LINEAR CONTROLLER DESIGN FOR THE SINGLE LIMB MOVEMENT OF PARAPLEGICS

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Abstract: This paper deals with the problem of modelling and control of the knee joint dynamics. Using controlled functional electrical stimulation (FES) of the quadriceps muscle group the knee joint will be positioned at a commanded angle, subject to disturbances and displacements of the hip and ankle joints. A simple ARX-model has been estimated directly from measured input-output plant data. Further, a discrete linear pole-placement controller will be presented as solution for the tracking and regulation problems. Advantages of this approach in comparison with nonlinear controllers based on physiological models will be discussed and natural limitations are shown. A simple tuning rule for an already determined controller will be given to achieve robust stability with respect to the plant characteristics, which depend on the actual operating point. Experiments with neurologically intact subjects show encouraging results.

Keywords: Biocybernetics, Electrical stimulation, Pole assignment, Discrete time, Parameter estimation

1. INTRODUCTION

The control of single limb movements of paraplegics represents an important preliminary stage towards more complex motor function restoration of handicapped persons suffering from spinal cord injury. Of particular interest in this case are tasks like standing-up, standing and sitting-down. Using functional electrical stimulation (FES) with surface electrodes results in muscle contraction and subsequent joint movement. A feedback control system will be investigated for knee joint movement. Here, the quadriceps muscles (rectus femoris and vasti) are stimulated by one pair of electrodes. The resulting knee joint angle is measured. Although this experiment seems simple, it is a good example in the study of the general characteristics of human kinematics. Because of the relatively large movement range a non-isometric muscle contraction arises, so that the moment generated at the joint depends very strongly on the muscle length and velocity. Both of these variables are functions of the set of joint angles and their derivatives. Movements of the hip joint for example have an influence on the knee joint due to the biarticular behaviour of the rectus femoris muscle group. Additional complicating effects are the presence of a nonlinear recruitment curve for the muscle activation as well as some nonlinear terms in the equations of motion due to gravity and elastic passive moment. Constraints on the input (stimulation level) and output (limited range of knee angle) make the controller design more difficult.

To deal with this problem several authors have developed neuromusculoskeletal models and them used for controller design (Palazzo *et al.*, 1998; Riener and Quintern, 1996; Riener and Fuhr, 1998; Ježernik and Riener, 1999; Chizeck *et al.*, 1999). These models take into account the major properties of the muscle and segmental dynamics

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during FES. This approach requires a large effort to determine the actual values of the model parameter for each new individual. In (Riener and Quintern, 1996; Palazzo *et al.*, 1998) a feedforward controller consisting of an analytic inverse of the model captures the major nonlinearities and gives a nominal control action to let the knee angle follow a commanded trajectory. A simple feedback controller, e.g. proportional-integral-derivative (PID), then has to reduce the tracking error and achieve stability. However, the controller parameters are not chosen on the basis of a parameterized model but rather through an empirical procedure. This is often time consuming and does not lead to an optimal feedback controller, as the results in (Palazzo *et al.*, 1998; Riener and Quintern, 1996) show. Since the neuromusculoskeletal model has continuous time character the design of the required discrete controller becomes a problem. A quasi-continuous assumption fails due to the relatively long sample time which is defined by the stimulation frequency. A comparison between a PID controller and a sliding mode controller can be found in (Ježernik and Riener, 1999) based on a simulation using a computer model of the knee joint (Riener and Fuhr, 1998). The application of the Ziegler-Nichols tuning rule for the PID controller yields, as expected, unsuitable tracking performance. This is because the tuning rule used results in a closed loop with a damping ratio less than 0.25. The tracking performance of the sliding mode controller is good. However, the amplitude and rate of change of the control signal are unacceptably high. A very interesting continuous time robust sliding mode control approach is proposed and tested by simulation in (Durfee, 1993). It is assumed that the moment generated at the knee joint can be changed continuously in time. In (Chang *et al.*, 1997) the inverse knee joint dynamics were modelled as a neural network and used in the forward path of a controller with two degrees of freedom (2DOF). As in (Palazzo *et al.*, 1998; Riener and Quintern, 1996), a PID-controller is placed in the feedback path and determined by the Ziegler-Nichols method based on the step response of the open-loop system. Previdi et al. (1999) describe a discrete NARX polynomial model which is identified using data collected in an isotonic stimulation session. Based on the NARX model and its linearisation a gain scheduling controller and a linear LQG controller are determined. In comparison to the linear LQG controller the gain scheduling controller works in a larger operating regime, including various equilibrium points. However, the extrapolation behaviour of NARX polynomial models is well known to be poor. In the case of the knee joint dynamics for example good extrapolation behaviour is required to guarantee good performance near the knee lock position (full extension). A model reference adaptive controller (MRAC) knee joint control in paraplegics was developed by Hatwell et al. (1991). The design is based on a discrete ARX model of 3rd order and uses a nonlinear recruitment characteristic compensation scheme. To avoid problems with parameter convergence the parameter estimator is only switched on in the linear region of the operating range. Consequently, the tracking performance suffers in situations such as full knee extension or flexion.

Based on the above analysis the design of knee-angle controllers can generally be improved. Approaches like (Hatwell *et al.*, 1991; Previdi *et al.*, 1999) point in the right direction, but the tracking performance can still be bettered. The 2DOF structure (Palazzo *et al.*, 1998; Chang *et al.*, 1997; Riener and Quintern, 1996) can be unsuitable if the feedback controller is locally designed and cannot guarantee stabilisation in the whole operating range of interest. As a consequence the stability of the 2DOF structure can be lost. In this paper it will be shown that with a linear discrete pole-placement controller good tracking performance and stability can be achieved. The design is based on a simple model of the knee joint dynamics whose parameters are empirically determined.

2. EXPERIMENTAL SETUP AND METHODS

To carry out our investigations the following experimental setup (cf. Fig. 1) was used: The subject was seated on a table with the unloaded shank free to swing. The knee angle $\Theta(t)$ is measured by an electrogoniometer (Wood *et al.*, 1998) and varies from $\Theta(t) = 0$ (rest-position) to $\Theta(t) = 1$ (full-extension). Using a PCMCIA data acquisition card the angle was sampled with $T_s = 100$ms. The stimulator (Phillips *et al.*, 1993) is connected to the laptop via the serial port and delivers constant current rectangular pulses with pulsewidth up to $500\mu s$. During the experiments the pulsewidth serves as a variable control signal whereas the current amplitude $I = \{50, 60, 70, 80, 90\}$mA and the stimulation frequency $f = 20$Hz are fixed. All implementation was done in Matlab® using the Real Time Toolbox® (Humusoft, 1999). The surface electrodes were positioned as shown in Fig. 1(a) (Kralj and Bajd, 1989).

2.1 *Modelling*

A simple discrete-time ARX-model is estimated directly from measured input-output plant data. The model is described by

$$y(k) = \frac{B(q^{-1})q^{-n_k}}{A(q^{-1})}u(k) + v(k) \qquad (1)$$

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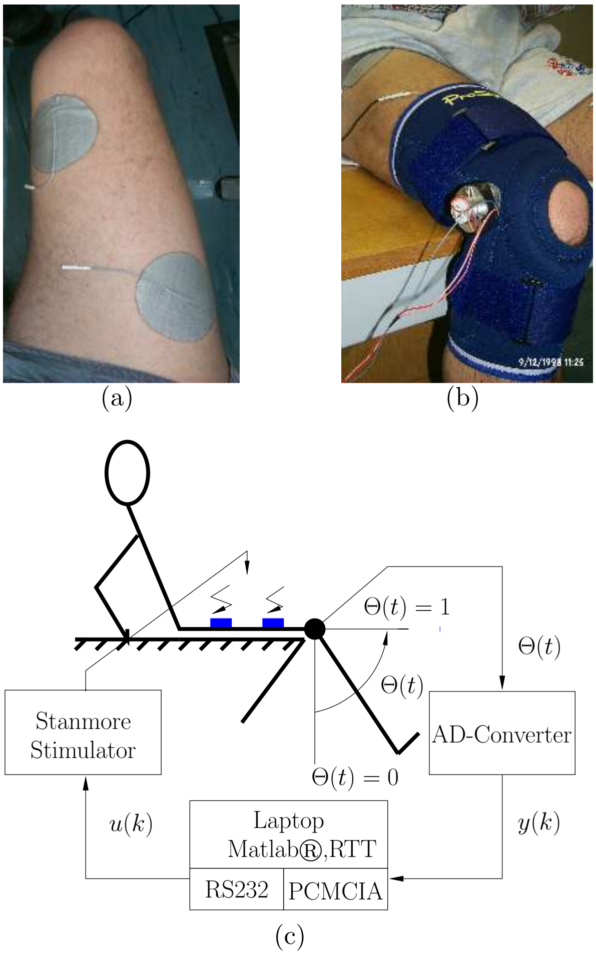

Fig. 1. Experimental setup: (a) electrode placement on the quadriceps, (b) goniometer and (c) schematic of experimental setup.

with

$$A(q^{-1}) = 1 + a_1 q^{-1} + \cdots + a_{n_a} q^{-n_a} \qquad (2)$$

$$B(q^{-1}) = b_0 + b_1 q^{-1} + \cdots + b_{n_b} q^{-n_b}. \qquad (3)$$

Here, y is the system output (sampled knee angle), u is the input (pulsewidth), v represents an output disturbance and q^{-n_k} is a time delay of n_k sample steps. A and B are polynomials in the unit delay operator q^{-1}. The coefficients of the polynomials are determined by the standard least squares method. In the first step the current amplitude is selected in such a way that the knee is fully extended at the maximum of the control signal range ($u(k) = 0 \ldots 500\mu$s). To achieve this a ramp in the pulsewidth is applied to the plant. From this static experiment an operating point is taken which represents the middle of the knee angle range. The necessary data for the identification are collected at this equilibrium point using a pseudo binary random signal (PRBS) as input with a duration of about 40s. Dividing the data set in an estimation and validation part the model structure and parameters have been determined in the usual way (Åström and Wittenmark, 1997).

2.2 *Controller Design*

Based on the identified model a linear discrete input-output pole-placement controller with two degrees of freedom (2DOF) can be designed (Åström and Wittenmark, 1997) for the selected operating point. The controller has the general form (see Fig. 2)

$$u(k) = \frac{1}{R}\left(Tr(k) - S(y(k) + n(k))\right). \qquad (4)$$

Here, $r(k)$ is the reference signal and $n(k)$ a measurement noise. R and S are the controller polynomials in the delay operator, which are defined by

$$R(q^{-1}) = 1 + r_1 q^{-1} + \cdots + r_{n_r} q^{-n_r} \qquad (5)$$

$$S(q^{-1}) = s_0 + s_1 q^{-1} + \cdots + s_{n_s} q^{-n_s}. \qquad (6)$$

These polynomials and the pulse transfer function T have to be determined in such a way that the desired output response y_m to command signals r becomes:

$$y_m(k) = H_m(q^{-1})r(k) = \frac{B_m(q^{-1})}{A_m(q^{-1})}r(k). \qquad (7)$$

It is further assumed that the controller can cancel some of the plant poles and zeros. Assume that the polynomials A and B are factorized as $A = A^+A^-$ and $B = B^+B^-$ where A^+ and B^+ are the stable factors that will be cancelled. To obtain perfect model following the numerator B_m of the reference model H_m must contain the factor $q^{-n_k}B^-$, because $q^{-n_k}B^-$ cannot be cancelled. Thus, B_m can be written as $B_m = q^{-n_k}B^-\overline{B}_m$. For the attenuation of constant disturbances the controller is required to have integral action. This means that the controller polynomial R has to contain the factor $(1 - q^{-1})$. The above requirements and assumptions give

$$R = (1 - q^{-1})B^+\overline{R} \qquad (8)$$

$$S = A^+\overline{S} \qquad (9)$$

$$T = \frac{\overline{B}_m A_o A^+}{A_m} \qquad (10)$$

where the polynomial A_o is denoted as the observer polynomial. Combining (1)-(10), the closed-loop characteristic polynomial A_{cl} is easily found by

$$A_{cl} = AR + q^{-n_k}BS = A^+B^+A_o$$

$$A_{cl} = A^+B^+(A^-(1 - q^{-1})\overline{R} + q^{-n_k}B^-\overline{S}) = A^+B^+A_o. \qquad (11)$$

To obtain the controller the Diophantine equation (11) has to be solved for $\overline{R}, \overline{S}$ and R, S, T computed from equations (8)-(10). The specifications for tracking are governed by the pulse transfer function $H_m = B_m/A_m$. The desired regulation behaviour is given by the observer polynomial A_o. Separation of the disturbance and command signal response is achieved. To see this the transfer functions from the command signal r, the output disturbance v and the measurement noise n to the output y are calculated as

$$\begin{aligned} y &= \frac{q^{-n_k} BT}{AR + q^{-n_k} BS} r + \frac{AR}{AR + q^{-n_k} BS} v \\ &\quad - \frac{q^{-n_k} BS}{AR + q^{-n_k} BS} n \\ y &= \frac{B_m}{A_m} r + \frac{(1 - q^{-1}) A^- \overline{R}}{A_o} v \\ &\quad - \frac{q^{-n_k} B^- \overline{S}}{A_o} n. \end{aligned} \quad (12)$$

The control structure is shown in Fig. 2. Here, the constant gain k_p will be explained later and can be assumed as $k_p = 1$ at the moment. To achieve unity gain of the reference model and to leave the system zeros unchanged ($B^- = B, B^+ = 1$) the polynomial B_m is chosen as

$$B_m = q^{-n_k} B A_m(1)/B(1). \quad (13)$$

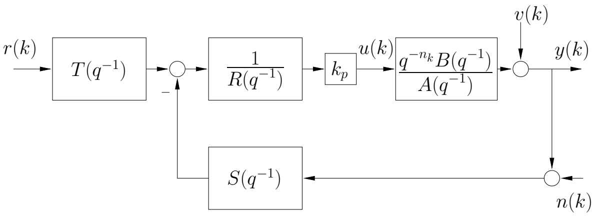

Fig. 2. Controller Structure

3. RESULTS AND DISCUSSION

In the following experimental results with a neurologically intact subject will be presented. The local model and controller are designed for the static operating point $\Theta^s = 0.7$ and $u^s = 150\mu s$, corresponding to a nearly extended knee. For the plant model, obtained from the identification trials, the structure is given by $n_a = 2$, $n_b = 0$ and $n_k = 2$. Fig. 3 shows data from the identification and from the identified model, given by the following pulse transfer function:

$$y(k) = \frac{0.0005534 q^{-2}}{1 - 1.21 q^{-1} + 0.41 q^{-2}} u(k) + v(k) \quad (14)$$

The plant characteristics are dependent upon the equilibrium point. In summary, the system poles for low knee extension are complex conjugate and underdamped and become real and overdamped for near-full knee extension. The system gain decreases by a factor of approximately 5-15 in the same range of operation. The significant change in

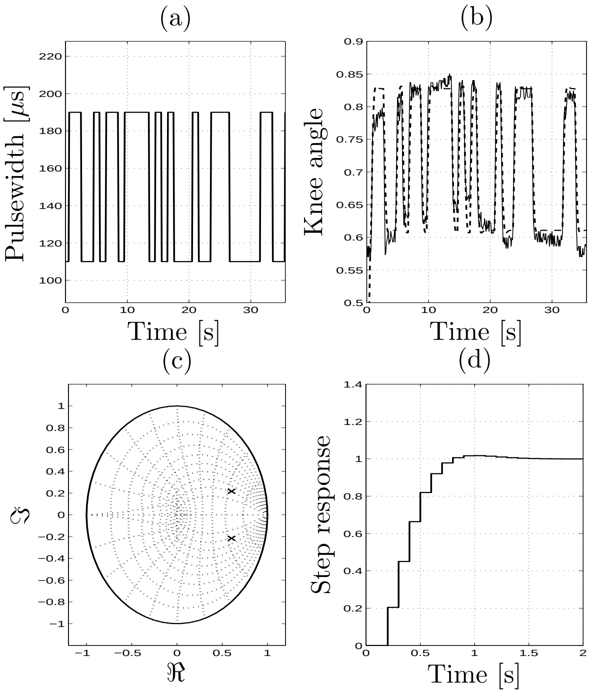

Fig. 3. Identification results for a neurologically intact subject ($I = 60$mA): (a) Applied PRB-signal (b) Measured (solid) and simulated (dashed) system response (c) Pole/zero-map and (d) Normalized step response of the ARX-model.

the system gain is critical for controller design. A controller designed for an equilibrium point with smaller gain can destabilise the closed loop during operation at an equilibrium point with larger plant gain. To achieve a controller with stability robustness it is recommended to use the linear model with the largest gain for controller design. Another possibility is to adapt the open-loop gain to compensate the changing system gain. This procedure works if a controller has already been designed. In both cases it is assumed that the system time constants do not vary widely for different operating points. Adapting the open-loop gain means that the factor k_p in Fig. 2 will be changed. Results of control experiments are shown in Fig. 4 and 5. Here, the controller C1, as well as the later modification, is based on the model (14). To specify the polynomials A_m and A_o with degree $\deg A_m = \deg A_o = 2$ discretisation of a continuous time system of 2nd order was used. This system is defined by the rise time $t_{r_{A_m}}$ or $t_{r_{A_o}}$ and the damping coefficient ζ_{A_m} or ζ_{A_o}. For all controllers the design parameters are shown in Table 1. The control signal is relatively smooth and the knee angle of the subject follows the filtered reference angle y_m very accurately. Such a control signal is desirable for paraplegic patients, because a smooth control signal is less likely to excite unwanted spastic reflexes. Although the controller was designed for the knee angle range $\Theta = 0.6 \ldots 0.8$ it works very well in the range $\Theta = 0.3 \ldots 0.9$ (a range of approximate 60^o). Due

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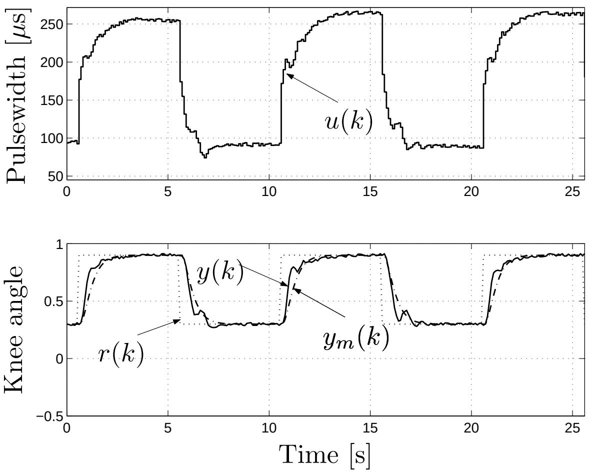

Fig. 4. Controller C1: Test of the tracking performance; r reference signal, y_m filtered reference signal (desired knee angle), y measured knee angle, u pulsewidth.

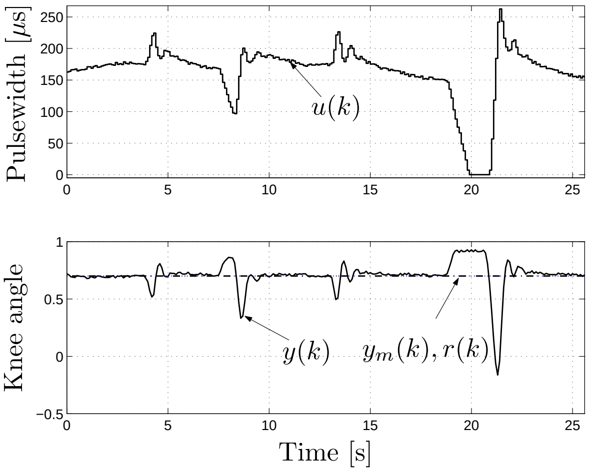

Fig. 5. Controller C1: Test of disturbance attenuation ($t = 4$ and $t = 13s$ pushing down of the leg, $t = 7.5s$ and $t = 18s$ lifting up of the leg).

to the characteristics of the muscle groups during FES it was impossible to reach an angle higher than $\Theta = 0.9$. For angles $\Theta < 0.3$ stability could not be achieved (without loss of performance). The controller is also able to counteract disturbances successfully (cf. Fig. 5). The leg was disturbed by pushing it downwards at $t = 4$ and $t = 13s$ and by lifting it up at $t = 7.5s$ and $t = 18s$. At $t = 20s$ the controller goes into saturation but comes back at $t = 21s$ because of the implemented antiwindup scheme (Åström and Wittenmark, 1997). Further, the subject was allowed to alter hip angle slightly during the ex-

Table 1 Controller design parameters ($A^+ = A, B^+ = 1$)

Controller	t_{rA_m}	ζ_{A_m}	$t_{r_{A_o}}$	ζ_{A_o}	k_p
C1	1s	1	0.9s	1	1
C2	0.5s	1	0.4s	1	1
C3	0.5s	1	0.4s	1	0.5

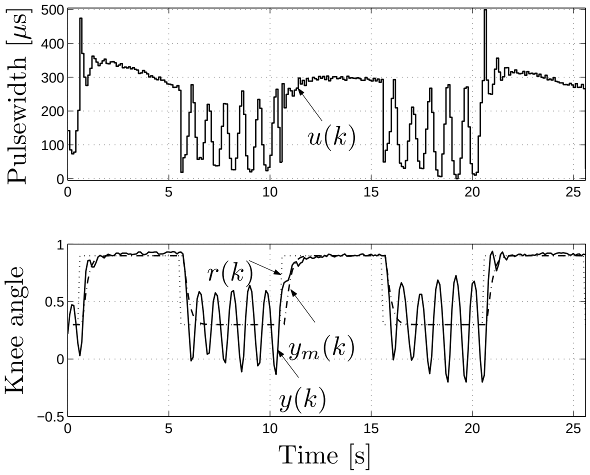

Fig. 6. Unstable closed-loop system with controller C2; r reference signal, y_m filtered reference signal (desired knee angle), y measured knee angle, u pulsewidth.

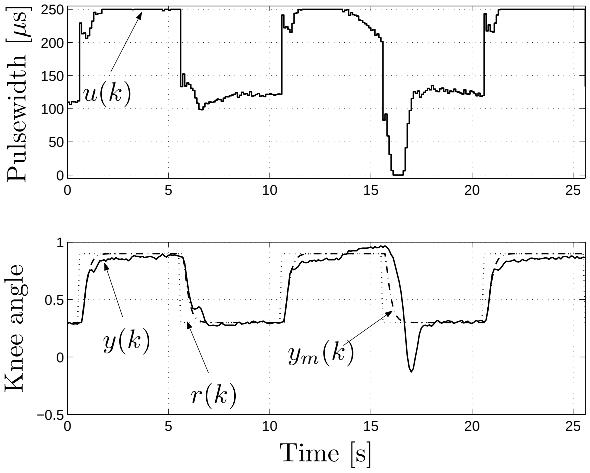

Fig. 7. Stabilised closed-loop system with Controller C3 which is identical to C2 but has $k_p = 0.5$. At $t = 13s$ the leg was lifted up as disturbance test.

periment. Changing the rise time of the reference model to $t_{r_{A_m}} = 0.5s$ (controller C2), the results shown in Fig. 6 were obtained. Now the closed-loop system becomes unstable for the equilibrium point $\Theta = 0.3$ because the system gain is too high with the fast controller specifications. Changing the open-loop gain as suggested above by the factor $k_p = 0.5$ (controller C3) stabilises the system but gives poor performance for upward movement of the leg (cf. Fig. 7). In this experiment the upper saturation limit in the controller was set to $u = 250\mu s$ so that the reference value for the angle $r = 0.9$ could not be reached. At $t = 13s$ the leg was lifted up as disturbance test.

4. CONCLUSIONS

It was shown that good tracking performance and disturbance attenuation for the knee joint movement can be achieved by controlled FES using

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surface electrodes. The use of a parameterized linear model and pole-placement design provides a fast way to determine a simple but reliable and robust controller. Closed-loop specifications can be easily formulated in the time-domain and are model independent. There are plans to continue this investigation with experiments on paraplegic subjects. The integration of this approach into problems like standing with a loaded knee joint (Wood *et al.*, 1998) will be investigated in the near future.

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