Control of Ankle Joint Stiffness using FES while Standing

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Abstract - *We have been developing feedback systems for control of unsupported standing in paraplegia. In the present paper we report results of an investigation of ankle stiffness control via functional electrical stimulation (FES), motivated by the need for sufficient ankle stiffness within unsupported standing schemes which aim to harness voluntary and stabilising motor control inputs from the upper body.*

The work was carried out with intact subjects using apparatus in which the subject stands with all joints above the ankles braced, and where ankle moment is provided via FES of the ankle flexor and extensor muscles.

The results show: (i) that accurate ankle stiffness control, up to the fundamental strength limits of the muscles, can be achieved with controlled FES; (ii) that ankle stiffness control has the potential to ease the task of stabilising upright posture by application of additional upper-body forces.

Keywords: Feedback Control, Stiffness Control, Function Electrical Stimulation, Unsupported Standing, Posture Control.

1 Introduction

This paper presents new results from an investigation of feedback control of ankle joint stiffness during standing, using functional electrical stimulation (FES) of the plantarflexor and dorsiflexor muscle groups. In previous work we have investigated *dynamic* control of unsupported standing via FES of the calf muscles only [2]. That work was carried out using an apparatus known as the "Wobbler" in which the subject stands with all joints above the ankles braced and does not use the arms for support [1]; there is therefore no voluntary control input from the subject.

Matjačić *et al* [3] have investigated control of unsupported standing using a Mechanical Rotating Frame (MRF). In contrast to the Wobbler, the MRF allows voluntary motion of the subject's upper body (e.g. using the trunk muscles) while the lower limbs are braced in a frame. The frame is controlled by a hydraulic actuator at the ankle joint. Stable posture is achieved by combining artificial control of the hydraulic joints with the subject's voluntary control input from the upper body. With this 2-link arrangement it has been shown that stability is achieved provided that a certain level of static stiffness (approximately 10Nm/deg) is implemented at the ankle joint.

This motivates our investigation of ankle stiffness control using FES: can ankle stiffness be achieved using FES of the ankle flexor and extensor muscles as an alternative to the hydraulic actuators in MRF-like devices? As a first step this investigation is being carried out in the Wobbler apparatus. Thus, our initial goal has been to determine the accuracy with which ankle stiffness control can be achieved with FES.

In the Wobbler arrangement *external control* must be applied during stiff standing experiments to maintain postural stability. In this work external control moment has been applied by the experimenter, and a second aspect of this investigation has therefore been to determine the ease with which the experimenter can stabilise the subject for a range of values of ankle stiffness.

2 Methods

2.1 Apparatus and Subjects

The Wobbler apparatus is described in detail in [1]. While standing in the Wobbler the subject wears a custom-fitted body shell which locks the knee and hip joints, allowing motion only around the ankle joints. For safety four light ropes are attached to the shoulders of the body brace and from there to a frame attached to the ceiling. When the ropes are tight the body cannot move. The ropes can be slackened sufficiently to allow movement in the sagittal plane within predefined limits. A string attached to the body brace at shoulder level is wound round a pulley attached to a potentiometer placed well behind the subject. This potentiometer is used to measure the inclination angle.

The subject's feet are positioned in footboxes connected to a shaft aligned with the ankle axis which can be driven by an electric motor. When the motor is switched on a gearing mechanism induces a rocking motion in the shaft and the feet are "wobbled". The shaft angle is measured by a second potentiometer.

The results reported here are for a fit and healthy male subject with no neurological deficit.

2.2 Control strategy

The stiffness control structure is shown in figure 1. M represents the dynamics of the stimulated muscles and B are the body dynamics of the inverted pendulum. The strategy uses a desired stiffness k^s_{ref} (specified by the experimenter) and a measurement of ankle-joint angle θ to compute the required ankle moment $m_{\text{ref}} = k^s_{\text{ref}}\theta$. This required moment is achieved by a feedback controller C_m which uses a measurement of total ankle moment m and adjusts the dorsiflexor/plantarflexor stimulation intensity p appropriately. m_e is an externally-applied moment acting on the body. The moment controller C_m is

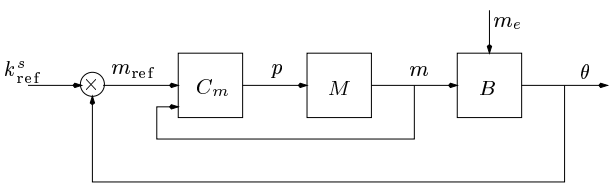

Figure 1: Stiffness control structure. The control structure incorporates a switching mechanism (not shown) which translates positive values of p to plantarflexor stimulation and negative values ($p < 0$) to dorsiflexor stimulation.

designed using an empirically-determined linear dynamic model of the muscles' response from p to m i.e. a series of stimulation test signals of PRBS form is applied in open loop and a family of linear models is estimated. This family includes models of both the dorsiflexor and plantarflexor responses, but only one model is chosen and used to design a single linear controller for ankle moment. The analytical methods used for model identification and controller design are described in detail elsewhere [2].

3 Results

After tests for setting the current, and after model identification and controller design, the following *control* tests are carried out:

Test AS: Ankle **S**tiffness control. Here the body is fixed (ropes taut), but the feet are wobbled in order to achieve a change in ankle angle by manual adjustment of the wobbling speed. This is a test of ankle stiffness control, achieved via ankle moment control, i.e. the reference moment $m_{\text{ref}} = k^s_{\text{ref}}\theta$ is determined as in figure 1 by the product of the current ankle angle θ (measured by the shaft potentiometer) and the *desired stiffness* k^s_{ref}.

Results from the ankle stiffness control test with a desired stiffness value $k^s_{\text{ref}} = 4$Nm/deg, are shown in figure 2. The top graph shows the ankle angle θ as measured at the shaft potentiometer. The second graph shows the reference moment $m_{\text{ref}} = k^s_{\text{ref}}\theta$ and the controlled moment m.

It is seen that the controller switches stimulation between the plantarflexors and dorsiflexors to achieve good moment tracking. The bottom graph shows the approximate value of stiffness obtained during the test by computing $k^s(t) = \frac{m(t)}{\theta(t)}$. Note that this computation is sensitive to the fact that θ becomes very close to zero each time the ankles cross the neutral position (where $\theta = 0$), and also to the fact that θ and m change sign at slightly different times. This explains the peaking in the plot of $k^s(t)$. It can nevertheless be seen that $k^s(t)$ is centred almost exactly on the desired value of 4Nm/deg.

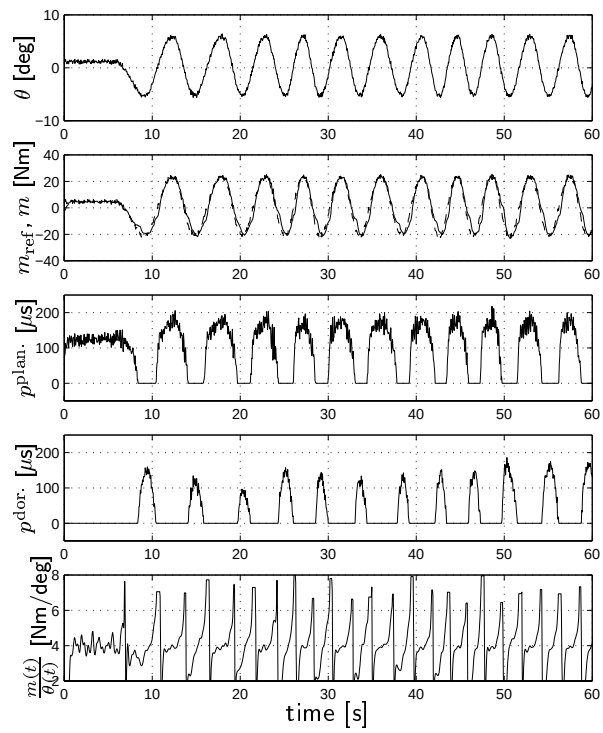

Figure 2: Test AS: ankle stiffness control with $k^s_{\text{ref}} = 4$ Nm/deg. The dashed line in the second graph from top is the reference moment; solid lines are measured quantities.

A second view of the accuracy of stiffness control during this test is given in figure 3 which is a scatter plot of m against θ. The solid line in this plot represents a "pure" stiffness of 4Nm/deg.

Test SS: Stiff **S**tanding. In this test the feet are fixed (no wobbling) but the body is allowed to sway (slack ropes). This is again a test of stiffness control via ankle moment control, with $m_{\text{ref}} = k^s_{\text{ref}}\theta$, where the current body angle θ is measured by the potentiometer behind the subject.

After selecting the desired stiffness the experimenter stands in front of the subject holding on to the body brace and moves the body manually through a range of angles. The required external moment m_e can be estimated by neglecting dynamic effects with the static moment balance $m_e = m - \tilde{m}gl\theta$. (Here, $\tilde{m}$ is the body mass, g is gravitational acceleration, and l is the distance from the ankle joint to the body's centre of mass. $\tilde{m}$ is readily measured and l is estimated.)

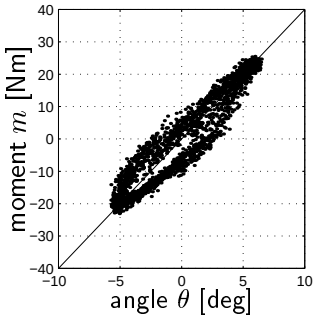

Figure 3: Test AS: ankle stiffness control with $k^s_{\text{ref}} = 4\,\text{Nm/deg}$ (same data as in figure 2). This is a scatter plot of moment against angle. The solid line represents an exact stiffness of 4Nm/deg.

Results from one stiff standing test are shown in figure 4. The corresponding scatter plot of m against θ is shown in figure 5. Thus the angle profile shown in the top graph is achieved by the experimenter slowly pulling and pushing with an external moment m_e whose estimate is plotted in the bottom graph of figure 4. The desired stiffness in this test was $k^s_{\text{ref}} = 20\text{Nm/deg}$.

It can be seen that for forward-leaning postures ($\theta > 0$, when 0s$< t <$12s and $t >$22s) good moment tracking is achieved (second graph in the figure) via plantarflexor stimulation, and that $k^s(t) = \frac{m(t)}{\theta(t)}$ is close to the desired 20Nm/deg.

When the subject is pushed backwards, however, the relatively weak dorsiflexor muscles are not able to generate the moments required: in the time periods 13s$< t <$15s and 17s$< t <$21s there is a significant difference between the desired and actual moments because the dorsiflexor stimulation has reached its maximum value (400μs). During these periods the desired stiffness level of 20Nm/deg cannot be achieved. However, it is interesting to note that with constant, maximal, dorsiflexor stimulation the stiffness achieved settles on an approximately constant level of around 5Nm/deg. These two stiffness regimes are clearly reflected in the scatter plot of figure 5.

Test EPC: External **P**osture **C**ontrol. This test is carried out under similar conditions to Test SS. In this case the experimenter provides *external control* (moment m_e) by standing in front of the subject and holding onto the body brace at a measured height from the ankles. A desired reference angle θ_{ref} is generated and displayed on a screen to the experimenter in realtime together with the measured inclination angle. The task of the experimenter is to make the posture follow the reference angle as accurately as possible.

The main point here is to assess the ease or difficulty with which external **tracking control** can be achieved. This can be assessed for various desired stiffness values k^s by comparing the required external forces and the tracking errors. As in Test SS the external moment can be estimated here from $m_e \approx m - \tilde{m}gl\theta$.

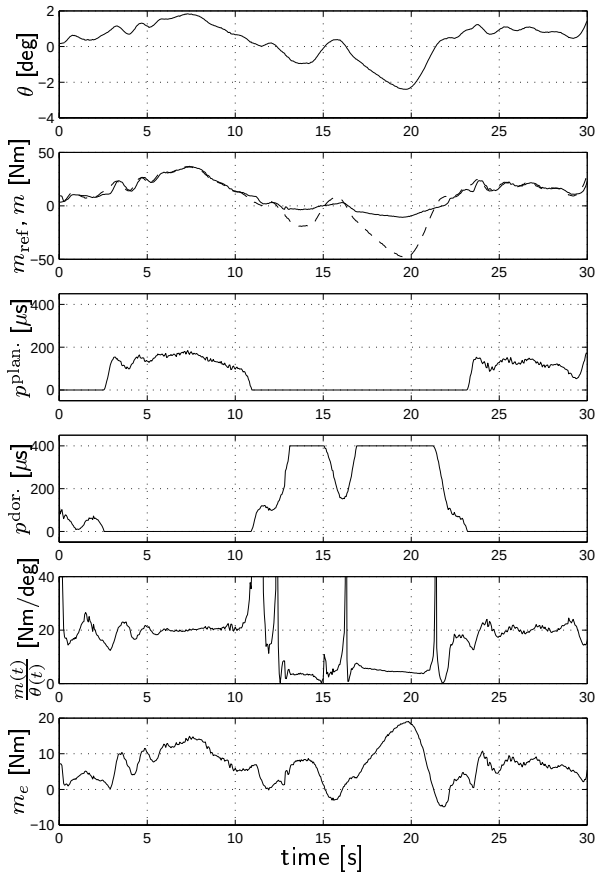

Figure 4: Test SS: stiff standing control with $k^s_{\text{ref}} = 20\,\text{Nm/deg}$. The dashed line in the second graph from top is the reference moment; solid lines are measured quantities.

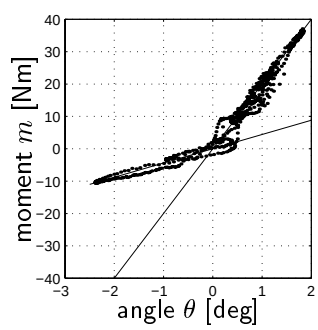

Figure 5: Test SS: stiff standing control with $k^s_{\text{ref}} = 20\,\text{Nm/deg}$ (same data as in figure 4). This is a scatter plot of moment against angle. The solid lines represent exact stiffnesses of 5 and 20 Nm/deg.

Two tests of external posture control were carried out for two different values of desired stiffness: 10Nm and 30Nm. When $k^s_{\text{ref}} = 10\text{Nm/deg}$ the experimenter is able to achieve reasonable angle tracking with only modest hand moments m_e of magnitude up

to 20Nm (data not shown). Qualitatively, the experimenter reported that the task was easy to perform at this level of desired stiffness.

When $k^s_{\text{ref}} = 30\text{Nm/deg}$ (figure 6) the increased stiffness generates much higher levels of demanded ankle moment. Good moment tracking is achieved during forward lean, with stiffness approximately equal to the desired value of 30Nm/deg. In backward lean the dorsiflexors again quickly saturate giving large moment tracking errors and a stiffness value around only 5Nm/deg.

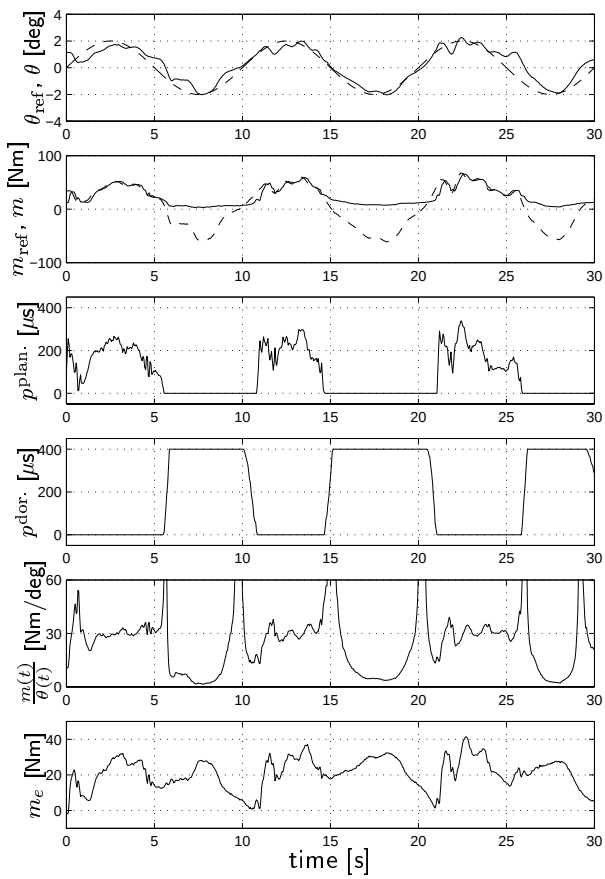

Figure 6: Test EPC: external posture control with $k^s_{\text{ref}} = 30\,\text{Nm/deg}$. The dashed lines in the two top graphs are reference values; solid lines are measured quantities.

The experimenter described this angle tracking task as very difficult, and this is presumably due to the large moments required from the experimenter (up to 40Nm - see the bottom graph in figure 6).

4 Discussion and Conclusions

The results have shown that ankle stiffness control can be achieved with FES of the ankle flexor and extensor muscles. The ability to achieve specified ankle stiffness is fundamentally limited by actuator force (i.e. the controlled muscles). The results have shown in particular that stiffness control in backward-leaning postures is limited by the relatively weak dorsiflexor muscles. Accuracy of stiffness control is also dependent upon the bandwidth of the ankle moment control loop. This is because the stiffness control structure (figure 1) attempts to achieve a desired stiffness k^s_{ref} by setting the *reference* for the ankle moment control loop to $m_{\text{ref}} = k^s_{\text{ref}}\theta$. However, the stiffness level achieved in closed loop is determined by the actual moment m which is $m = G_m k^s_{\text{ref}}\theta$, where $G_m = m/m_{\text{ref}}$ is the closed-loop transfer function of the moment loop. The quantity $k^s(t) = m(t)/\theta(t)$ has been computed in the above experiments. With this definition we note that $k^s = G_m k^s_{\text{ref}}$ which shows that closed-loop stiffness is governed by the ankle moment control loop G_m. By implication the accuracy of stiffness control can be improved by increasing this bandwidth.

The results of further stiff standing tests have shown, as expected, that ankle stiffness alone is not sufficient to stabilise the body. Stabilisation of an ideal inverted pendulum requires a stiffness satisfying $k^s > \tilde{m}gl$. A subject with mass $\tilde{m} = 70$kg and centre of mass at length $l = 1$m from the ankle joint therefore requires a stiffness of at least 12Nm/deg. Note, however, that $k^s > \tilde{m}gl$ is a necessary, but not sufficient, condition. It was observed in stiff standing tests that a desired stiffness of 20Nm/deg resulted in ankle moments which tended to push the subject back towards the neutral position, but did not stabilise the body.

Of central importance in this study was not the property of absolute stability, but rather the degree to which ankle stiffness can facilitate external control of the body. The external posture control tests showed that higher levels of stiffness make it more difficult to perturb the body away from the neutral position. We postulate therefore that it will be easier to stabilise around the neutral position with externally applied damping or with voluntary upper-body motor control inputs when the controlled stiffness is significantly greater than the critical value $\tilde{m}gl$. It remains to experimentally verify this hypothesis.

References

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