set the stimulation intensity using two control buttons.

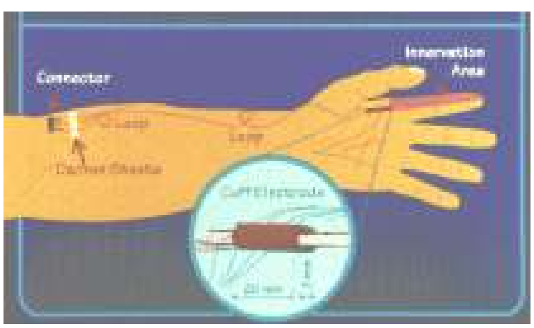

Fig. 1. Example of implantable FES device in hand

3. SKELETAL MUSCLE CONTROL

When considering the physical requirements of the control system, a set of time-domain specifications will result. In the task of interest something which is meaningful to a controller is a very complex process. Such connectors have here - locomotion, the desired activation of the muscle (force) can be described as a stepwise signal (set-point). The amplitude of the electrical activation will be constant, as well as the impulse period (unit impulses). The desired force will be obtained with a variable frequency of the electrical stimulus.

The actual activation of the muscle, as achieved through feedback control, is required to meet the following specifications:

(i) a reasonable rise time,
(ii) no overshoot, and
(iii) zero steady-state error.

In addition, this performance and closed-loop stability should be robust to expected variations in the controlled muscle dynamics – since they are subject to rehabilitation techniques.

A DIRAC (DIRect Adaptive Control) control strategy is the subject of this feedback-control application study. The DIRAC algorithm is both an auto-tuning as well as an adaptation method for the controller parameters. Since paralyzed or paretic muscles are time-varying systems, an adaptive/auto-tuning method is therefore justified. A simple simulation on a muscle model adapted from the literature is performed and the controller performance for reference tracking is depicted. A comparison between a classical control approach (PI and PID) and a DIRAC-PI controller is given in [10]. However, the study in [10] tackles some implementation aspects of a 2nd order model, *without considering time-delay.*

3.1 Linear muscle model.

Using experimental data, Gollee and Hunt [4] extracted the impulse response from which the step response was derived by integration. For linear dynamic systems, the dominant time constants can be extracted from its step response. Based on this information, estimates for the optimal sampling period and the model structure are possible.

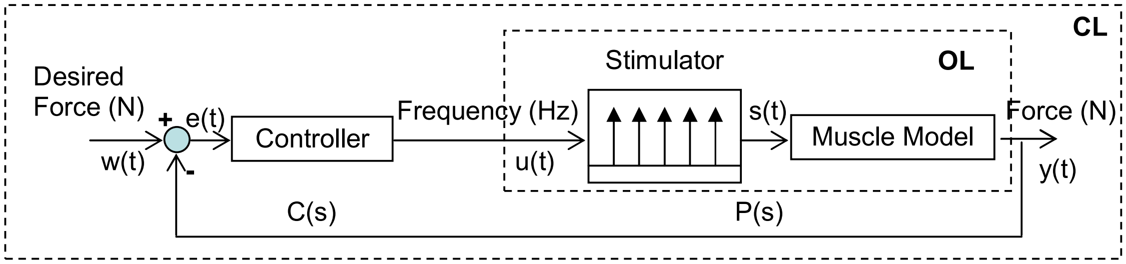

Fig. 2. Block scheme of muscle model simulation (open loop OL and closed loop CL configuration)

A linear model capturing the properties of a muscle under isometric conditions [3] can be represented by a 2nd order transfer function with time-delay:

$$M_{\text{nominal}}(s) = M_1(s) = \frac{450e^{-0.005s}}{s^2 + 25s + 100} \qquad (1)$$

The parameters in (1) are a set of *nominal* parameters and their value can change (considerably) from person to person. They also depend on the physical condition of the muscle, age *etc*. In the case of a paralyzed muscle, important variations are observed during the rehabilitation period. The time delay (5ms) has been taken into account and is the time between the nervous activation and the calcium release in the muscle in order to obtain contraction.

The input to such a model is a stimulus which occurs with a certain frequency and the output is the force resulted from the contraction of the muscle, as in Figure 2. The input frequency is limited to 5Hz – 50Hz (the frequencies for which the static characteristic is linear).

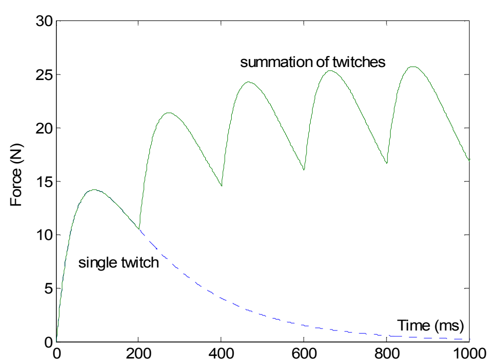

Fig. 2. Single Twitch (dotted) and Summation of Twitches (continuous) at 5Hz (unit) impulse frequency in open loop.

From a mathematical standpoint, the response *y(t)* of the process *P*(*s*) (here: the stimulus and the muscle model) to an input $u(t)$ (= frequency *F*) is the effect of a series of impulses with period *T* (= *1/F*) applied as input. A simple experiment (unit impulses) depicted by Figure 3 shows the output of the process corresponding to (1) and having the time constants: 1/20=50 ms and 1/5=200 ms, a total open-loop settling time of 1.1 second.

3.2 A DIRect Adaptive Controller (DIRAC).

Controller design is based on the use of the auto-tuning principle, which automatically finds a set of PI(D) parameters without an *a priori* process identification (i.e. no model required). A brief description is provided in this section and more details can be found in [1].

The PI(D) parameters can further be used in a discrete-time control scheme, with a software implemented controller:

$$u(t) = u(t-1) + c_0 e(t) + c_1 e(t-1) + c_2 e(t-2) \qquad (2)$$

with the error being the difference between the desired force w(t) and the measured force y(t):

$$e(t) = w(t) - y(t) \qquad (3)$$

Denoting the shift-operator: $q^{-1}e(t) = e(t-1)$, results:

$$u(t) = \frac{C(q^{-1})}{1-q^{-1}} e(t) = \frac{c_0 + c_1 q^{-1} + c_2 q^{-2}}{1-q^{-1}} e(t) \qquad (4)$$

and the control loop is depicted in Figure 2.

As De Keyser [1] mentioned, “the DIRAC algorithm can be considered as an auto-tuning as well as an adaptation method”. Indeed, since the identification of the controller parameters is done within the DIRAC strategy, there is no need for specifying a model of the process *a priori*, thus functioning as an *auto-tuning* method. Secondly, if used *on-line*, the PI(D) parameters are adapted continuously, thus resulting in a direct *adaptive* controller. The use of auto-tuning or adaptive control seems appropriate for the control of skeletal muscles, since they are known to be time-varying. The adaptive control method described in this section is easy to understand and simple to apply. In the context of an unknown process model, the assumption that *the muscle and the stimulator* are described by an unknown (discrete-time) transfer function $P(q^{-1})$ leads to the closed loop transfer function:

$$y(t)=\frac{C(q^{-1})P(q^{-1})}{(1-q^{-1})+C(q^{-1})P(q^{-1})}w(t) \tag{5}$$

The *design performance* of the closed loop is specified by a *reference model,* $R(q^{-1})$, given *a priori*. For example, one of the desired characteristics of the closed loop response can be the settling time. The task of controller tuning is to find $C(q^{-1})$ (i.e. c_0, c_1 and c_2) such that the closed-loop transfer function from (5) should approximate the desired reference model $R(q^{-1})$. This can be written as:

$$C(q^{-1})(1-R(q^{-1}))P(q^{-1})\cong(1-q^{-1})R(q^{-1}) \tag{6}$$

Applying (6) to the time-signal $u(t)$, results in

$$\begin{aligned}&C(q^{-1})(1-R(q^{-1}))\underbrace{P(q^{-1})u(t)}_{y(t)}\cong\\&(1-q^{-1})R(q^{-1})u(t)\end{aligned} \tag{7}$$

and becomes:

$$C(q^{-1})(1-R(q^{-1}))y(t)\cong(1-q^{-1})R(q^{-1})u(t) \tag{8}$$

Defining the filtered signals:

$$\begin{aligned}&u_f(t)=(1-q^{-1})R(q^{-1})u(t),\\&y_f(t)=(1-R(q^{-1}))y(t)\end{aligned} \tag{9}$$

and introducing the error signal $\varepsilon(t)$, (8) becomes:

$$\boxed{u_f(t)=C(q^{-1})y_f(t)+\varepsilon(t)} \tag{10}$$

The final step is to estimate (e.g. via least-squares estimator) the parameters in the polynomial $C(q^{-1})$ such that the errors $\varepsilon(t)$ are minimized. The overall DIRAC strategy is schematically presented in Figure 4.

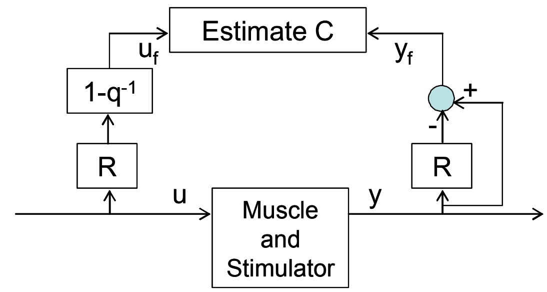

Fig. 4. Scheme of the DIRAC Strategy

4. PERFORMANCE ANALYSIS

Notice that for the simulation presented in this contribution, DIRAC algorithm has been used *off-line* for initial tuning of a PI controller; nevertheless, the method can easily be implemented *on-line* as a direct adaptive controller. In this section, set-point tracking for force is performed with a DIRAC-PI controller. The sampling period was T_s=10ms. The reference transfer function was set to:

$$R(s)=\frac{1}{1+s} \tag{11}$$

which for the nominal model parameters from (1), results in the controller:

$$K_1(s)=\underbrace{0.012}_{K_p}(1+\frac{1}{\underbrace{0.0124}_{T_i}s}) \tag{12}$$

The experiment consisted in changing the set-point from 45N to 95N. On a scale of 22.5N-225N, a set-point change of 50N is about 25%. The result given by the controller is depicted in Figure 5 along with its corresponding control input. It can be observed that the control performance meets the specifications mentioned in Section 3. The output force goes to the desired set-point value of 95N with no overshoot and zero steady state error. Controller output - frequency - stabilizes at the value of 40Hz. Taking into account that the muscles to be activated are paralyzed muscles, a smooth convergence to the set-point force value is more justified than a fast and aggressive convergence, thus avoiding damage of its mechanical properties and supporting its rehabilitation procedure.

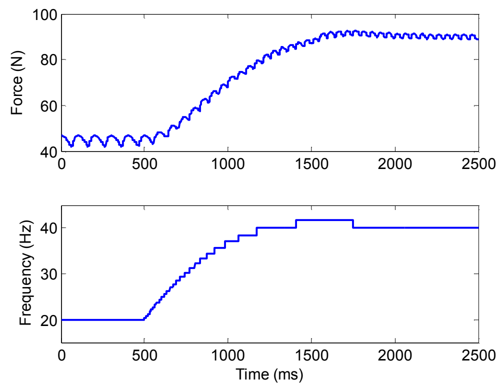

Fig. 5. A closed loop response with a DIRAC-PI controller for the nominal case

A better analysis can be summarized from the Nichols plot. This is done using an in-house developed toolbox for Matlab®, called "Frequency Response Toolbox". The toolbox is detailed in [2] and is freely available at: www.autoctrl.UGent.be/rdk.

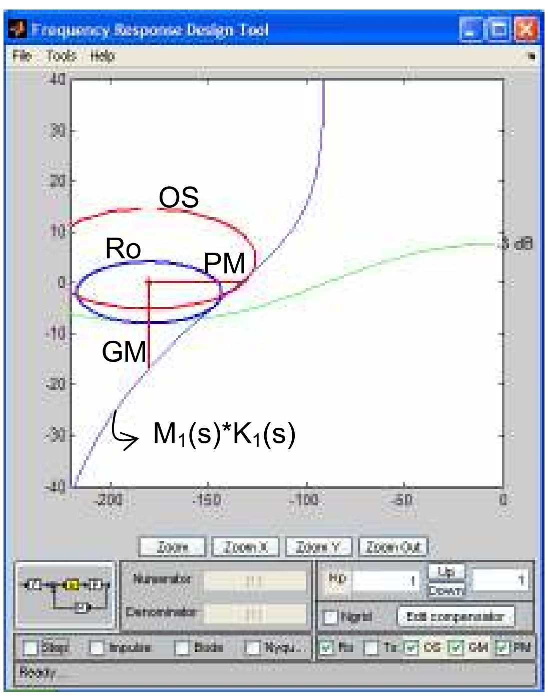

Fig. 6. Nichols plot for the *nominal* situation

For the system in (1) and PI-controller in (12) the Nichols chart is depicted in Figure 6. The performance indexes such as gain margin, phase margin, overshoot and robustness are visually depicted in the figure and their values are:

$$\begin{aligned} GM_1 &= 17dB \\ PM_1 &= 50° \\ OS_1\% &= 20\% \\ Ro_1 &= 0.6 \end{aligned} \tag{13}$$

Notice that the gain of the model in (1) and the controller are both included in the Nichols plot. Therefore, given the gain margin of 17dB (or linearly: 7.07), results that the overall gain can increase seven times before reaching instability. Also the robustness index of 0.6 is rather high (on a scale from 0→1).

The application of adaptive control techniques for skeletal muscle control is motivated by the time-varying character of the system (i.e. in rehabilitation systems). Earlier studies have shown similar performance of the adaptive control strategy with a controller designed based on the explicit model of the system [10]. However, the obvious advantage is that the DIRAC strategy does not require an *a priori* knowledge of the model. As stated previously in Section 3.1, variations in model parameters are expected as a result of the rehabilitation process. Significant changes can occur (see Figure 7, dashed line), changing both the gain and time constants of the muscle, such as:

$$M_2(s) = \frac{500e^{-0.005s}}{s^2 + 16s + 80} \tag{14}$$

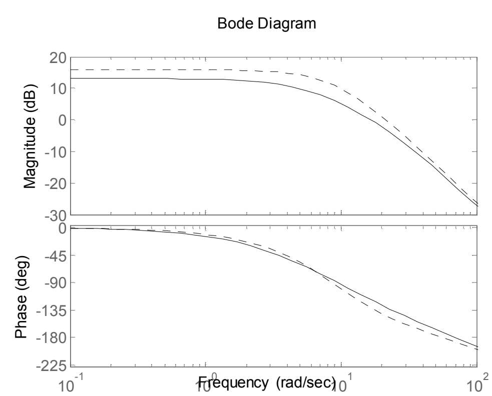

Fig. 7. Bode plots comparing the *nominal* (-) and the *changed* (--) muscle models

This means not only an increase in the overall number of muscle fibres (gain) but also a change in their dynamic response (time constants).

Consequently, without changing the reference function $R(s)$ from (11), the controller adapted to these changes becomes:

$$K_2(s) = \underbrace{0.0114}_{K_p}(1 + \frac{1}{\underbrace{0.0117}_{T_i} s}) \qquad (15)$$

The resulted Nichols plot for the second muscle model and controller is given in Figure 8. The performance indexes such as gain margin, phase margin, overshoot and robustness became:

$$\begin{aligned} GM_2 &= 10dB \\ PM_2 &= 35° \\ OS_2\% &= 35\% \\ Ro_2 &= 0.45 \end{aligned} \qquad (16)$$

Similarly to the first case, the gain of the process and controller are included in the Nichols plot. With a gain margin of 10dB (or linearly: 3.16), results that the overall gain can still increase three times before reaching instability.

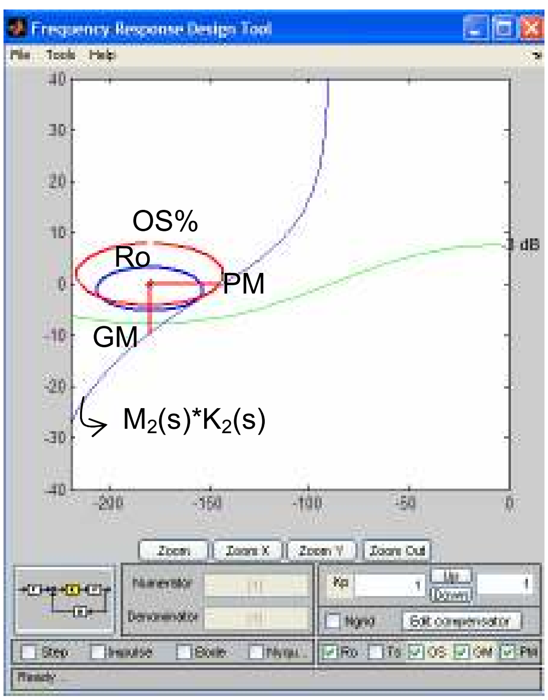

Fig. 8. Nichols plot for the *changed* situation

Comparing the Nichols plots for both control situations, can be stated that although the process has changed significantly, stability margins and robustness are still within reasonable values. It should be emphasized the fact that the reference transfer function $R(s)$ denoting the *desired closed-loop behaviour* was not changed when re-tuning the controller.

In order to compare the performance in both these cases, Figure 9 depicts the step response for the same change in output force set-point as tested previously in Figure 5. It can be seen that the closed-loop dynamics are changed, as well as the gain. For the same output force of 95N, the second controller (dotted line) requires less frequency (from 40Hz decreased to 28.6Hz). This effect is the result of increased muscle mass and improved properties due to the rehabilitation process.

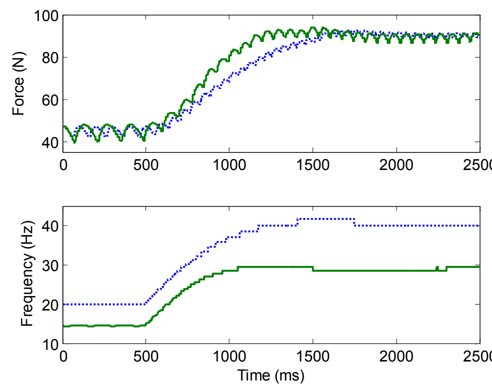

Fig. 9. Comparison of step responses for the nominal control case (-) and changed control case (--).

Although the control strategy can also be used *on-line,* this possibility has to take into account the fact that muscles variations are observed in time intervals of days, weeks, even months. Therefore, continuous on-line adaptation of the controller parameters is not necessary. However, a re-tuning procedure which can be evaluated online during the normal operation can be effectuated every week, to account for small changes in the muscle properties.

5. CONCLUSIONS

One issue that complicates biomedical control and poses modelling problems is the significant inter- and intra-patient variability observed. Therefore, a key component in a control system is the adaptive element. Although it is more difficult to guarantee stability and performance levels for these systems than for static control algorithms, the ability of the algorithm to adapt to a patient's specific behaviours may lead to remarkable performance improvements.

The use of *adaptive and auto-tuning control strategies* such as DIRAC in closed-loop control of paralyzed skeletal muscles is justified by *inter-* and *intra- patient variability*. The muscle properties differ from person to person, as well as from one time-interval to another. Comparable results can be obtained either with or without knowledge of a muscle model [10]. This means that a wearable device can be supported, applied and used successfully on any patient, due to the auto-tuning and adaptive properties of the DIRAC control strategy. The elimination of the identification step considerably simplifies the control task. Taking into account the time delay present in the system is important for real life applications and satisfactory and stable results are obtained.

The present contribution has given a brief overview of the practical problems posed to a control engineer by this application. The task of developing stable and robust control algorithms is not limited to a simple mathematical description. Real-life constraints and hardware/software limitations are to be tackled in an optimal manner, providing a feasible, practical oriented solution.

Although more advanced control can be applied to this example, it may hold significant disadvantages. For example, an important drawback is the need for a model in the prediction algorithm when a model based predictive control strategy is applied. The aim of this presentation has been limited to a simple but satisfactory solution: an auto-tuning algorithm that does not require the identification of a model.

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