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Application of Virtual Reference Direct Control Design in Rehabilitation Engineering

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Abstract

This paper deals with design of feedback controllers for knee joint movement of paraplegics using *Functional Electrical Stimulation* (FES) of the paralysed quadriceps muscle group. The controller design approach, *Virtual Reference Feedback Tuning* (VRFT), is directly based on open loop measured data and fits the controller in such a way that the closed-loop meets a model reference objective. The use of this strategy, avoiding the modelling step, significantly reduces the time required for controller design and considerably simplifies the rehabilitation protocols. Linear and nonlinear controllers have been designed and experimentally tested, preliminary on a healthy subject and finally on a paraplegic patient. It is shown that the control design is effective in controlling the leg position and rejecting disturbances and it could be successfully used in a rehabilitation protocol.

1 Introduction

The aim of this work is the design of neuroprostheses for paraplegic patients suffering from spinal cord injury. In a paraplegic subject the muscles of the lower limbs are intact and capable of effective contraction, but there is the fundamental lack of stimulation from the brain through the injured spinal cord. Functional Electrical Stimulation (FES) can provide the missing electrical stimuli to induce muscle contraction and the corresponding joint movement. So, in the case of this paper, by stimulating the quadriceps muscle group a movement of the knee joint can be obtained. By continuous monitoring of the knee joint angle it is possible to design a feedback controller to decide the electrical stimulation patterns necessary to achieve predetermined movements of the knee useful in more complex motor functions such as cycling, standing up, sitting down or stepping.

A preliminary step for the design of a controller is often a modelling stage, where an input-output dynamical system describing the relationship between the input FES pattern and the knee joint position is obtained (see for instance [1, 2, 3, 4]). In most cases, black-box models are obtained starting from an input-output sequence collected on the patient; then a controller is designed and applied to the subject (see [2, 3, 5]). Unfortunately, in designing a neuroprosthesis, the identification stage is time consuming, basically because it could be difficult to make a choice on the model structure [4, 6]. The longer the time used for model estimation, the more the patient feels not comfortable and the more the muscle properties could change. In fact, in the design of a controlled neuroprosthesis, the time spent to design the controller represents a fundamental issue which could lead to rapid fatigue of electrically stimulated muscles. Furthermore, the interest in the well-being of the patient do not allow long lasting open loop tests for data collection, identification and time consuming modelling procedures. Moreover, due to the day-to-day change of the neuro-musculoskeletal system, an update of the controller parameters dependent on currently measured open loop Input/Output data is highly recommended.

So, it is advisable to investigate control design strategies that avoid the identification step, to be used in all those cases where the subject is in particular physical or psychological conditions and cannot bear long stimulation sessions. In this paper, the application of *Virtual Reference Feedback Tuning* (VRFT) strategy to the challenging problem of controlling joint positions by means of electrical stimulation of muscles will be presented and results obtained in experimental sessions will be discussed. The VRFT method gives a solution to the problem of designing a controller for a system, whose I/O behavior is unknown, on the basis of a single set of I/O data without resorting to the identification of a model of the system and with no need of any on-line update of the controller parameters. The idea on which VRFT is based was originally proposed in [7] and developed in [8] as a complete and ready to use method for data-driven control design in a noisy environment. Obviously, the aim of the VRFT approach is not the design of the "*best*" possible controller, but the design of a controller with good tracking and robustness properties, based on a reduced set of data and so tunable in short time. Preliminary results based on simulations and a first experiment on a healthy subject were presented in [9]. These encouraging results have pushed us to extensive testing on healthy subjects and, finally, on paraplegics. In this paper, experimental results obtained using linear and nonlinear controllers on both healthy persons and paraplegics will be discussed, testing tracking of typical reference signals and robustness of the closed-loop system.

The structure of the paper is the following: in Section 2 the experimental setup is described. This experimental protocol is the fruit of a long and consolidated work on this control problem, which can be considered as a benchmark in FES control applications [2, 5]. In Section 3 the control strategy adopted in the experiments is described. Then, in Section 4 experimental results obtained with the linear controller on a healthy subject are presented. Also a nonlinear modification of the control algorithm is proposed and tested on a paraplegic patient.

2 Experimental Setup and Protocol

The experiments have been carried out on two subjects: a 27 years old healthy male and a 58 years old male paraplegic subject, four years past from an injury at T10 level. During each session the subject is sitting on a bench with the unloaded shank free to swing (see Fig. 1). The absolute knee angle Θ in degree is measured in real-time by an ultrasound based 3D-motion analysis system [10]. The angle is directly calculated from the measured positions of three markers placed at the ankle, knee and hip. This sensor system is connected to a laptop by a proprietary interface. First of all the knee angle is normalised so that it varies from $y = 0$ at rest position without any stimulation to $y = 1$ at maximum knee extension when the muscle is fully contracted. Rest position and maximum knee extension are in general not equal to $\Theta = 90°$ and $\Theta = 180°$ respectively. In particular, in a paraplegic these values could be considerably reduced (about $110° - 150°$), as a consequence of the higher stiffness of the joints in a paraplegic compared to a healthy person.

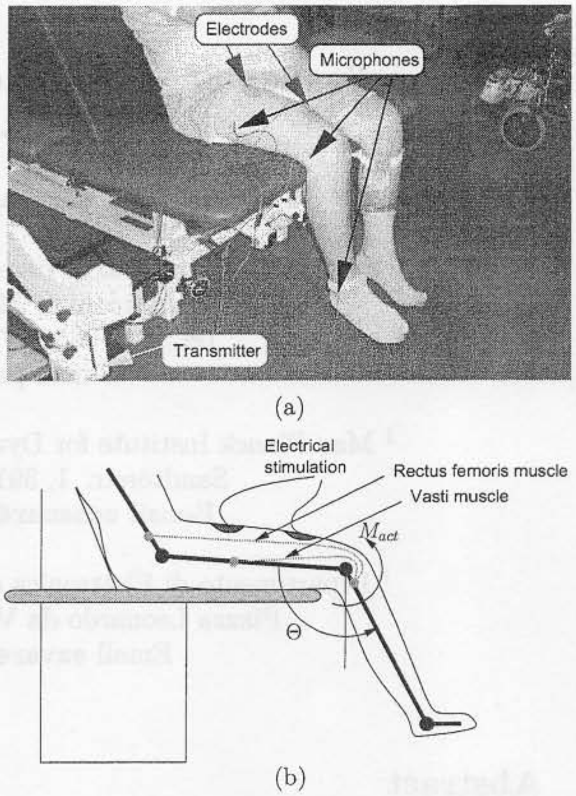

Figure 1: (a) Position of the subject during the experiments. The three markers (microphones) of the ultrasound motion analysis system are visible at the knee, at the ankle and at the hip. The electrodes are visible just above the hip and knee markers. On the left side a part of the ultrasound transmitter unit can be seen. (b) Definition of the knee joint angle Θ. The active knee moment M_{act} is induced by electrical stimulation of the quadriceps muscle group (rectus femoris and vasti muscle).

The data are collected with a sampling time $T_s = 0.05\,\mathrm{s}$. The quadriceps muscle group is stimulated in a non-selective way using surface electrodes which deliver the electrical pulses generated by a portable stimulator connected to the laptop computer. The frequency is $f = 20\,\mathrm{Hz}$ and the electrical charge delivered to the muscle is modulated by changing the pulse width. So, the actual control variable is the modulated pulse width measured in μs. All the implementation is done in Matlab© using Real Time Toolbox© [11].

Each experimental session is made of three main steps.

At the beginning, a *preliminary test* is made in order to choose the stimulation current (usually about $I = 40 \ldots 80\,\mathrm{mA}$). Starting with a low current, the pulse width is ramped up in steps from 50 to about $400 - 500\,\mu s$ and the angle is measured. Then the current is incremented by 10mA and the stimulation pattern is repeated. This process continues until the knee is close to full extension at high pulse widths. The current is fixed at this level for the remainder of the experiment. The input (pulse width) is desaturated and normalised to the range 0–1. This is done using the result from the previous preliminary test. The pulse width at which the knee comes out of low-level saturation (rest position) is taken as $u = 0$ and the pulse width at which the knee goes into high-level saturation (extension) corresponds to $u = 1$. Finally, an estimation of the system's equilibrium curve for normalised pulse width and angle is obtained from the measured data which will be useful to prepare the input pattern for controller design.

After this first stage, the *data set for control design* is generated. A carefully designed input signal is applied to the electrically stimulated muscle. The used pulse width signal possesses specific stochastic characteristics. The goal is to generate a sequence of input steps which leads to an almost uniform distribution of the output over the achievable range of knee joint motion. The normalised pulse width signal changes at each sampling time to a new value with a predetermined switching probability; if the input is to change, the new value is drawn from a uniform distribution over the interval 0–1 and afterwards processed by the inverse static (equilibrium) curve of the knee joint dynamics, estimated in the preliminary test. An example of I/O data used for controller design is shown in Fig. 2. Finally, the last step of the experimental session consists in testing the controller designed according to the method described in Sect. 3 with the data collected at the previous stage. Examples of theses results will be presented in Sect. 4.

3 Controller design using Virtual Reference Feedback Tuning

Using the data collected at the beginning of the experimental session, the VRFT allows the design of a controller for an unknown system without resorting to the identification of a model of the system.

The VRFT method approximately solves a model-reference problem. This means that the control specifications are assigned via a reference model $M(q)$ which describes the desired behaviour of the closed-loop system. The model $M(q)$ is used to define in a simple and effective way the basic characteristics of the control system such as its settling time, the allowed overshoot etc. In standard applications, a second order model is usually satisfactory.

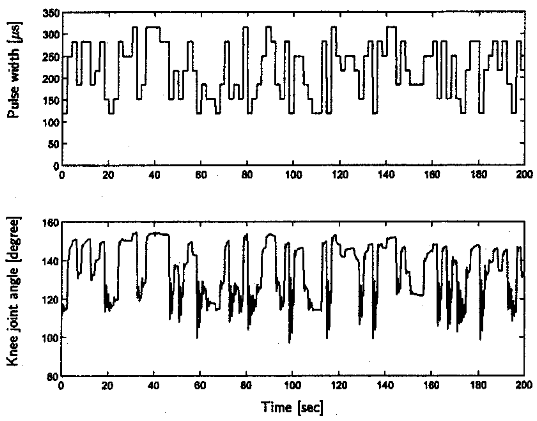

Figure 2: Input/Output data collected on a paraplegic. This is the data set used for the design of a linear controller according to the VRFT strategy. The upper section represents the stimulation input; the lower part is the measured knee joint angle.

In the VRFT approach a given SISO linear plant is considered, whose (unknown) pulse-transfer function is $P(q)$:

$$\begin{cases} y(t) = P(q)u(t) \\ u(t) = C(q;\theta)(r(t) - y(t)) \end{cases} \tag{1}$$

In Eq. (1) q is the forward shift operator, $C(q;\theta)$ is a linear one-degree-of-freedom controller belonging to a given family of linear controllers $\{C(q;\theta)\}_{\theta\in\mathbb{R}^n}$ parameterised by the n-dimensional real vector θ. The signals $u(t)$ and $y(t)$ are respectively the input and the output of the considered plant. The signal $r(t)$ is the reference signal of the control loop.

Let us assume that a set of I/O data $\{u(t), y(t)\}_{t=1,..,N}$ has been collected from an experiment on the plant and that a reference model $M(z)$ has been chosen. The goal is to solve a model reference problem, i.e. to find

$$\bar{\theta} = \arg\min_{\theta} J_{MR}(\theta) \tag{2}$$

where

$$J_{MR}(\theta) = \left\| \frac{P(z)C(z;\theta)}{1 + P(z)C(z;\theta)} - M(z) \right\|_2^2 . \tag{3}$$

To this aim, given the measured $y(t)$ (i.e. the actual signal measured at the output of the plant), consider a *reference* $r_v(t)$ such that $M(q)r_v(t) = y(t)$.

Such a reference is called "*virtual*" because it does not exist in reality and in fact it was not used in the generation of $y(t)$. So, in this framework, $y(t)$ is the desired output of the closed-loop system when the reference signal is $r_v(t)$. Finally, compute the corresponding *tracking error* $e(t) = r_v(t) - y(t)$.

Notice that, even though the plant transfer function $P(z)$ is not known, when the plant is fed by $u(t)$ (the actually measured input signal), it generates $y(t)$ (the corresponding measured output signal). Therefore, a good controller generates $u(t)$ when fed by $e(t)$, at least in the condition when the reference signal is the *virtual reference* $r_v(t)$. The idea is then to search for such a controller. Since both signals $u(t)$ and $e(t)$ are known, this task reduces to the *identification problem* of describing the dynamical relationship between $e(t)$ and $u(t)$ by using the family of linear models $\{C(q;\theta)\}_{\theta\in\mathbb{R}^n}$.

In the following, the algorithm implementing the above idea will be briefly outlined. In the algorithm, the identification of the controller is addressed by minimising the classical least-squares identification criterion (see [12]).

Design Algorithm

Given the reference model $M(q)$, the family of controllers $\{C(q;\theta)\}_{\theta\in\mathbb{R}^n}$ and the set of data $\{u(t), y(t)\}_{t=1,..,N}$, do the following:

1. Calculate:
   - a virtual reference $r_v(t)$ such that $y(t) = M(q)r_v(t)$, and
   - the corresponding tracking error $e(t) = r_v(t) - y(t)$.
2. Filter the signals $e(t)$ and $u(t)$ with a suitable filter $L(q)$, obtaining $e_L(t)$ and $u_L(t)$:
   $$e_L(t) = L(q)e(t), \qquad u_L(t) = L(q)u(t).$$
3. Estimate the controller parameter vector
   $$\hat{\theta}_N = \arg\min_{\theta} J_{VR}^N(\theta) \tag{4}$$
   where
   $$J_{VR}^N(\theta) = \frac{1}{N}\sum_{t=1}^{N}\left(u_L(t) - C(q;\theta)e_L(t)\right)^2. \tag{5}$$

Notice that Eq. (5) is quadratic in the parameter vector θ and all the computations are directly performed on the measurement data, assumed that the filter $L(q)$ is given.

In this work, we consider the class of PID controllers as having the following form:

$$C(q;\theta) = K_p + K_i\frac{T_s}{1-q^{-1}} + K_d\frac{1-q^{-1}}{T_s} \tag{6}$$

where T_s is the sampling time of the system. Notice that since the controller is linear in the parameters $\theta = [K_p\ K_i\ K_d]$ the performance index (5) is quadratic and the parameter estimate $\hat{\theta}$ can be easily found.

The practical implementation of the controller has been realized with a standard implementation of an antiwindup scheme [13].

The proposed algorithm could be modified in order to obtain nonlinear controllers. The use of nonlinear controllers is strongly advisable in FES control. In fact, it has been evidenced by experiments that the muscle–joint system is not a linear plant. The main sources of nonlinearity are gravity and nonlinear muscle contraction dynamics. So, a nonlinear version of VRFT has been designed, where the control action is given by

$$u(t) = \sum_{i=1}^{S} g_i(v(t-i+1)) \tag{7}$$

where

$$v(t) = \frac{1}{1-q^{-1}}e(t) \tag{8}$$

and

$$g_i(v(t-i+1)) = \sum_{k=1}^{M}\theta_k e^{-(r_i(v(t-i+1)-c_k))^2} \quad \text{for all} \quad i = 1,\dots,S \tag{9}$$

with $r_i \in \mathbb{R}$ for all $i = 1,\dots,S$ and $\theta_k, c_k \in \mathbb{R}$ for all $k = 1,\dots,M$.

The previous representation corresponds to the use of S nonlinear functions built using M radial basis functions for each of them. The parameters c_k and r_i are chosen a priori: they are respectively the centres and the widths of the basis functions. So, the control action is linear in the parameters θ_k which could be estimated by using the same algorithm as described before.

4 Experimental Results

Once that a controller has been designed for a subject following the algorithm of Sect. 3, two kind of tests were performed in order to qualitatively evaluate the properties of the design: *tracking tests* and a *disturbance rejection test.*

In a *tracking test* stair-like or square wave reference signals are used. One of the reasons for using such reference signals is that they can be safely used by physicians in rehabilitation protocols. When performing such tests using linear design, the reference signal is usually not spanning all the extension range of the knee joint and it is limited to about $120°-150°$ in healthy subjects and $130°-140°$ in paraplegics who have smaller extension ranges.

The *disturbance rejection test* consists in pushing or pulling the shank (or putting a known weight on the ankle) while the controller is maintaining it at a fixed reference angle. Observing how fast and smoothly the controller leads the leg back to the reference position, one can obtain an impression on the degree of robustness of the closed-loop system. For the sake of brevity, results about this kind of test will be shown only for paraplegics.

One of the problems that must be usually faced in the experimental sessions is the increasing fatigue in the stimulated muscles: so, sessions could not last for a long time, in particular when paraplegics are involved. Anyway, since the controller includes an integral action, it is able to counteract slowly time-varying effects.

First, results of the linear control for a healthy subject will be presented. Then, results of the nonlinear control on a paraplegic will be shown.

4.1 Healthy subject

In Fig. 3, an example of a *tracking test* is shown for the linear controller (6). Specifically, a stair-like reference signal is used. The reference range has been kept limited to the central region (about $120° - 150°$) of the extension range of the healthy knee, so that the linear controller could be effective.

Healthy subjects have of course the ability to perform voluntary muscle contractions. Although the subject was blindfolded during the experiment and could not see the reference trajectory on the laptop screen, significant interaction of feedback controller and subject could be observed. In fact, the healthy subject had the tendency to induce normal muscle contractions by himself which were initially triggered and generated by FES when the reference changed to a higher knee extension level. As a result, the controller used a reduced intensity of the control action to track the reference. In Fig. 3, the pulse width is continously declining after a higher knee joint extension level is reached; this indicates that increasing voluntary torque is present.

When the reference changed to a lower level the subject needed some time to relax the muscles, which can be seen in the slowly raising pulse width after the lower level has been reached; voluntary muscle contraction is decreasing during this period.

However, the closed-loop system shows good tracking properties. Small tracking offsets are resulting from the voluntary muscle contractions. The knee joint angle follows the reference angle accurate while the control signal is varying smoothly.

4.2 Paraplegic subject

A paraplegic subject does not have the ability of contracting muscles, so he cannot interact with the controller. The tracking tests with the linear controller yielded comparable good results as with the intact

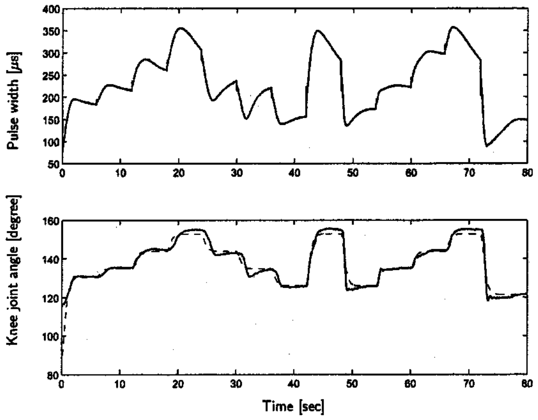

Figure 3: Tracking of a sequence of steps using a linear PID controller applied to a healthy subject. The upper part represents the control action (stimulation input). The lower section is the knee joint angle (– measured angle, - - reference filtered by the tracking reference model).

subject. As expected, no significant interaction of subject and controller could be detected.

From Fig. 4, disturbance rejection properties of the linear controller can be evaluated. When the disturbances occur the knee joint angle initially drops or increases depended on the kind of disturbance but the controller brings the shank automatically back to the position to be maintained.

In Fig. 5, the tracking of a square wave is shown whereas the experiment was performed by using a nonlinear controller. Excellent tracking performance could be achieved.

5 Concluding Remarks

In FES muscle fatigue plays a crucial role, making it difficult to stimulate a subject (healthy or paraplegic) for a long time. In particular, in identification or control design experiments, this could lead to models that are no longer valid or to poorly performing controllers, since muscle fatigue could change even dramatically the muscle/joint system response to stimulation. In this sense, controller designed directly from data could be of great help since they can considerably shorten the session duration.

The proposed Virtual Reference Feedback Tuning strategy has been successfully tested in experiments with intact and paraplegic subjects.

Based on a single measured I/O data set, the design is yielding a feedback controller with excellent tracking and disturbance rejection properties.

Since integral action is embedded into the controller, the controlled neuroprosthesis can successfully counteract slowly varying system properties due to fatigue or constant disturbances. During the tests performed, muscle fatigue was not significantly affecting the muscles as durations of single control sessions have been kept short.

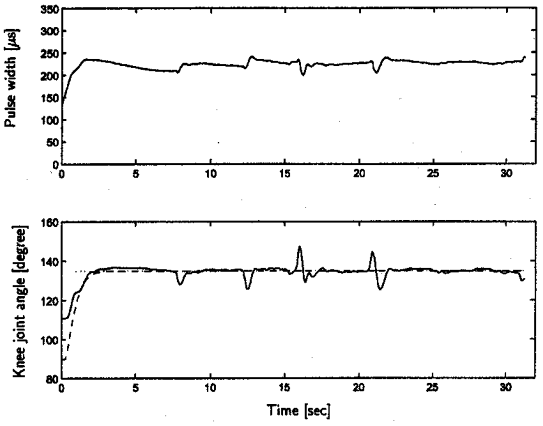

Figure 4: Verification of the PID controller's disturbance rejection properties on a paraplegic subject. The upper section represents the control action (stimulation input). The lower part is the knee joint angle (– measured angle, $\cdots$ reference filtered by the tracking reference model). At about $t = 8\,\mathrm{s}$ and $t = 12\,\mathrm{s}$ the subject leg has been pushed down. At about $t = 16\,\mathrm{s}$ and $t = 21\,\mathrm{s}$ the subject leg has been lifted up.

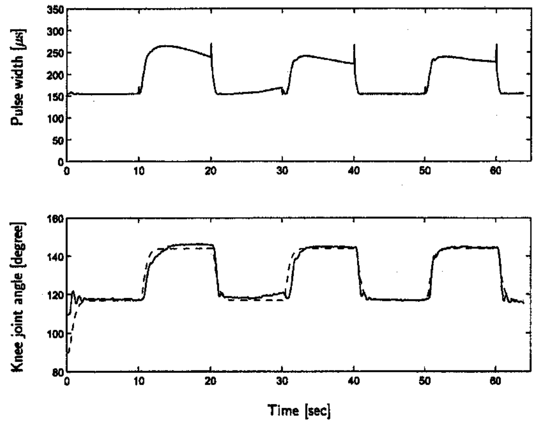

Figure 5: Tracking test of a nonlinear controller on a paraplegic subject. The upper section represents the control action (stimulation input). The lower part is the knee joint angle (– measured angle, $\cdots$ reference filtered by the tracking reference model).

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