CONTROL OF A HYBRID MOTOR PROSTHESIS FOR THE KNEE JOINT

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Abstract: Functional Electrical Stimulation (FES) is used for gait restitution in paraplegic patients. One of the major problems is the heavily increased muscle fatigue due to unphysiological stimulation. As a possible solution, this paper introduces a hybrid neuroprosthesis for the knee joint, where muscle stimulation is complemented by a motor-driven exoskeleton. A control concept based on model predictive control is designed that employs both actuators cooperatively and varies their participation corresponding to their capabilities.*Copyright ©2005 IFAC*

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1. INTRODUCTION

Motor neuroprostheses for the lower extremities using Functional Electrical Stimulation (FES) already allow the realisation of functional gait (Fuhr, 2004). However, increased muscle fatigue due to unphysiological recruitment poses a serious problem and limits the achievable gait duration.

To relieve the strain on the muscles, an additional motor supported orthosis (exoskeleton) can be applied. This combination of exoskeleton and electrical stimulation is denominated “hybrid neuroprosthesis” (Andrews and Baxendale, 1988; Popovic and Tomovic, 1989). The stimulated muscles of the patient are relieved and the necessary aid of the upper trunk is noticeably reduced. Consequently, the muscles fatigue less rapidly and the gait duration increases.

This paper introduces a new approach for control of hybrid neuroprostheses that focuses on a cooperative control concept for the redundant system. Both actuators FES and external drive are placed in a single control loop with a model predictive controller. An adaptive distribution of the necessary torque (via a convex combination) between the two actuator types reduces muscle fatigue in comparison to mere FES. The system takes advantage of the reliable and fast performance of the motor, but exploits the muscles as much as possible when a predominantly constant torque is required. Thus, a better performance is accomplished without excessively increasing motor participation and energy consumption.

The control is implemented and tested on a single joint, the knee. To generate reference trajectories, an ergonomic patient interface is used. The patient wears a data glove as input device in order to control his movements. As the motion of the finger articulations bears an obvious analogy to the lower extremities, a highly intuitive control is realised. The knee angle tracks the index bend angle.

Section 2 outlines the system structure and the model of FES-stimulated muscles that has been

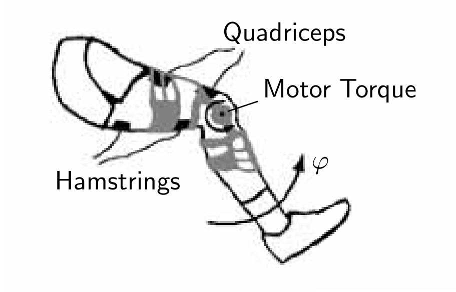

Fig. 1. Motor-driven orthosis and electrically stimulated muscles (quadriceps and hamstrings) to control the knee bend angle φ.

used in this work, section 3 treats the general control, section 4 explains the proposed concept to distribute the manipulated variable between the two actuators. Results are presented in section 5.

2. SYSTEM STRUCTURE AND MATHEMATICAL MODEL

2.1 The Hybrid Motor Neuroprosthesis

The system has 3 components: An exoskeleton, that consists of motor and orthosis, a stimulator, and a data glove, that serves as user interface. The motor, in addition to the stimulated muscles, exerts a torque on the patient's knee (Fig. 1).

For muscle stimulation, a neurostimulator from Krauth + Timmermann GmbH called "MOTIONSTIM8" is used, which can be controlled online in a scientific mode via a computer's serial port. The Max-Planck Institute in Magdeburg provides a C-library which is used in this work. The exoskeleton is based on a knee orthosis and is driven by a DC-motor from Maxon in combination with a gear box. As a patient interface the data glove "P5Glove" from Essential Reality is used. This product, that was originally developed for the game market, is accessed via USB port. The producer provides an open source C-library to access the glove in Linux.

The knee angle is measured with an encoder attached to the motor, the angular velocity is derived via differentiation. These variables are fed back to the control algorithm.

The operating system used for this application is Real-Time-Linux on a Pentium 3 Processor. The control was implemented using the Real-Time-Workshop from Matlab/Simulink (Version 6.0).

2.2 Mathematical Model of Leg and Muscles

The lower thigh is modelled as a pendulum. The degree of freedom of the ankle joint is neglected. The anthropometrical data provided by Winter

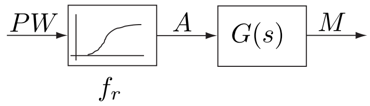

Fig. 2. Muscle response to FES is modelled with a nonlinearity f_r that correlates pulse width PW and activation level A, followed by a linear transfer function $G(s)$ to the torque M.

(2004) allows to estimate model parameters like the weight of the joints in relation to the entire body mass, as well as centre of mass and inertia of the segments.

In FES, the correlation between stimulation parameters and exerted torque by muscle contraction is strongly nonlinear and time-variant. This work adopts a simple model (Winters, 1990), which calculates the muscle force in function of the so-called "activation level" A (see Fig. 2). The activation level is a nonlinear function of the pulse width PW of the stimulation pulse, where frequency and current are held constant. This nonlinear function is called "recruitment curve" and is experimentally obtained. The transfer function that describes the torque M produced at an activation level A is assumed to be a critically damped second order delay element with dead time (damping D=1):

$$G_{\mathrm{FES}}(s) = \frac{M(s)}{A(s)} = \frac{\omega_m^2}{s^2 + 2\omega_m s + \omega_m^2}\, e^{-sT_t} \quad . \tag{1}$$

In conformity to literature, the dead time T_t is assumed to be 25 ms and the muscle's eigen angular frequency ω_m is identified as 9.4 rad/sec.

3. CONTROL CONCEPT

The control goal is that the knee angle tracks the bend angle of the index. The system is redundantly actuated by the two actuators, which both exert a torque on the knee joint. The control strategy must cope with a fundamental problem of redundant systems: identical results can be achieved with different participation of the redundant actuators.

It is desirable to use the human muscles as much as possible to provoke a training effect. Nevertheless, both actuators should be exploitet according to their capabilities and restrictions: The motor reacts quickly, whereas the muscles have to receive a stimulation signal with slow changes. The concept therefore splits the predicted necessary torque in function of its frequency domain characteristics: nearly constant torques are realised by the muscles and high-frequency torques by the motor. A constraint is the stipulation that both actuators must always produce a torque with the

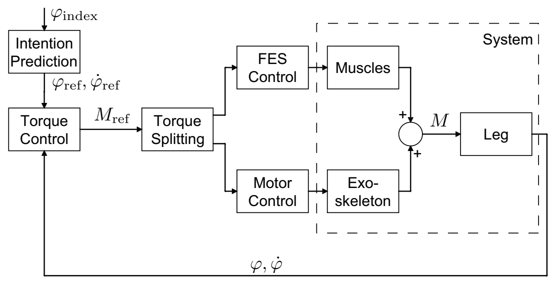

Fig. 3. Based on the predicted reference angle trajectory, the model-based torque controller predicts an optimal torque trajectory, which thereafter is split up to the two actuators.

same orientation, thus reducing energy consumption. With this split-range concept, the overall system is optimised with regard to performance, energy consumption and muscle fatigue.

The chosen control strategy is based on model predictive control (MPC). In order to force the control variable to track a certain future reference trajectory, a future sequence of the manipulated variable is predicted. This is done via an optimisation procedure (Bryson, 2002), where control performance and control effort are traded off. In each time step, the trajectory of the manipulated variable is re-calculated and only the first value is emitted. The considered time span for the optimisation is denominated "prediction horizon".

The control concept (Fig. 3) divides the task into four subproblems: prediction, torque control, torque distribution and actuator control.

3.1 Prediction of the reference trajectory

A future reference trajectory for the knee angle is provided for the predictive controller, which is based on the momentary reference angle and its history. Simple rules have been implemented: The historical reference angle trajectory is extrapolated, limited by the knee angle constraints. This consideration of constrains exploits a special advantage of predictive control in this application: The maximally allowed knee angle cannot be exceeded by overshoot, since the model predictive controller is aware of the limitation in advance and can decrease the torque adequately.

3.2 Torque Control

A model-predictive controller calculates the future torque trajectory in order to force tracking of the knee angle. The optimisation is based on a model with idealised (proportional) actuators and does not consider the actual realisation of the torque. The calculated future torque trajectory forms the base for the next two elements of the control loop: torque distribution and predictive FES control. The manipulated variable of the torque controller (the torque M) represents the reference variable for the FES controller.

3.3 Actuator Control

The DC motor can be controlled by standard techniques. In contrast, the muscles control is considerably more complex because of strong nonlinearities and delay. A subordinate control compensates the nonlinearity (see section 2.2) by an inverse muscle recruitment curve f_r^{-1}, which determines the pulse width corresponding to the desired torque, as suggested by Hunt (1998). After the compensation, the open loop relationship between desired and actual torque is described by a second order lowpass with dead time (see Eq. 1). As a result of the predictive strategy, stimulation pulses can be anticipated in order to decrease phase delay and dead time. Now the manipulated variable in the predictive optimisation procedure is the pulse width, whereas control variable is the torque produced by the muscles. Since the exerted torque cannot directly be measured, it is estimated by an observer that contains a model of the entire system (muscles, motor and leg). Angle and angular velocity of the leg are fed back to the observer to reduce estimation error. This subordinate controller ensures the simultaneous activation of motor and FES.

4. TORQUE DISTRIBUTION

By comparison of the two actuators, the motor has superior control performance. However, there are two reasons to integrate the muscle force as much as possible (limited by fatigue): a positive training effect for the patient and a reduction of the consumed electrical energy.

A very simple approach to distribute the torque is a convex combination via the multiplication with a fixed factor k: the stimulator realises the part $k{\cdot}M_{\text{ref}}$ and the exoskeleton the part $(1{-}k)M_{\text{ref}}$. Under the supposition that the stimulator in combination with compensator and controller can be considered a proportional element, there is no delay between motor torque and muscle torque, both actuators work in the same direction.

A different approach for distribution is based on filtering of the reference torque. Since the stimulated muscles only work up to a frequency of approx. 2 Hz, it is straightforward to perform filtering and to give the motor the higher-frequency part of the signal and the stimulator the lower one. One major objection to this distribution is that it inhibits the cooperative work of the two actuators,

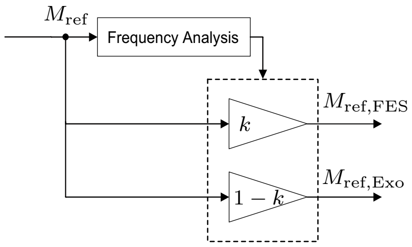

Fig. 4. The reference torque is split up to the FES-stimulated muscles and the motor-actuated exoskeleton via an adaptive, frequency-dependent convex combination.

i.e. it is not guaranteed that both actuators work in the same direction.

With a frequency dependent variable factor, as illustrated in Fig. 4, the advantages of filtering and of distribution with a factor can be combined.

4.1 Adaptive Distribution

The requirements of the adaptive distribution are:

- Both actuators exert a torque with the same orientation.
- In the presence of a predominantly low-frequency signal the stimulator is preferred, whereas high-frequency torques are realised by the motor.
- The transitions between the modes should be smooth and gradual to allow the muscles to slowly build up their strain.

Therefore, the factor k has to meet the following criteria:

- $k \in [0, 1]$.
- The value of k depends on the frequency of the desired torque, predominantly high frequencies lead to a value near 0, low frequencies result in a value near 1 (high participation of the stimulator).
- Changes in frequency lead to changes in k in time, in order to increase the corresponding actuator participation, but not too jerkily.

The expression “in time” is defined in the following sense: In the case of an abrupt (predicted) frequency change in a sinusoidal reference signal, the actuator distribution that corresponds to the new frequency is reached at the instant when the frequency change takes place. This requirement can be achieved via a predictive control strategy, because this way the future reference torque trajectory is known and can be used for an anticipated actuator preparation.

In order to analyse a signals's frequency distribution, Fast Fourier Transformation (FFT) could be used. However, simulations have shown that this method, in order to produce meaningful results, demands for a large number of supporting points, which in turn requires an excessively long prediction horizon. Therefore, two more robust time-domain approaches to determine the factor k have been developed, which are based on integration.

4.2 Frequency Analysis by Integration

The introduced method to calculate the distribution factor is based on filtering and consequent integration of the predicted reference torque. By evaluation of the relationship between lowpass-filtered signal and original signal, a characteristic is derived which indicates the portion of low frequencies.

As the signal is predicted and therefore entirely known at the instant of filtering, the application of a so-called “zero-phase” discrete filter is possible, achieved by backward and forward filtering with an IIR lowpass-filter.

A first approach for the calculation of k is:

$$k = 1 - \frac{\int_0^T |w - w_{\mathrm{filt}}| \mathrm{d}t}{\int_0^T |w| \, \mathrm{d}t} \quad , \tag{2}$$

where w_{filt} is the zero-phase-filtered reference signal w (in this case the reference torque M_{ref}) and T the prediction horizon.

The demand for the limitation of k between 0 and 1 is fulfilled: Since w_{filt} is the result of zero-phase filtering of w with a filter magnitude $< 1, \forall\ t \in [0, T]$:

$$\begin{aligned} |w_{\mathrm{filt}}| \leq |w| \quad &\wedge \quad \mathrm{sign}(w_{\mathrm{filt}}) = \mathrm{sign}(w) \\ \Rightarrow \quad |w - w_{\mathrm{filt}}| \quad &\leq |w| \\ \Rightarrow \quad \int_0^T |w - w_{\mathrm{filt}}| \mathrm{d}t \quad &\leq \int_0^T |w| \mathrm{d}t \\ \Rightarrow \quad k \in [0, 1] & \end{aligned} \tag{3}$$

In case that w is a sinusoidal signal, i.e. it contains only one, constant frequency, w_{filt} equals:

$$w_{\mathrm{filt}} = w \cdot |G_{\mathrm{filt}}| \quad , \tag{4}$$

where $|G_{\mathrm{filt}}|$ is the magnitude of the zero-phase filter for the considered frequency. Substituting this for k in Eq. (2), the immediate result is:

$$k = |G_{\mathrm{filt}}| \quad . \tag{5}$$

For this special case the frequency evaluation represents a zero-phase lowpass-filtering. The stimulator receives the filtered reference signal and the motor the remainder.

The third criterium remains to be analysed. It demands little oscillation of k, but nevertheless reaction in time on upcoming frequency changes. One way to reach little oscillation of k is a large horizon T considered for the integration. However, a prediction horizon too far in the future impairs the reliability of the predicted data and requires excessive calculation time. Lowpass-filtering of k also limits oscillation, but causes an undesirable phase delay. This is opposed to the demand to adapt the actuator participation in time to the given situation. A small modification of Eq. (2) improves the considered behaviour significantly, as will be shown in the following section.

4.3 Time Weighted Integration

In order to meet the third requirement in section 4.1, k must adapt *gradually* to predictable changes. This can be achieved if values of w are given more weight depending on their vicinity to the present time. Thus, a change appearing at the end of the prediction horizon has little effect on k at first due to the small weight. Then its impact gradually increases with advancing time.

Hence, a *time-weighted integration* accomplishes the required gradual adaption to predictable changes. The modified calculation method is:

$$k = 1 - \frac{\int_0^T |w - w_{\text{filt}}| \cdot (T-t)\mathrm{d}t}{\int_0^T |w| \cdot (T-t)\,\mathrm{d}t} \quad . \qquad (6)$$

It can be shown that the limits of k and also its behaviour at constant frequency of w are not altered by this modification.

4.4 Dimensioning

Now the question for the quantitative parameters in the calculation arises. The following criteria are defined as desirable:

- The stimulator is used predominantly up to 2 Hz, afterwards the motor takes over.
- Each actuator is always employed with a minimum participation, which has been chosen to be 20%.

For the design, Eq. (5) for the behaviour of k in the presence of a sinusoidal reference signal proves helpful. It demonstrates that the filter defines the behaviour of the torque distribution. The specifications mentioned above result in a filter with an magnitude decreasing from 0.8 at 0 Hz to 0.2 at 2 Hz. This is a decrease by 75%; a filter scaled with 0.8 must therefore have an magnitude of 1/4 at 2 Hz. As the filter order of a phaseless filter doubles compared to the original filter and the magnitude decreases by square in relation to it, the original (Butterworth-) filter decreases 6 dB. After 2 Hz, the factor k is held constant at 0.2, so that the stimulator always receives 20% minimum. The resulting dependance of k on the reference signal frequency is displayed in Fig. 5.

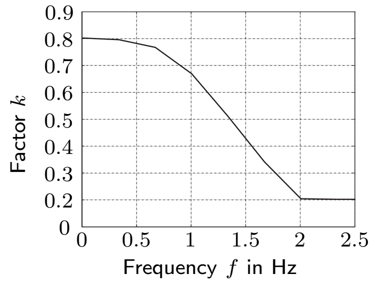

Fig. 5. The parameter k as a function of the frequency f of a sinusoidal reference signal. Both actuators always participate with a minimum portion of 20%.

5. TESTING AND SIMULATION

The purpose of the time-weighted integration is to avoid strong oscillations of k in the presence of abrupt changes of the reference torque. To display the impact of this modification, Fig. 6 shows the course of k as a response to a sawtooth-formed torque signal. Within a second the torque arises to its maximum value and rapidly descends afterwards. At t_1, the abrupt change appears in the prediction horizon. One can see clearly that in the case of non time-weighted integration, the factor k falls just as abruptly. The motor is employed jerkily and the muscles are relieved, although there is still enough time for a gradual shift. Using time-weighted integration this behaviour does not occur, the factor k changes its value with considerably less oscillation, but eventually reaches the same final value. The time-weighting of the integration thus accomplishes the task of avoiding abrupt changes of k.

Fig. 7 shows the performance of the variable distribution in simulation. The employed reference signal for the knee angle is a sine sweep. Thus, the torque reference trajectory calculated by the torque controller also shows a continuous increase in frequency. The amplitude first rises due to the larger necessary torques to produce higher accelerations, but later decreases due to the optimisation procedure which trades control effort

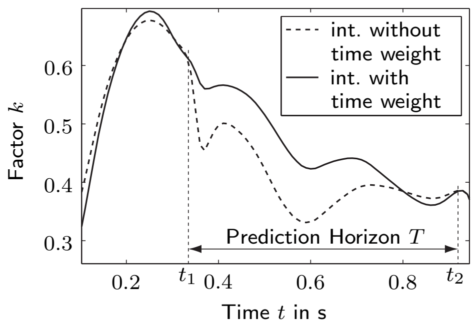

Fig. 6. With time weighted integration, the factor k adapts more gradually to an upcoming instantaneous frequency change of the reference signal w (at t_2), which is predicted at t_1.

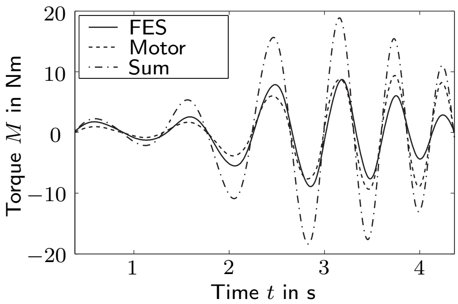

Fig. 7. Torque distribution: The motor portion increases with the frequency of the reference torque, the muscle portion decreases.

against regulation performance: When the system can no longer track the desired trajectory in an acceptable manner because of the intertia of the leg, the torque controller decreases control effort. The figure also shows the shift in the participation of the two actuators on the entire torque: At first, the motor has a small percentage, later at higher frequencies it takes over more and more. Thus, the requirement of a distribution adapted to the actuator capabilities is accomplished.

In control performance, the variable torque distribution does not show any essential advantage compared to the fixed distribution. In control effort, however, the variable distribution leads to a significant reduction of approx. 5-10% (muscles + motor). This can be explained considering the adaptive strategy: The model-based predictive feedforward control calculates torques which are adequate to perform a desired motion, assuming that they are realised ideally. However, the bandwidth of the muscle dynamics is constrained and therefore the muscle torque cannot follow high-frequency reference signals. If the control strategy, as in fixed distribution, does not consider the properties of the muscles, they receive a reference torque which they cannot realise. In this case, the control error of the knee angle accumulates, provoking the torque controller to calculate a reference torque trajectory with higher amplitude. This way the motor is engaged more and more. Although this does not affect significantly the control performance, control effort and thus energy consumption increase. Applying adaptive distribution, the muscle participation is reduced in such challenging situations, but intensified in situations when the muscles can meet the requirements. That way the control error of the closed-loop control and thus the control effort decrease, which results in less energy consumption.

Opposed to these savings there is the additional consumption of calculation time not to be neglected, as well as a stability risk arising with model uncertainties, which are considerable in the case of FES. The calculation time does not only increase by the integration itself, but also by the still considerably prolonged prediction horizon of the torque controller in order to avoid oscillations of k.

6. SUMMARY

A hybrid neuroprosthesis for the knee joint was investigated, which controls the motion of the knee employing electrically stimulated human muscles and a motor-driven orthosis. In order to control this redundantly actuated system, a control concept was designed and tested. The main aim was the operation of the hybrid neuroprosthesis in a manner that employs both actuators cooperatively and corresponding to their specific capabilities.

For the realisation of this system a method of frequency evaluation in the time domain was derived. The results are promising and show a considerable reduction of overall energy consumption.

The results of this work concerning control of redundant actuators are transferable to a complete hybrid neuroprosthesis for the lower extremities.

7. ACKNOWLEDGMENTS

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