ADVANCES IN EXTERNAL CONTROL OF HUMAN EXTREMITIES IX

CLOSED LOOP CONTROL OF STIMULATION DURING STANDING

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ABSTRACT

One aspect in the control of walking using functional electrical stimulation is the stabilisation of the standing position. We study strategies for closed loop control of knee-angle during standing using quadriceps stimulation. The feedback parameters are knee-angle and angular velocity.
Due to the gravitation influence the system is open loop nonstable. To find a robust controller and an optimal feedback strategy the skeletal-muscle system was studied using state equations. The system, consisting of a three segment body model and a first order muscle model, was described by a set of first order linearized differential equations. Matrix calculations were used to determine the poles of the system. By means of root-locus plots the stability of the controller was judged for different feedback strategies. It was found that at least a PD-controller is needed to stabilize the standing position. To find the most robust controller, the sensitivity of the controller adjustment to changes in the muscle recruitment curve was determined. This was also done by means of root-locus plots. State space plots were used to find the influence of fysiological constraints to the controller adjustment. Computer simulations were used for verification.

INTRODUCTION

In the clinical application of Functional Electrical Stimulation (FES) the stabilisation of the standing position is usualy realised by open-loop stimulation of the Quadriceps muscles. On the occurance of fatigue or other disturbances the stimulation is adapted by hand. In this way the muscles are also more stressed than needed. By applying feedback of kinematical information from the body towards the stimulation equipment these problems may be overcome. In this way the reproducability can be improved, and the muscle force needed to maintain an upright position can be minimized.

At this moment in many centres research has started to develop closed-loop stimulation systems [1, 2, 3]. However in most literature the controllers are adjusted by trial and error. We study strategies for closed loop control of knee-angle during standing, with automatic compensation of inter and intra patientvariability.

The main problem in controlling the standing position while the knees are unlocked is its unstable behaviour due to the gravitation influence. Besides the recruitment curve of the muscle is highly nonlinear and is time-dependant partly due to fatigue [4].

The system which we consider consists of a controller, a muscle model and a skeletal model (figure 1).

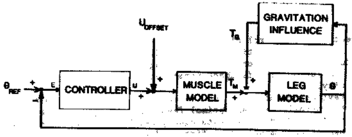

Figure 1: Block-diagram of the system, where:

θ_{ref} = desired knee-angle
θ = knee-angle
T_m = muscle torque over the knee joint
T_g = gravitation torque on the knee joint
U_{offset} = offset stimulation for compensation of T_g
u = controller output
e = error signal

To find a robust controller and an optimal feedback strategy we applied state-space methods, in which the model is described by a set of first order differential equations. Root-locus plots were used to estimate the influence of the different system parameters on the stability of the controller.

METHODS

The skeletal model

To describe the mechanical system during standing, it is modelled as a set of three coupled rigid bodies: underleg, upperleg and upperbody (figure 2).

The joints between the body segments are considered to allow rotation in the sagital plane only. Each segment is described by means of it's mass, centre of gravity and moment of inertia. The upperbody is considered to stay in a vertical position and vertically above the ankle joint. This means that all the angles in the model are determined by the knee-angle.

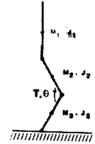

Figure 2:

The three-segment body-model in the sagital plane, where:

m_i - mass of segment i
J_i - moment of inertia of segment i
θ - knee-angle
T - muscle torque over the knee joint

The knee-torque T can now be described by the following differential equation using the Lagrangian formalisme:

$$T = \underbrace{[A+B.\cos(\theta)].(\ddot{\theta})}_{1} + \underbrace{C.\sin(\theta).(\dot{\theta})^2}_{2} + \underbrace{D.\cos(\theta/2)}_{3} + \underbrace{E.(\dot{\theta})}_{4} \qquad (1)$$

where:
1 - inertia torque
2 - centrifugal torque
3 - gravity torque
4 - damping torque
θ - knee-angle
A,B,C,D,E - coeficients containing the mechanical properties of the three segments (mass, moment of inertia, centre of gravity)

From (1) it can be seen that the dynamic behaviour of the system is knee-angle dependant. Besides it is important to notice the the influence of the gravitation. Our interest concerns specially the small-signal behaviour of the system for an equilibrium position near full leg-extension (θ=180 degrees). Linearizing equation (1) around equilibrium state θ=180° and $\dot{\theta}$=0 gives:

$$T = \underbrace{A'.(\ddot{\theta})}_{1} + \underbrace{D'.\theta}_{2} + \underbrace{E'.(\dot{\theta})}_{3} \qquad (2)$$

where:
1 - inertia torque
2 - gravity torque
3 - damping torque
θ - deviation of the knee-angle
A',D',E' - coeficients containing the mechanical properties of the three segments (mass, moment of inertia, centre of gravity)

The muscle model

Considering the small-signal behaviour of the muscle, the muscle force alters slowly and with a certain delay when the muscle activation is changed. As a first approximation this is modelled as a first-order transfer function. The static force is supposed to be linear for small changes. Additional influences of e.g. force-length and force-velocity effects are neglected.
The muscle transfer function which is used can be written as:

$$\frac{T(s)}{u(s)} = \frac{K}{(1 + s.d)}$$

where: $T(s)$ = torque
$u(s)$ = activation
K = gainfactor
d = time constant

The dominant non-linearity and time dependance of the recruitment can now be modelled as a time varying gainfactor K.

The controller

In the development of a suitable controller for the stabilisation of the standing position two important aspects have to be considered: first the possible changes in the dynamic properties of the system, mainly due to changes in the gain factor K of the muscle, secondly the influence of the gravitation.

While the system as shown in figure 1 is open-loop unstable, bodeplot methods can not be used to adjust the controller. Therefore we describe the behaviour of the system by means of the state-space method. The behaviour of the linearized system is described by means of a set of first order differential equations. The state-variables are the knee-angle θ, the angular velocity and the torque T over the knee-joint.
The state-equations can be written in matrix notation as follows:

$$\dot{X} = [A].X + [B].u \tag{4}$$

where: $X = [\theta, \dot{\theta}, T]^T$, the state-vector
$[A]$ = system matrix
u = control variable
$[B]$ = influence of the control variable u

Feedback can be incorporated considering the control variable u to be dependent of the state-vector X. In our case u can be seen as the muscle activation. In the skeletal-muscle system two state-variables can be measured and therefore we use output-feedback; i.e. a static control strategy only depending on knee-angle and the angular velocity.
Defining:

$$u = F_1.\theta + F_2.\dot{\theta} \tag{5}$$

where F_1 and F_2 are the control gains, we can write:

$$\dot{X} = [A'].X \tag{6}$$

where: $X = [\theta, \dot{\theta}, T]^T$, the state-vector
$[A']$ = system matrix of the closed-loop system

It can be derived that the eigenvalues of the system matrix $[A']$ are equal to the poles of the corresponding system [5]. In general the poles of a system describe the systems behaviour completely. The poles can be plotted in the s-plane. For a stable behaviour of the system all poles should be positioned at the open left half of the s-plane.

When applying variations to one of the system parameters the locus of the poles can be plotted in the s-plane as a function of this parameter, and changes in the stability of the system can be judged. In this way the robustness of the controller for variations in K can be visualized.

The calculation of the eigenvalues or poles of the system is done by means of a special computerprogram for matrix calculations.

RESULTS

Considering the linearized mechanical model of equation (2) and the muscle model of equation (3), various controllers can be studied with respect to the stability in controlling the knee-angle θ after small disturbances.

For a Proportional-controller the feedback signal u is a function of the knee-angle θ, and can be written as: $u = F_1.\theta$.
Figure 3 shows the root-locus plot of the system with P -controller, for $F_1.K$ varying from 0 to infinity. The position of the poles for the open-loop situation (F_1=0) is marked by x. It can be seen that one of the poles is always in the right half-plane, so the system is unstable for all values of $F_1.K$.

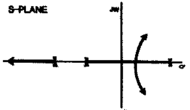

Figure 3: Root-locus plot for variations in gain, for the linearized system of figure 1 with P-controller, where x gives the position of the poles in the open-loop situation.

For a PD-controller the feedback signal u is also a function of the angular velocity (see (5)). In figure 4 the position ofthe poles of the system for the open-loop situation (F_1=0, F_2=0) is marked by x. From figure 4 it can be seen that when using a PD-controller values for F_1 and F_2 can be found to obtain a stable system (all poles situated in the open left half-plane).

To find the most robust controller adjustment with respect to variations in K, the influence of F_1 and F_2 is investigated. Therefore, the poles are positioned in the s-plane for a nominal value of K to obtain a stable system (marked N). The according values of F_1 and F_2 are calculated, and the root-locus plot is used to determine within what range of variations of K the system will be stable. Repeating this for different values of the poles, the optimal values for F_1 and F_2 can be found.

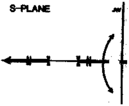

Figure 4: Root-locus plot for variations in the muscle gain K for the linearized system of figure 1 with PD-controller, where x gives the position of the poles in the open-loop situation, and N the position of the poles for a nominal value of K, F_1 and F_2.

To evaluate the characteristics of a controller adjustment, the responses of the system to disturbances in knee-angle or angular velocity are examined. This is done by means of computer simulations on the nonlinear system. For modelling mechanical systems the bondgraph technique is suitable [6]. Therefore we use the simulation program TUTSIM, which includes the bondgraph elements. State-space plots as well as time plots of the response of the system are made. As an example figure 5 shows the response of the system to disturbances when using a PD-controller.

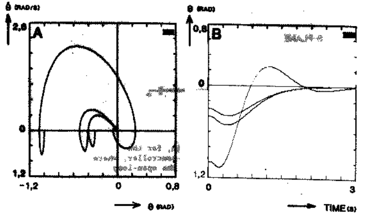

Figure 5: Responses of the system with PD-controller to disturbances in knee-angle and angular velocity around the state of equilibrium.
A: state-space plot
B: time plot

From figure 5 it can be seen that the response of the system is changing for large disturbances in the knee-angle and slight oscilations occur. Although the system is still stable, it is out of its linear area. This is due to the fact that the system is linearized and valid for small-signal behaviour only. Outside the linear area the controller is over-compensating the torque of gravity.

Apart from adjusting the controller to be most robust to alterations in muscle gain, fysiological constraints also have to be considered. Problems will occur when the controller is adjusted so that, after a disturbance, the muscle force or knee angle will need or will become values out of the fysiological range. The system can be considered to be open loop in that case or can be damaged. Therefore, with respect to the state variables, areas in the state space can be indicated which must be avoided or even can not occur (figure 6). Maximum and minimum knee angle and the constraints for the stimulation current with respect to the minimum and maximum muscle force can be indicated as lines.
For a PD-controller e.g.: $U = F_1.\theta + F_2.\dot{\theta} < U_{max}$

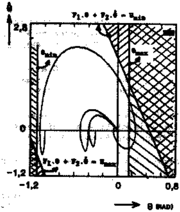

Figure 6:

State space plot of figure 5. Indicated are the limits of the stimulation current and the range of the knee angle.

CONCLUSIONS

To find an optimal control strategy to control the standing position, the classical methods used to adjust a controller (e.g. by means of bodeplots) can not be used.
Using the state-space methods we are able to describe the small signal behaviour of the system and determine the system parameter sensitivity of the controller. A preleminary conclusion might be that a PD-controller is needed to stabilize the system, when a first-order muscle model is used. Besides the most fast controller seems to be also the most robust to changes in the muscle gain factor K.

The system was linearized in the equilibrium position. This means that the controller characteristics are limited to small changes in the state-variables. To enlarge the linear area and the stability of the controller feedback linearization can be applied; i.e. nonlinear state feedback which makes the overall system linear in the whole state space. A major problem might occur when the process of linearization is not performed accurately. Further research on this matter has to be performed.

Special account has to be directed also towards determining the influence of fysiological constraints to the optimal adjustment of the controller. A first conclusion can be that when the desired knee angle is near full leg extension, overshoot has to be avoided. In that case the system has to be critically damped and the poles of the system must be real.

REFERENCES

Jaeger R.J., 'Design and Simulation of Closed-Loop Electrical Stimulation Orthoses for Resoration of Quiet Standing in Paraplegia', J. Biomechanics, VOL. 19, No. 10, pp. 825-835, 1986.

Wilhere G.F., Crago P.E. and Chizeck H.J., 'Design and Evaluation of a Digital Closed-Loop Controller for the Regulation of Muscle force by Recruitment Modulation', IEEE Trans. on Biomed. Eng., VOL BME-32, No. 9, 1985.

Petrofsky J.S., Phillips C.A. and Stafford D.E., 'Closed-Loop Control for Restoration of Movement in Paralyzed Muscle', Orthopedics, VOL. 7, pp. 1289-1302, 1984.

4. Stanic U., Trnkoczy A., 'Closed-Loop Positioning of Hemiplegic Patient's Joint by means of Functional Elecrical Stimulation', IEEE. Trans. on Biomed Eng., VOL. BME-21, No. 5, 1974.

5. Luenberger D.G., 'Introduction to dynamic systems', John Wiley, New York, 1979.

6. Karnopp D.C., Rosenberg R.C., 'Systems dynamics: a unified approach', John Wiley, New York, 1975.