FEEDBACK CONTROL OF UNSUPPORTED STANDING

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Abstract

This paper presents the results of continuing work on feedback control of unsupported standing in paraplegia. Our experimental setup considers a situation in which all joints above the ankle are braced, and stabilising torque at the ankle is generated by stimulation of the plantarflexors. A previous study showed that short periods of unsupported standing with paraplegic subjects could be achieved. In order to improve consistency and reliability of unsupported standing we are currently investigating several modifications to the control strategy. The paper reports progress towards this goal.

Introduction

We are investigating the use of feedback control systems which enable paraplegics to stand without the support of their hands and arms. Feedback is used in combination with functional electrical stimulation (FES) of the ankle plantarflexor muscles to provide stabilisation of the upright posture. We previously proposed a nested control structure in which an inner loop controls the moment generated at the ankle [1,2]. An outer-loop controller regulates the angle of body inclination. In our experimental setup all joints above the ankle are locked using a special body brace, allowing us to isolate the effects of the artificial control system from the remaining motor control actions of the intact upper body. While we believe that functional systems must integrate the natural and artificial controllers, our experimental setup allows us to study the potential benefits and fundamental limitations of the artificial system. Matjacic and Bajd have recently carried out a study of unsupported standing in which the upper body is free to move [3].

The previous study showed that our approach is feasible; a neurologically-intact subject can be stabilised for long periods of time, and the controller is able to maintain stability in the face of significant disturbances (pushing and pulling the subject, raising and lowering the arms while holding weights etc.). The study also showed that short periods of unsupported standing could be achieved with a paraplegic subject [4]. We concluded that the principal limitations to the approach included limited power, rapid fatiguing, and significant spasticity of the paralysed muscles. While these limitations are dependent on the condition of the particular subject in question, the underlying parameterisation and design of the artificial controller has a crucial effect on the length of time during which successful standing is achieved. In short, the controller must be robust enough to deal with these sources of uncertainty and disturbance, and must maintain upright postural stability as reliably as possible.

We are currently investigating a number of modifications to the control structure proposed in [2], with the aim being to improve the consistency and reliability of unsupported standing. Preliminary results are reported in the paper.

Methods

To perform dynamic tests of unsupported standing, an apparatus called the "Wobbler" was constructed. Full details of the construction and functionality of the Wobbler are given elsewhere [5]. The Wobbler apparatus allows measurement of the moments generated at each ankle, and the angle of inclination of the body is also measured. A nested control structure for unsupported standing is shown in figure 1. An inner-loop controller regulates the ankle moment. The outer-loop controller regulates the body inclination angle by providing a desired moment (reference signal) for the inner loop.

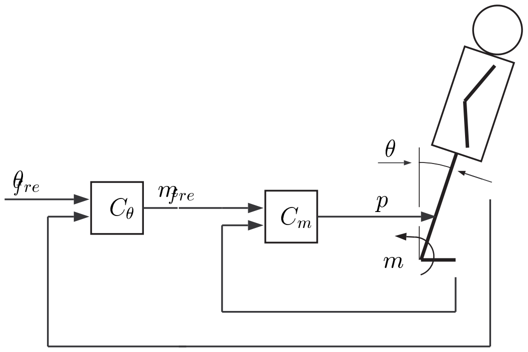

Figure 1. Nested control loop structure. θ inclination angle; m ankle moment; p pulsewidth

Figure 2. Subject standing in the Wobbler

Figure 2 shows a subject standing in the apparatus.

In our previous study [2,4] two separate controllers were implemented for left and right ankle moment. In this case half of the total desired moment was demanded from each side. Since the results of paraplegic standing showed very large differences in the moment-producing capacity of the left and right sides, our current approach is to implement a single moment controller which aims to generate the desired total ankle torque. This is achieved by sending an identical level of muscle stimulation to both sides. Design of the moment controller is based upon identification of the dynamic response from this single stimulation level to the total moment output. This modification means that if one muscle is stronger than the other then this muscle will automatically make a higher contribution to the total moment (since the left/right stimulation is equal). In the previous approach an equal moment is demanded from each side, without regard to the ability of the respective sides to generate the desired moment. One effect of this, seen in the results with paraplegic subjects, was that one side would typically be operating in saturation (i.e. full stimulation) while the reserve of the stronger muscle on the other side was not utilised.

A further significant difference in the current series of experiments is that we are now using a closed-loop pole assignment control strategy for inner and outer loop design. The approach outlined in [2] used an LQG (Linear-Quadratic-Gaussian) design. A feature of the LQG method is that the closed-loop response achieved with a given set of control design parameters depends on the parameters of the model used for control design. With pole assignment, it is the desired closed-loop response itself (e.g. specified in time-domain terms of risetime and damping) which is the design target. Thus, the nominal closed-loop response is independent of the plant model. Typically, the identified models are second order [6].

Results

Results of standing experiments with a neurologically intact subject are shown in figures 3 and 4. The sampling time is 50ms in both control loops. The muscle activation is varied by the pulsewidth of the stimulation pulses, where a pulsewidth of 500µs corresponds to a fully activated muscle. The outer loop controller is designed with a closed-loop risetime of 1s and an observer rise time of 0.7s.

Control results with a sufficiently fast inner loop controller (closed-loop risetime 0.2s, observer rise time 0.15s) are shown in figure 3. After an initial settling period (until t=5s), the inclination angle of the subject follows the reference angle very accurately.

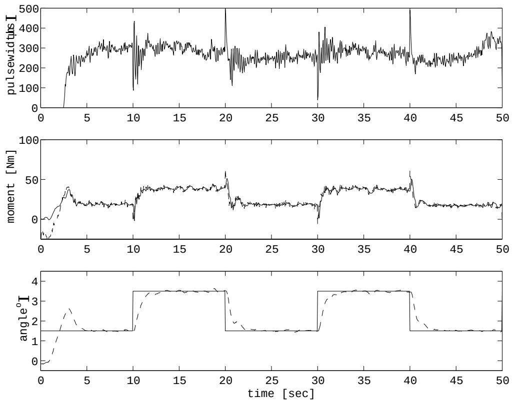

Figure 3. Result with fast inner loop controller. For moment and angle, the solid lines show references and the dashed lines denote measured values.

The risetime of the inner control loop is a crucial design parameter as the delay introduced in the outer loop by a slower inner loop controller can destabilise the system. This is illustrated by the experimental results shown in figure 4. Here the inner loop controller (closed-loop rise time 0.3s, observer rise time 0.15s) is too slow to enable stable standing, and the outer loop oscillates.

Discussion

The results show that unsupported standing can be achieved by stimulation of the plantarflexors in a feedback control setup. As the body dynamics represent an unstable inverted pendulum, it is important to obtain a fast response from the inner moment control loop to ensure stability of the outer loop. If the inner loop is too slow the delay introduced in the outer loop can lead to an overall unstable system. This result implies that the inner loop controller design is aimed at making the dynamics of the closed loop as fast as physically possible. We also observed in our experiments that it is more difficult to achieve a response fast enough for safe standing when the muscles are fatigued or weak.

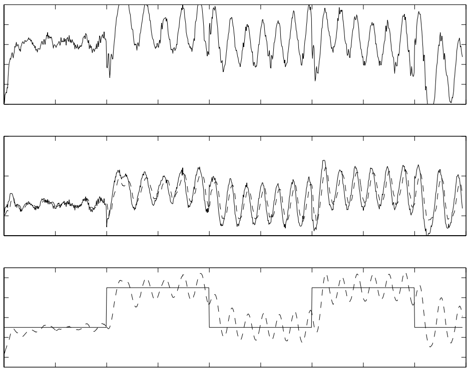

Figure 4. Result with slow inner loop controller. For moment and angle, the solid lines show references and the dashed lines denote measured values.

Conclusions

It was shown that unsupported standing can be achieved with a stabilising torque from the plantarflexors obtained by surface stimulation and feedback control. The importance of a fast and controlled moment generation from the muscles for stable standing was illustrated.

The use of pole assignment for the design of the controllers provides a way to consistently achieve the desired closed loop properties as the design parameters are directly related to characteristics of the closed control loop.

The use of a single stimulation level for right and left ankle ensures that both muscles are activated equally, and that the required moment can be obtained robustly. The identification of the muscle dynamics is simplified as only a single plant needs to be identified which combines the left and right ankle.

We plan to continue this investigation with experiments with a paraplegic subject in the coming months.

References

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