Feedback Control of Unsupported Standing in Paraplegia—Part II: Experimental Results

Marko Munih, *Member, IEEE*, Nick de N. Donaldson, Kenneth J. Hunt, and Fiona M. D. Barr

***Abstract*—This is the second of a pair of papers which describe an investigation into the feasibility of providing artificial balance to paraplegics using electrical stimulation of the paralyzed muscles. By bracing the body above the shanks, only stimulation of the plantar flexors is necessary. This arrangement prevents any influence from the intact neuromuscular system above the spinal cord lesion. In this paper, we present experimental results from intact and paraplegic subjects.**

***Index Terms*—Artificial balance, feedback control, optical control, paraplegia, unsupported standing.**

I. Introduction

In Part I of this two-part paper [1], we introduced this work in which feedback controllers are used to try to stabilize the inverted pendulum of a paraplegic's body by stimulation of the ankle plantarflexors. We described the control structure which, by using three nested feedback loops, should be robust; the use of linear quadratic Gaussian (LQG) control which allows these loops to be tuned, each with two parameters ("tuning knobs"); and the methods of measuring the properties of the stimulated muscle and the biomechanical properties of the body. We continue in this paper by describing the experimental methods and some of the results from a neurologically intact and a paraplegic subject. These results show that the LQG controller can conveniently be tuned and then gives satisfactory results with the intact subject. The results from the paraplegic are most interesting because they show the limitations of even a good control strategy: the performance is less satisfactory due to the rapid muscle fatigue and spasticity.

II. Experimental Procedure and Apparatus

A. Test Procedures

The sequence of tests which can be carried out during a laboratory session begins with *system identification*. As described in [1] of this paper, and in [2] and [3], the impulse response test allows determination of the muscle's recruitment nonlinearity. Following this, a pseudo-random binary sequence (PRBS) test is used to determine the dynamic part of the muscle response. Note that the muscle parameters are estimated under isometric conditions. The mass and moment of inertia of the body are determined using the method described in [1]. Having obtained the muscle and body measurements, LQG controller parameters can be calculated quickly.

We distinguish between two types of test for the control of standing: "Imitation Standing" and "Actual Standing." Actual Standing consists in fixing the feet and allowing the body to move in the sagittal plane. The nested control structure employed is shown in [1, Figs. 2–3]. In Actual Standing all controllers are active: two moment controllers regulate the moments produced in the left and right ankles, and the angle controller aims to stabilize the body at some desired reference angle.

As mentioned above, the muscle parameters for both ankles are determined under isometric conditions; the ankles are fixed during the impulse response and PRBS tests. During Actual Standing, the muscles will not operate isometrically due to sway and to changes made to the reference angle. The "Imitation Standing" procedure was devised as a prelude to Actual Standing to test whether the moment controllers could produce the desired moments during simulated standing conditions when the muscles are not operating isometrically. If this is so, we can reasonably expect that the moment controllers will function properly during Actual Standing. In Imitation Standing the body is fixed upright and the ankles are wobbled sinusoidally at various amplitudes and frequencies. Imitation Standing is depicted schematically in Fig. 1 (cf., [1, Fig. 3]). The body angle is held fixed and the ankle angle induced by wobble (θ_{wobble}) is injected into the feedback path of the outer loop.

Operation under nonisometric conditions will lead to some mismatch between the muscle model and the actual muscle dynamics, but the feedback nature of the moment controllers gives a certain degree of robustness against this type of uncertainty. Following control design, the stability margins (gain and phase margins) are checked to ensure a sufficient degree of robustness and then the controllers are tested by Imitation Standing.

Of all the possible tests [2]–[4], only Actual Standing tests are presented in this paper.

B. The Wobbler Hardware

The Wobbler apparatus has been built to explore, in the laboratory, the practical feasibility of using functional electrical stimulation (FES) of the ankle plantarflexors to achieve

Manuscript received February 24, 1997; revised September 14, 1997. This work was supported by the British Medical Research Council under Grant G9220653.

M. Munih is with the Faculty of Electrical Engineering, University of Ljubljana, Ljubljana, Slovenia.

N. de N. Donaldson is with the Department of Medical Physics and Bioengineering, University College London, London WC1E 6JA U.K.

K. J. Hunt is with Daimler-Benz AG, Alt-Moabit 96 A, Berlin, Germany.

F. M. D. Barr is with the Royal National Orthopaedic Hospital Trust, Stanmore, U.K.

Publisher Item Identifier S 1063-6528(97)08932-5.

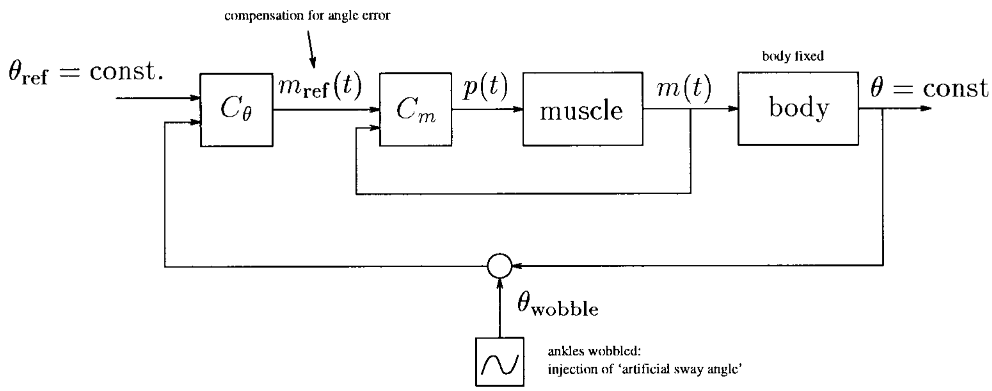

Fig. 1. Imitation Standing: ankles are wobbled while body is fixed. This arrangement tests whether the moment controllers can produce the required moments during simulated standing conditions, i.e., in nonisometric muscle conditions and without Actual Standing verifies if position controller responds reasonably. θ_{wobble}: ankle angle due to wobbling, m: measured ankle moment, p: muscle stimulation pulse width, θ_{ref}: angle set-point held constant, m_{ref}: moment set-point, C_θ: angle controller C_m: moment controller.

unsupported standing of completely paralyzed persons while the other joints are braced. The details of the hardware and software are presented in [4]. The most important components are two open-top boxes into which the shoes are fixed, and these are mounted on a shaft, which may or may not be fixed, with independent torque measurements for the left and right ankles [4]. Also mounted on the shaft are a precision encoder, safety rotation stops, an electromagnetic clutch, and a 150 Nm torque limiter. The shaft is driven by a crank-rocker mechanism from a flywheel which is itself propelled through a speed-reducing toothed-belt drive from a dc motor. The motor speed is adjusted by setting the armature voltage.

The device allows different tests in a normal upright standing posture: ankle muscle identification using various methods [3], ankle stiffness measurements [5], ankle moment control [2], and closed-loop position control.

In the standing tests the subject is strapped into a brace made of plastic shells, reinforced and joined by steel strips. The feet are normally tied in sport shoes which are aligned horizontally and vertically so that ankle plantarflexion centres lie on the shaft's axis. The shoes are glued on aluminum plates which are bolted into the foot boxes. In Imitation Standing four light ropes are used to fix the body brace's left and right shoulders forward and backward in the sagittal plane. This prevents any swinging of the upper body. With such fixation, no segments above the ankle joints can move, while the feet remain free to rock. These elements together represent, in effect, a fixed single inverted pendulum with a support which can be rocked. The arrangement described above has been used for measuring moment and position controller frequency-response characteristics at various wobbling frequencies.

For the Actual Standing experiments, the Wobbler boxes are boxed in a horizontal position while the ropes are slightly released to allow body pendulum fore-and-aft sway. As the subject may fall, we only slacken the ropes attached to the subjects' shoulders so that they cannot fall far.

The left and right feet moments are measured with two torque load cells: one measures the right moment only and the second measures both ankle moments together. The left ankle moment signal is then realized by subtraction in an operational amplifier. For the position controller and for Imitation Standing the ankle angle is measured with a precision shaft encoder mounted on the shaft. It has a resolution of 0.018 degrees. For measurement of the inclination angle in Actual Standing, a high-resolution 2% linearity potentiometer, mounted approximately 1.6 m behind the person at shoulder height is used. A thread passes from the body brace round a pulley fixed to the potentiometer and is held taut by a hanging 60 g weight. The resolution can be set by selecting one of two pulley diameters. The resolution is normally 0.014°, with worst case peak-to-peak noise of 0.03° for a maximum excursion angle of 14.5°.

C. Wobbler Software: Experimental Programs

The Wobbler hardware is supported by specially-written real-time programs, display programs, data conversion programs, and MATLAB scripts and functions [4]. In the real-time program the position controller runs at $f_\theta = 6.7$ Hz. This signal is input to left and right moment controllers, which sample at $f_m = 20$ Hz. Outputs of the two moment controllers are passed separately to the serial link program handler, and from there to the stimulator (for each pulse separately).

D. Experimental Subjects

The intact subject in the measurements shown here was 43 years old, had height 170 cm, mass $\tilde{m} = 70$ kg and inertia $J = 88.7$ kgm^2.

The paraplegic subject has a complete T5 lesion, was 35 years of age, 13 years after injury, had height 175 cm, mass $\tilde{m} = 75$ kg, and inertia $J = 95$ kgm^2. He undertook an isometric exercise programme for the ankle plantarflexors and dorsiflexors using bilateral reciprocal stimulation of the gastrocnemius and tibialis anterior muscles. A plantargrade position was maintained at the ankle during exercise, utilising bespoke plaster of paris splints. Exercise was performed while sitting for 30 min daily. He exibited severe spasticity (Ashworth Scale, grade 4), presenting with flexor and extensor spasms in the lower limbs which were not reduced by passive movements and stretching prior to the experiments.

During the experiments both subjects held their arms crossed on the chest, and the intact person also had closed eyes. The paraplegic's spasms were not present during quiet standing in the Wobbler but were elicited by postural changes and higher levels of stimulation.

E. Preliminary Imitation Standing Tests

Results from the intact subject, with f_m and f_θ at 20 Hz, demonstrate that the position controller, as well as both moment controllers, respond as expected. However, with the paraplegic subject, spiky signals at the moment controller input led to instability in the activations. Due to the finite torsional stiffness of the foot boxes, attributable mainly to the the compliance of one torque load cell, unsteady activation caused small but noticeable angle disturbance, which in propagating through the position controller, amplified the oscillatory frequency due to the wide controller bandwidth. We then changed f_θ to 6.7 Hz, which gave much better results in further Imitation Standing tests, so we proceeded to Actual Standing tests with this position loop sampling rate.

III. Actual Standing Results: Intact Subject

The aims of the Actual Standing tests were as follows:

1) to investigate the sensitivity to the "control knob" settings and find satisfactory settings with intact subjects;
2) to use these settings for intact standing trials at fixed and varying reference angles (tracking test), and for various disturbances (disturbance test);
3) finally, to implement and evaluate paraplegic standing.

The disturbances were introduced during constant reference angle tests in order to test closed-loop balance capabilities under realistic conditions.

With the intact person as the test subject, the effect of the position controller tuning knobs ρ_θ and t^θ_{obs} was studied in the same experimental session as the balance disturbance tests. The inner loops, adjusted beforehand as described in [2],[1] were taken account of in the design of position controllers having various ρ_θ and t^θ_{obs} values. In the first experiments a deadbeat observer $(t^\theta_{\mathrm{obs}} = 0)$ was compared to an observer value of $t^\theta_{\mathrm{obs}} = 0.4$, while ρ_θ was held constant. In simulation, $t^\theta_{\mathrm{obs}} = 0.4$ gave a satisfactory response, but as one expects from consideration of the sensitivity and complementary sensitivity functions [2], the controllers with t^θ_{obs} greater than zero, while maintaining good reference tracking, introduced increased phase lag in the loop and thus degraded the disturbance rejection performance. This was confirmed experimentally for the position controllers, where regulation to a constant angle setpoint was degraded when the observer rise-time was increased.

All graphs of experimental results have the same structure, and are arranged as follows: the top graph shows three moments: measured left and right ankle moments and also the required moment m_{ref}. The latter is the output of the position controller and determines an equal reference moment input for the left and right moment controllers. The left moment is presented as a dashed line, the right moment as a solid line, while the reference moment is plotted as a dotted line with large dots. The middle graph of all figures shows the input and output of the position controller; the pendulum (body) reference angle is dotted and the measured inclination angle is a solid line. The outputs from both moment controllers are input to the stimulator as left and right muscle activations, shown in the bottom graph of the figures. Again, the left side is presented as a dashed line and the right side as a solid line.

[1] Initially the stimulation current for each channel was set so that, at maximal pulse width (500 μs), it was just below the level at which either the stimulation or the resulting tightness at the ankle became painful.

A. Tracking Tests

To show the effect of varying ρ_θ with a fixed deadbeat observer, a number of controllers were tested in experiments with varying reference angle signals:

1) **High** $\rho_\theta = 0.1$ **value:** The moment controllers were set to good values [2] of $\rho_m = 0.000\,05$ with a deadbeat observer. The response of this very "lazy" position controller is shown in Fig. 2 for the first 60 s of a 3 min long trial. Curves for the remainder of the experiment were tested in experiments with varying reference angle signals After an initial transient lasting 10 to 15 s, during which the position controller "locks in" (see Discussion), the pendulum inclination angle roughly agrees with the average input angle with a just-distinguishable tendency to track up and down. In the moment graph the left and right moment controllers track the desired moment (the position controller output) closely. The noise, evident in the activation signals, is an unavoidable consequence of the high bandwidth of the moment controllers used here. Notice that although the position controller bandwidth is not high enough to track the desired position waveform, the average required pendulum angle is maintained and the subject is balanced.
2) **Medium** $\rho_\theta = 0.001$ **value:** The moment controllers were in this case set to $\rho_m = 0.000\,01$ with a deadbeat observer. This much better reference angle tracking is shown in Fig. 3. Again 1.5° step changes in the required angle, from −1.5 to −3° were applied. The response is much faster, with transients of more than 5 but less than 10 s. Compared to the results in Fig. 2, the calculated reference moment is slightly more noisy, as expected through the reduction in ρ_θ, and this is also seen in the activation graph. However the most important effect noticed in this figure is the faster position loop response in tracking the desired inclination angle.
3) **Low** $\rho_\theta = 0.0001$ **value:** The measurement results with moment controllers set to $\rho_m = 0.000\,01$ with deadbeat observers are shown in Fig. 4. The test conditions and graph presentation are identical to the measurements presented above. The closed-loop position control rise times are now smaller, lying in the range around 3 s for a 1.5° position change, which is approximately three times faster than with $\rho_\theta = 0.001$. The reference moment, which is the output from the position controller, is slightly noisier due to the lower ρ_θ value. Both activations are similar to those in Fig. 3 with better angle

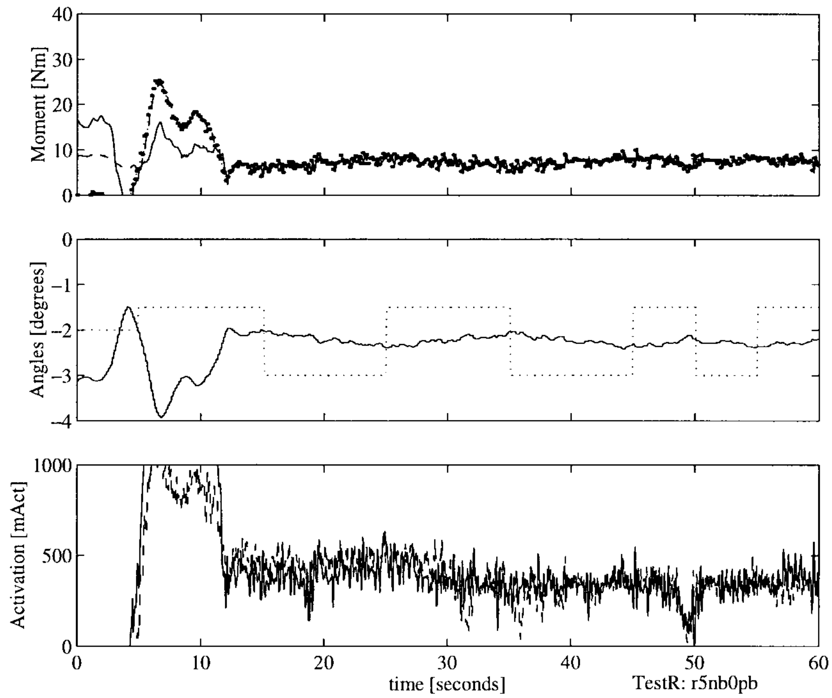

Fig. 2. Actual Standing of intact subject with slow (“lazy”) position controller and ankle angle reference signal varying between 1.5 and 3°. Controller parameters: $\rho_m = 0.000\,05$ and a deadbeat observer, $\rho_\theta = 0.1$ and a deadbeat observer.

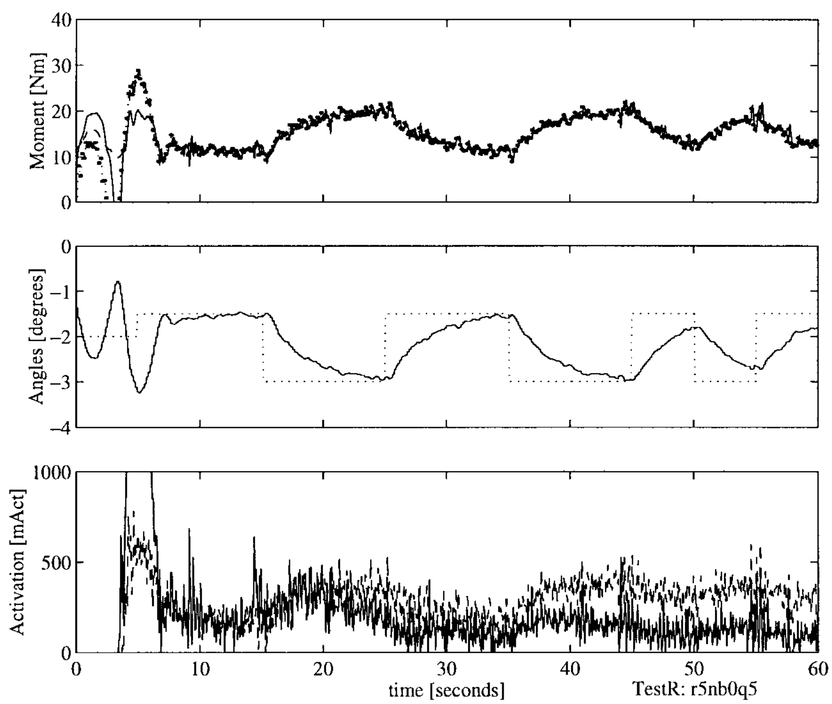

Fig. 3. Actual Standing of intact subject with medium-speed position controller setting and ankle angle reference signal varying between 1.5 and 3°. Controller parameters: $\rho_m = 0.000\,01$ and a deadbeat observer, $\rho_\theta = 0.001$ and a deadbeat observer.

tracking. We tested even smaller ρ_θ values but these did not give significantly faster responses. Note that with this setting, sometimes when fatigued, the control appeared to be less stable (see paraplegic results). This fatigue-induced instability significantly affected function during numerous experiments with paraplegic subjects, where fatigue is much more pronounced and may be accompanied by spasticity.

B. Disturbance Tests

A number of further tests are shown here in the intact subject with a constant reference angle in order to check the disturbance rejection capabilities of the controllers. The subject was disturbed in three ways: being pushed forward from behind (Fig. 5); moving his arms while holding weights (Fig. 6); and being pulled forward by a rope (Fig. 7). Initially,

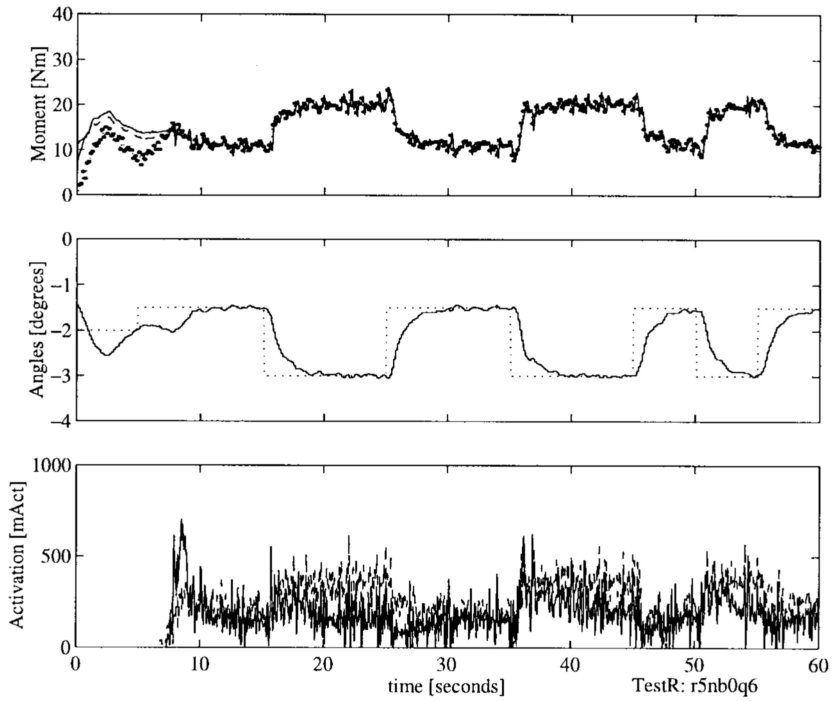

Fig. 4. Actual Standing of intact subject with fast position controller and ankle angle reference signal varying between 1.5 and 3°. Controller parameters: $\rho_m = 0.000\,01$ and a deadbeat observer, $\rho_\theta = 0.0001$ and a deadbeat observer.

various reference angle set-points in the range $-4 \leq \theta_{\rm ref} \leq -1$ degrees were investigated. The most comfortable settings for intact persons, and the easiest to regulate (lock-in), lie in the range 2 to 3.5°. This is similar to the inclination angle during normal upright standing with the Ground Reaction Force Vector being 3.5–6 cm in front of the ankles [6].

A standing trial with constant position reference and the position controller operating at a 6.7 Hz sampling rate is presented in Fig. 5. The two quiet standing periods, first from 5 to 23 s and then from 40 to 60 s, demonstrate how good the position regulation can be. The angle variations are within ±0.1°, though even narrower fluctuations were achieved during other laboratory tests (being dependent mostly on the level of angle measurement noise). Such accuracy is usually not possible during voluntary standing [6], proving that the results for artificial control of intact persons are not due to normal motor control, voluntary or involuntary. Moreover, precise control to the exact reference angle cannot be achieved voluntarily without numerical feedback of the actual angle (the subject's eyes were closed in the experiments).

The middle time period in this figure shows three similar disturbance events, with the subject being pushed from behind. The top graph shows the disturbance moment resulting from pushing the subject from behind; it is approximately 40 Nm peak on all three occasions, with the transient lasting about 3 s. As expected, similar disturbances in the moments cause similar disturbances in the angle response.

The second disturbance test was carried out with the subject holding a 5 kg weight in each hand. One or both weights were raised from hanging at arms' length to shoulder height with the arm in front of the trunk. The first lift (Fig. 6) by the left hand only started at 18 s and lasted for 10 s. During this asymmetrical lift, especially when the weight was quickly lowered, the required moments (dots), left actual moment (dashed line) and right actual moment (solid line) differ markedly. The second weight lift using both hands together started at 35 s and lasted 17 s. After the 5 s initial transient response, the ankle angle stabilized at the reference value despite the changed pendulum weight and moment of inertia.

In a final disturbance test (Fig. 7), the subject was pulled forward with a rope attached at belt height (approximately 1m above the ankles). The force of approximately 30 N was measured with a spring balance. During quiet standing, the rope was tugged three times starting at 27, 32 and 51 s. The second tug lasted 9 s. The top graph shows that during the first tug, the peak moment was approximately 40 Nm, while the second and third tugs resulted in greater peak moments. The vertical scales for moment and angle in this figure are kept the same as in the previous figures for ease of comparison. In the second tug, the muscle activation reached maximal level, resulting in left and right plantarflexor moments in excess of 40 Nm for 2 s. After that, by time 40 s, the pendulum angle returned to the reference value (2°) despite the superimposed load.

IV. Actual Standing Results: Paraplegic Subject

Finally, the nested-loop LQG controllers were tested with the paraplegic standing in the Wobbler. We warn readers that the paraplegic controller responses for the same settings of ρ_m, ρ_θ and observers should not be directly compared to the results for the intact subject because of the differences in their muscle transfer functions; the closed-loop control properties for an LQG design depend not only on these design parameters, but also on the muscle parameters.

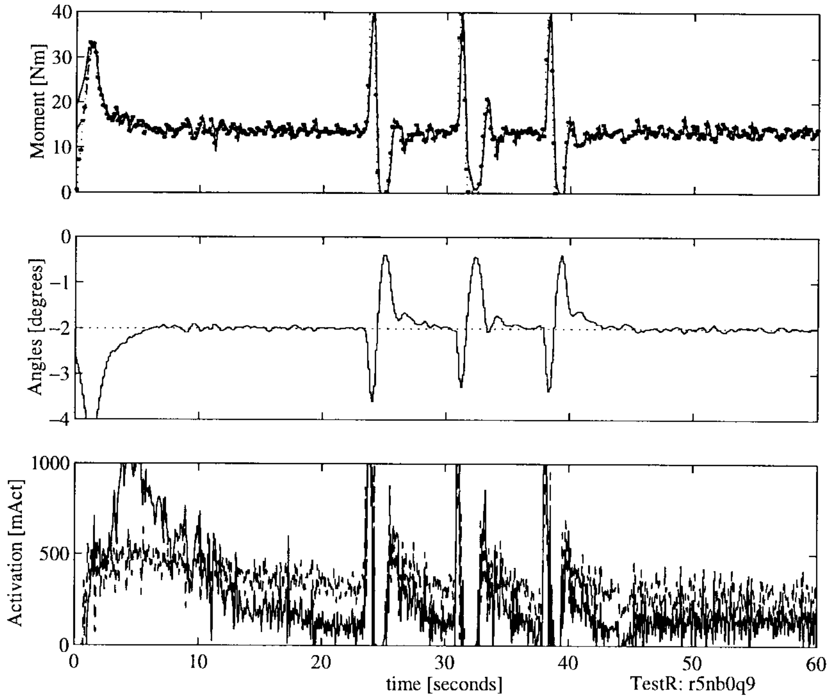

Fig. 5. Actual Standing of intact subject: quiet standing and three disturbances from the person's back. Ankle angle reference signal is fixed at 2°. Controller parameters: $\rho_m = 0.00001$ and a deadbeat observer, $\rho_\theta = 0.0001$ and a deadbeat observer.

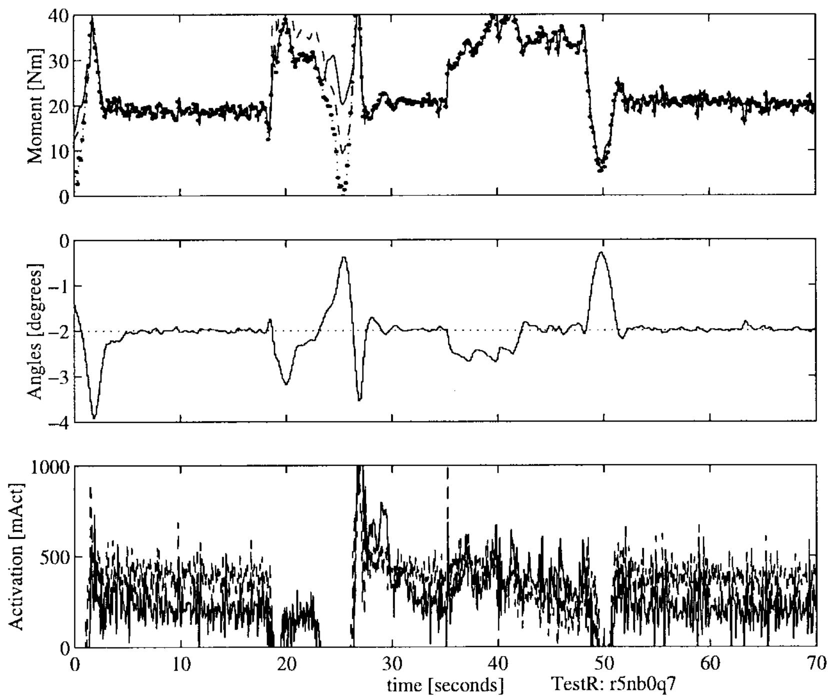

Fig. 6. Actual Standing of intact subject: quiet standing and twice extending hands forward with 5 kg weights in each hand (see text). Ankle angle reference signal is fixed at 2°. Controller parameters: $\rho_m = 0.00001$ and a deadbeat observer, $\rho_\theta = 0.0001$ and a deadbeat observer.

A number of successful, short, standing periods were achieved: see Figs. 8–11. The experiments lasted up to 60 s, but unlocked periods before and after stability are cut out and are not shown here. During "unlocked" periods the paraplegic subject was supported by an experimenter or by the ropes. In addition to the control of standing, each figure also demonstrates some phenomena encountered during that test. It is also characteristic of all the paraplegic trials that the body inclination angle is not as close to the reference value as we saw with the intact subject.

The test shown in Fig. 8 is characteristic because 1) left activation is on average much higher than the right showing left/right asymmetry and 2) the moment and activation signals are only marginally stable, especially on the right side. The

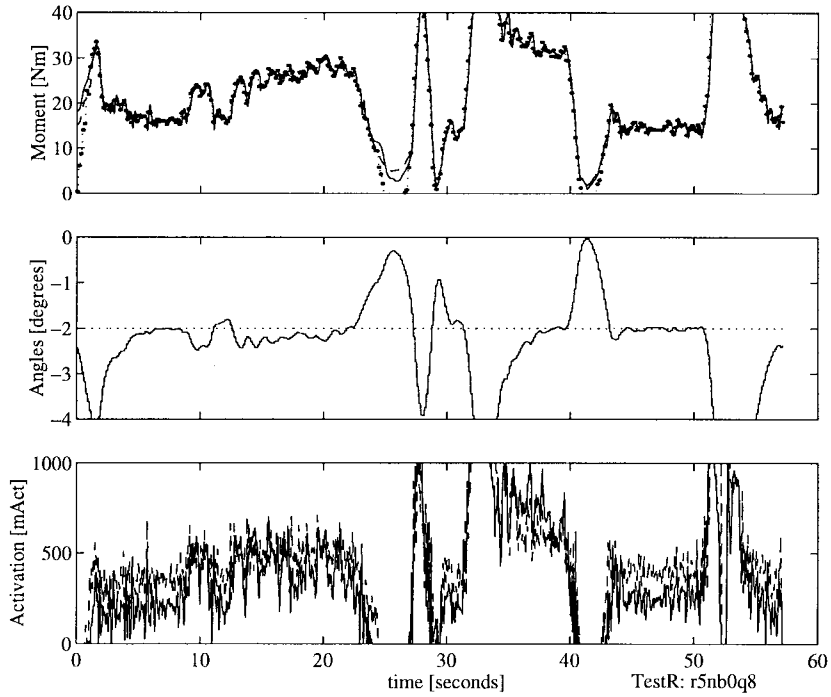

Fig. 7. Actual Standing of intact subject: quiet standing and three times being pulled forward with a rope attached at the waist 1m above the ankles. Force in the rope was approximately 30 N and ankle angle reference signal is fixed at 2 degrees.Controller parameters: $\rho_m = 0.00001$ and a deadbeat observer, $\rho_\theta = 0.0001$ and a deadbeat observer.

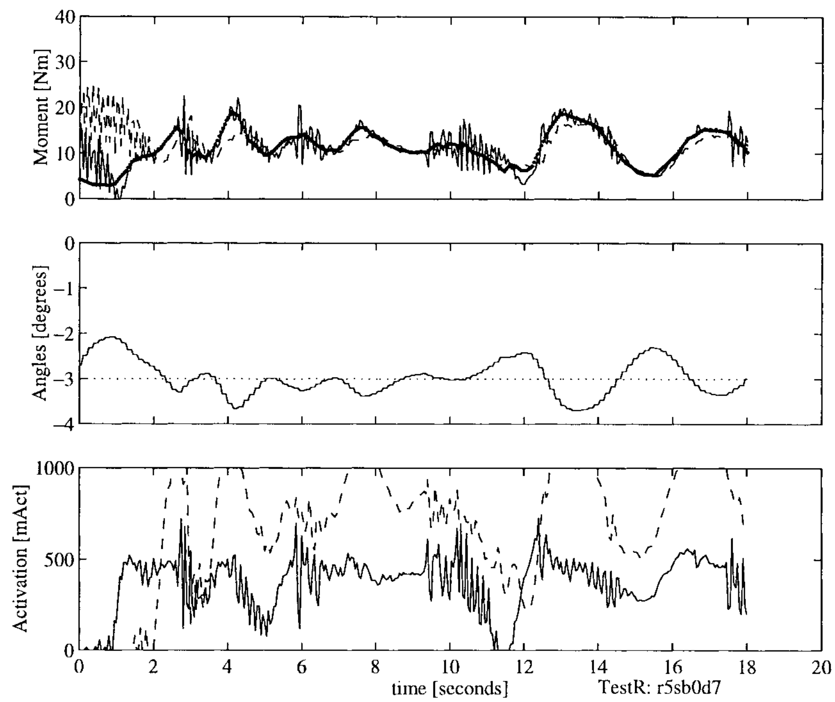

Fig. 8. Actual Standing of paraplegic subject: demonstrating left/right asymmetry. The ankle angle reference signal is fixed at 3°. Controller parameters: $\rho_m = 0.02$ and $t_{\rm obs}^m = 0.2$, $\rho_\theta = 0.5$ and $t_{\rm obs}^\theta = 0.4$.

asymmetry is more pronounced than we ever encountered in intact persons, suggesting that this is probably not merely due to poor electrode placement. More likely, the left and right muscles are unequally affected by fatigue.

The angle of the body is accurately maintained for the first half of the test shown in Fig. 9. However, after 12 s, the muscle activations saturate and the angle error increases. It is interesting that despite saturation, which must prevent feedback action, the body does not fall over, at least for the following 12 s, and then it falls over backward! The reference inclination angle is 3.5° and ankle moments are both more than 20 Nm, which causes rapid fatigue for the paraplegic during a long trial. The left/right asymmetry seen in the previous figure is noticeable here too.

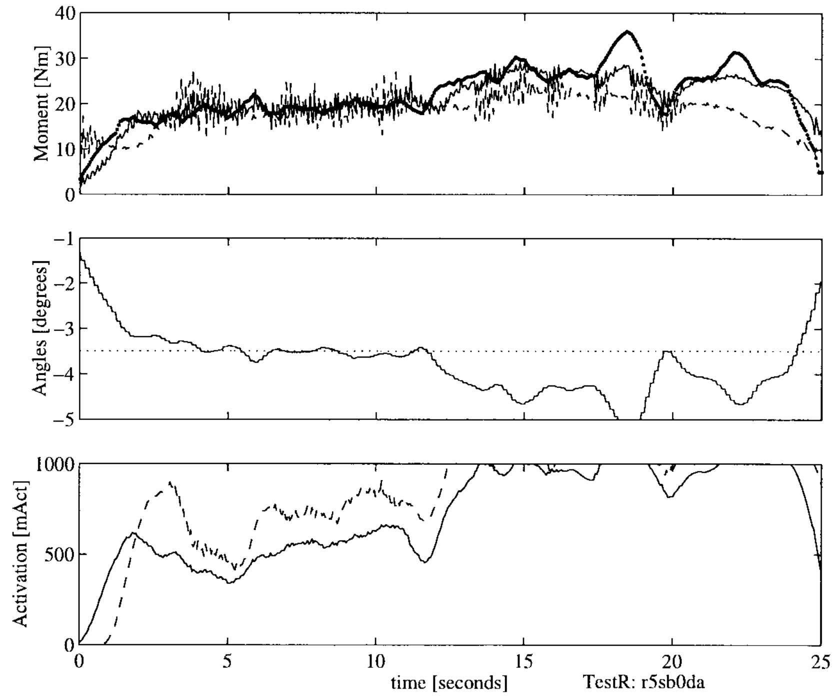

Fig. 9. Actual Standing of paraplegic subject: good position stability during first half of the trial and saturated activation later (fatigue). The ankle angle reference signal is fixed at 3.5°. Controller parameters: $\rho_m = 0.02$ and $t^m_{\text{obs}} = 0.2$, $\rho_\theta = 0.2$ and $t^\theta_{\text{obs}} = 0.4$.

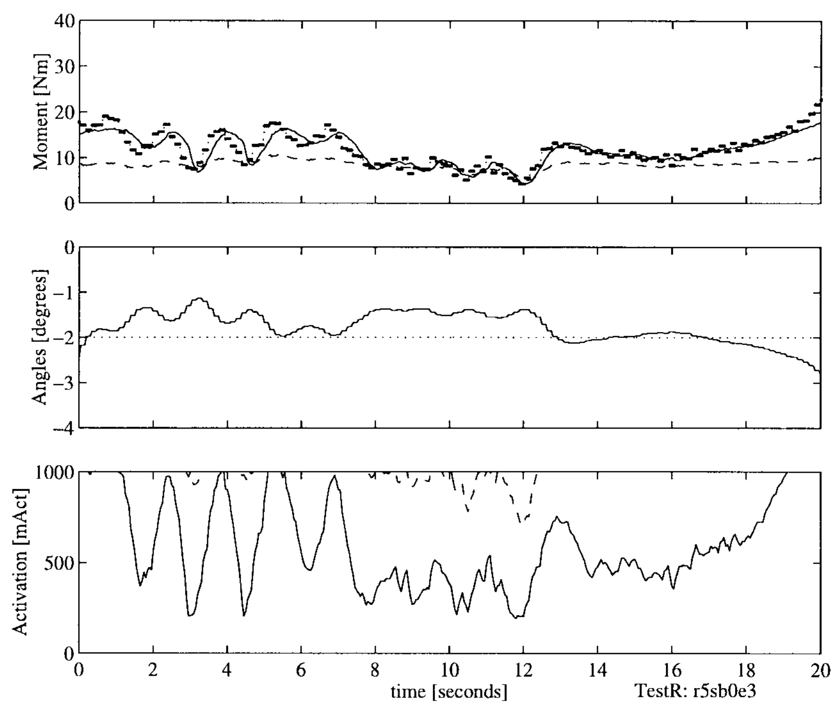

Fig. 10. Actual Standing of paraplegic subject: presentation of left/right asymmetry with left activation being saturated. Nevertheless, the subject does not fall until the end of the test. Ankle angle reference signal is fixed at 2°.Controller parameters: $\rho_m = 0.0005$ and a deadbeat observer, $\rho_\theta = 0.0001$ and a deadbeat observer.

In Figs. 10 and 11, the effect of fatigue is dominant. This means that the transfer functions, determined during identification measurements at the beginning of the session, would have changed significantly. The effect of fatigued muscles in both legs can be seen. Activations are very high while the generated moment is only about 10 Nm. It is interesting to note that the angle regulation is comparable to the other trials despite the extreme fatigue.

V. Discussion of Results

A. Results with Intact Subjects

Is it valid to test an artificial balance controller on intact subjects? When the intact subject stands in the Wobbler with his eyes shut, despite the body brace, he is easily able to maintain balance, presumably using his vestibular system,

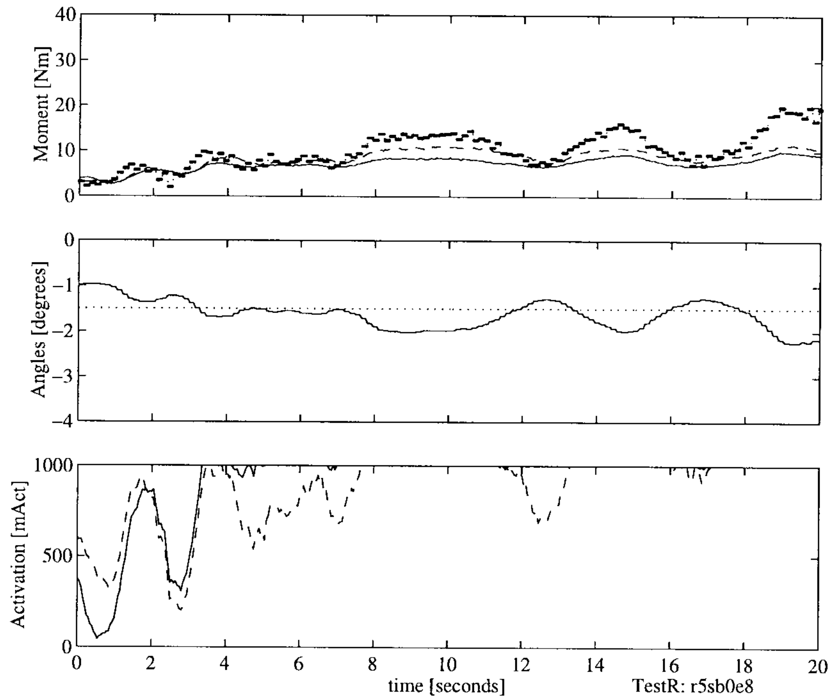

Fig. 11. Actual Standing of paraplegic subject: shows significant fatigue on both sides. Ankle angle reference signal is fixed at 1.5°. Controller parameters: $\rho_m = 0.00005$ and a deadbeat observer, $\rho_\theta = 0.0001$ and a deadbeat observer.

proprioception from his ankles and exteroception from his feet. He feels confident that he can do this while the foot boxes are fixed, and endeavours to relax before the stimulation of a "Actual Standing" test begins. If the stimulation is painful, which occasionally happens, or the subject is not relaxed, the onset of stimulation may cause involuntary contraction of the dorsiflexors (flexor withdrawal reflex?). In that case, trying to obtain the required level of plantarflexion moment, the controller increases activation of the calf muscles: a positive feedback which "latches up" at maximal pain and maximal activation. However this was rare (and no instances are shown in the Results). Usually, the stimulation activates the plantarflexors to produce moment, but also masks the sense of position, presumably because of the stimulation of the sensory nerves from the feet and ankles. The subject surrenders control to the artificial controller and has very little idea how well the artificial controller performs since there is no indication to him of the reference input to the angle controller. This transfer of control is shown in Figs. 2–7, for example, in Fig. 2, it occurs at 5 s.

Figs. 4–7 show that the tracking performance of the nested controllers with the intact subject can be excellent with reasonable transient response (time constant about 1.5 s). The narrow range of ankle angle is due to the available magnitude of the ankle moments at maximal activation compared to the upsetting moment of the inverted pendulum. For a 70 kg subject with CoG 1 m above the ankles, the static moment at each ankle (assuming an equal sharing of the required moment) is 24 Nm if the forward inclination is 4°. Greater moments will be required to prevent a falling-forward disturbance to decelerate the body mass. The intact subject may produce perhaps 50 Nm or more at each ankle by stimulation.

When we planned the Wobbler apparatus and the experiments, we wanted to know whether, if one could make the necessary measurements, one could design controllers using well-understood control engineering methods. These results from the intact subject show that this is so (but see the discussion of coronal plane motion of the paraplegic subject below). The nested control loop structure allows the controller to be built up and tested in several stages (muscle identification, moment controller testing, Imitation Standing, Actual Standing), and leads to a control structure which should be robust despite fatigue and spasticity. Typically, approximately 5–10 min are required for muscle identification and control design.

The LQG controllers, which are easy to tune with two "knobs" per feedback loop, have every broad optima for these knob settings: ρ_θ can take any value in the decade 0.0001 to 0.001. In our experiments, the best responses were obtained when both the position and moment control loops were designed with deadbeat observers (i.e., with $t_{\rm obs}^m = t_{\rm obs}^\theta = 0$). In general, however, it is still useful to have the option of increasing $t_{\rm obs}^m$ and $t_{\rm obs}^\theta$ as the sensitivity of the controllers to measurement noise will depend on the quality of the sensors used.

B. Results with Paraplegic Subject

Unlike the results from the intact subject, the results shown in Figs. 8–11 for the paraplegic are dominated by the effects of muscle weakness, fatigue and spasticity. These are manifested in the asymmetry of the activations and ankle moments, less apparent stability, and sometimes saturation of one or both the muscle activations.

- Less Stability: The paraplegic responses were less stable than those of the intact subject; so much so that the sampling rate of the angle-control loop had to be reduced to 6.7 Hz for the former. We presume that this tendency to oscillate was due to feedback actions of the subject's lower limb reflex arcs, giving a sort of clonus.

- Asymmetry: The paraplegic subject's plantarflexors were weaker on his left side, leading to higher activation to produce similar moments (see Figs. 8–10). The differential strength, and no doubt also spasticity, of the two ankles, led, despite their separate moment-control loops, to motion out of the sagittal plane. We had assumed that motion would occur only in the sagittal plane, and, for the intact subject, this has proved to be reasonable. However, significant transverse motion occurred for the paraplegic. To reduce this effect, we used a long light rope from the brace to one side to confine the motion approximately to the sagittal plane.
- Saturation of the Activation: The fact that sometimes the subject does not fall over when both activation levels have reached saturation (Figs. 9 and 11) can only be explained by the stiffness of the ankle joint which we have measured in this subject on a previous occasion as providing some 38% of the necessary stabilising moment [5, p. 220]. Presumably the cause of the stiffness is extension spasticity, but we did not investigate this by recording EMG.

These effects mask the changes in behavior of the controller due to the various parameter pairs used.

C. Discussion of the Control Methods

The controllers used in this work are linear except for the inverse recruitment curve (see Part 1, Fig. 13). We have shown that significant improvement in the consistency of the response of the moment control loop is possible if the Hammerstein muscle model is replaced by one with dynamics which change with activation level [7]. The "Local Model" approach described there seems to us much more elegant than the Radial Basis Function model we described in [8] and has the advantage that linear control methods are retained in the nonlinear controllers. We might wonder whether the controllers should take account of the muscle length and velocity, rather than treat them as if they were isometric as we do here. Such nonlinear controllers may have to be fully nonlinear, rather than a patchwork of local linearizations around a set of equilibria: scheduled local controllers cannot compensate unknown global dynamic terms [9], [10].

However, even though such a development might improve performance, in the paraplegic, such modifications are at present of minor significance compared to the major difficulties bulleted above. The same might be said of adapting the controllers for changing gain due to muscle fatigue; the fatigue itself is the major problem. We should comment that in situations where the muscle is not being used to maintain posture, so nearly isometric, and when it may be able to continue for long periods, the significance of these improvement to the controller might be quite different. For example, when controlling skeletal muscle which is surgically reformed for cardiomyoplasty and trained for extreme endurance, these improvements may be very significant [11].

To further improve the controllers in the Wobbler experiments, two possible significant improvements are outstanding.

1) At present (in the current LQG design), the values of the "control knobs" must be related to the muscle models, values from different muscles should not be compared directly. It would be much better if the knobs were model-invariant. One could then look for values which gave satisfactory performance from all subjects, because if these were found, no tuning of the controllers would be necessary; the controllers could immediately be determined from the muscle measurements. A possible method would be LQG with partial pole assignment.
2) Given the asymmetry which we have seen in the paraplegic muscles, and the tendency to cause coronal-plane motion, better use of the total muscle output should be possible if the pendulum model is made multivariable, and motion out of the sagittal plane is not prevented but measured as a further feedback signal.

D. Significance of Results for Paraplegic Standing

This work has highlighted the well-known limitations of functional electrical stimulation: spasticity and fatigue. Although the paraplegic subject of this work could produce moments in the region of 40 Nm for a few seconds, his maximal moments fell quickly to under 20 Nm and after a few tests to less than 10 Nm. In contrast, the intact subject could continue to produce about 50 Nm without discomfort for many tests. (It would be interesting to know why the difference is so great: in both cases the motor units will be recruited in nonphysiological order.) The effect of the difference is that while the intact subject can stand, inclined forward 2°, with muscles half activated (Fig. 3), and therefore with a reserve of moment to counteract disturbances (e.g., Fig. 5), the paraplegic's reserve is sapped after a few seconds (Fig. 9). This problem may be mitigated by use of implanted stimulating electrodes to ensure that all the motor units in the muscle can be recruited, by more frequent training of the muscles [12], [13] and possibly by selective stimulation so that slow motor units are recruited first [14].

Spasticity is unpredictable from moment to moment, even if the paraplegic is aware of its average level on any particular day. It may appear as spasms or joint rigidity due to coactivation of antagonists, and episodes may last for seconds or minutes. During that time we may see the activation signal jump from extreme to extreme as the controller endeavours to maintain the correct moment (no example is shown in Section V Discussion of Results).

If the paraplegic were standing with artificial balance control, these effects of fatigue and spasticity would cause falls. In fact, of course, when paraplegics stand outside the laboratory, they do not yet do so out of reach of support handles of some sort, and these they use both when the leg muscles become fatigued, to help support the body weight, and when the stimulator-controller cannot correct for disturbances, whether internal, such as leg muscle spasm, or external, such as lifting a weight [15]. Even if transient disturbances must be counteracted by resort to handles, the controller may still be useful for freeing the hands during nonspastic and undisturbed periods.

The Wobbler experiments were conceived because we thought that there was a wide gulf between the feedback control experiments which have been done on healthy animal preparations using comprehensive instrumentation, and the often rather informal clinical tests of feedback controllers—a complicated system in which many variables were not measured. The results from this and previous papers show quantitatively the well known but usually only qualitatively described effects of fatigue and spasticity on the performance of a properly designed robust controller.

This paraplegic subject, who is the only one we have tested in the Wobbler, exhibits severe spasticity, so he is an unfavorable subject. We expect that a less spastic individual would show results more like the intact subject while his muscles were not too fatigued. However, the standing endurance will always be limited by fatigue, and this can easily be assessed in new subjects simply by measuring the time for which the stimulated plantarflexors can produce moments at each ankle of at least, approximately, 30 Nm.

Imitation Standing allows dynamic testing of the controller and plant together under realistic conditions but without risk to the subject. Its value is shown by the results described in section 2.5, where the system exhibited poor stability which was corrected by changing the angle-loop sampling frequency from 20 to 6.7 Hz during subsequent Imitation Standing tests before Actual Standing was attempted. On the other hand, a shortcoming of the Wobbler apparatus is now evident: it cannot be used to measure the ankle stiffness during Imitation or Actual Standing, and consequently the observation that the paraplegic remained in balance despite saturation of the muscle activations (Figs. 8 and 11) could not be predicted. The Wobbler can be used to measure ankle stiffness in separate tests, but given the unpredictability of the stiffness due to spasticity, results from separate tests do not indicate the subsequent stiffness during standing controller tests. A possible development of the apparatus would allow the stiffness to be measured continually by applying small rapid angular displacements, perhaps during the 150 ms between angle samples, and measuring the change in the ankle joint moments. Techniques of this sort have been used by Robinson *et al.* [16] and Anderson and Sinkjaer [17].

These experiments have been conducted with two features which would not be present in a system designed for daily use: the body brace and the transducers. The transducers, which are mounted on the Wobbler shaft, are high-resolution with good absolute accuracy; it is unimaginable that similar sensors could be mounted in the footwear or implanted in the body. We chose these transducers because we did not want the performance of the controller to be limited by the transducers in the laboratory experiments and results like Fig. 5 show what is possible (angle stabilized to within $\pm 0.1°$).

The body can be regarded as a chain of segments, either open, if the standing is not "supported" by the hands, or closed, if it is supported. By use of the brace, these experiments involve a one-link open chain without interference from the intact upper body, since the trunk and head are restrained by the brace and the arms are held immobile. Without the body brace, but with support from the hands, the chain is closed which means that the body is largely controllable by the intact nervous system acting through the arms: feedback control from the handle forces may then be used to determine the leg muscle stimulation intensities [18]. When the hands do not support the body, the open chain is partly under voluntary control and the intriguing question arises: how should the stimulated paralyzed muscles be controlled using sensors on the paralyzed and intact parts of the body? One possibility, making explicit use of the voluntary activity of the trunk muscles, is being investigated by Matjacic *et al.* [19].

At present, our feedback control scheme for unsupported standing is the only one which has been implemented in experimental trials with paraplegics. A number of authors have proposed alternative approaches which have been tested only in simulation models [20]–[22].

VI. Conclusion

We have demonstrated a system which provides artificial balance to a paraplegic. The subject is braced above the ankles and stands in apparatus which allows both ankle plantarflexion moments and the common ankle angle to be measured. The controller is made robust by having three feedback loops. The controller is LQG, with two tuning "knobs" per loop. Muscle identification and controller tuning can be done quickly. We found that the optimal performance was acceptable for a wide range of control weighting (ρ) knobs. The major difficulties we encountered, when testing one paraplegic subject, but not the intact subject, were muscle weakness (fatigue) and spasticity. Together these limited the balance time to no more than a minute.

The value of setting up and testing FES control systems in special apparatus like the Wobbler, in enabling the system to be understood, seems to us to be beyond doubt.

We conclude that the "control" problem, in the development of functionally useful controllers for standing without support from the hands, is to devise a system in which the artificial controller acts in concert with the intact natural motor control system, using only practicable sensors. However, significant progress will be limited unless we can increase the muscle endurance and, in some patients, reduce unwanted spastic effects.

Acknowledgment

The authors would like to thank the British Medical Research Council for enabling them to build the Wobbler apparatus. They are grateful to our paraplegic volunteer for giving his time for their research.

References

[1] K. J. Hunt, M. Munih, and N. Donaldson, "Feedback control of unsupported standing in paraplegia. Part I: Optimal control approach," *IEEE Trans. Rehab. Eng.*, this issue, p. 331–340.
[2] K. J. Hunt, M. Munih, N. Donaldson, and F. M. D. Barr, "Optimal control of ankle joint moment: Toward unsupported standing in paraplegia," *IEEE Trans. Automat. Contr.*, to be published.

[3] K. J. Hunt, M. Munih, N. Donaldson, and F. M. D. Barr, "Investigation of the Hammerstein hypothesis in the modeling of electrically stimulated muscle," *IEEE Trans. Biomed. Eng.*, to be published.
[4] N. Donaldson, M. Munih, G. F. Phillips, and T. A. Perkins, "Apparatus and methods for studying artificial feedback-control of the plantarflexors in paraplegics without interference from the brain," *Medical Engineering and Physics*, 1996, to appear.
[5] N. Donaldson, F. M. D. Barr, G. F. Phillips, and T. A. Perkins, "Unsupported standing of paraplegics by stimulation of the plantarflexors: Some results from the wobbler apparatus," in *Neuroprosthetics: From Basic Research to Clinical Application*, A. Pedotti, M. Ferrarin, J. Quintern, and R. Riener, Eds. New York: Springer-Verlag, 1996.
[6] D. A. Winter, *A. B. C. (Anatomy, Biomechanics and Control) of Balance During Standing and Walking*, Waterloo, Ont., Canada: Waterloo Biomechanics, 1995.
[7] K. J. Hunt, M. Munih, and N. Donaldson, "Investigation of a nonlinear strategy for moment control in electrically stimulated muscle," in *Proc. IFAC Symp. Modeling and Contr. Biomed. Syst.*, Warwick, England, 1997.
[8] N. Donaldson, H. Gollee, K. J. Hunt, J. C. Jarvis, and M. K. N. Kwende, "A radial basis function model of muscle stimulated with irregular inter-pulse intervals," *Med. Eng. Phys.*, vol. 17, pp. 431–441, Sept. 1995.
[9] H. J. Chizeck, "Adaptive and nonlinear control methods for neural prostheses," in *Neural Prostheses*, R. B. Stein, P. H. Peckham, and D. B. Popovic, Eds. Oxford, England: Oxford University Press, 1992, pp. 298–328.
[10] K. J. Hunt and T. A. Johansen, "Design and analysis of gain-scheduled control using local controller networks," *Int. J. Contr.*, vol. 66, no. 5, pp. 619–651, 1997.
[11] H. Gollee and K. J. Hunt, "Nonlinear modeling and control of electrically stimulated muscle: A local model network approach," *Int J. Contr.*, 1997, to appear.
[12] A. Kralj and T. Bajd, *Functional Electrical Stimulation: Walking and Standing After Spinal Cord Injury*. Boca Raton, Florida: CRC Press, 1989.
[13] I. D. Swain, "Conditioning of skeletal muscle by long-term functional electrical stimulation—Implications for the development of practical systems," in *Spinal Cord Dysfunction, Vol. III: Functional Stimulation*, L. S. Illis, Ed. Oxford, England: Oxford University Press, 1992.
[14] J. T. Mortimer, "Motor prostheses," in *Handbook of Physiology—The Nervous System II*, J. M. Brookhart, V. B. Mountcastle, V. B. Brooks, and S. R. Geiger, Eds. Bethesda, MD, Amer. Physiol. Soc., 1981, pp. 155–187.
[15] T. Houdayer, R. Davis, S. Emmons, B. Andrews, J. Patrick, and A. Mortlock, "Hybrid paraplegic standing with the cochlear FES-22 implantable stimulator and Andrews FRO braces: A follow-up," in *Proc. IEEE EMBS Conf.*, Amsterdam, The Netherlands, 1996, pp. CD-ROM paper 478.
[16] C. J. Robinson, B. Flaherty, L. Fehr, G. C. Agarwal, G. F. Harris, and G. L. Gottlieb, "Biomechanical and reflex responses to joint perturbations during electrical stimulation of muscle: Instrumentation and measurement techniques," *Med. Biologic. Eng. Comput.*, vol. 32, no. 3, pp. 261–272, 1994.
[17] J. B. Anderson and T. Sinkjaer, "An actuator system for investigating electrophysiological and biomechanical features around the human ankle joint during gait," *IEEE Trans. Rehab. Eng.*, vol. 3, pp. 299–306, Dec. 1995.
[18] N. Donaldson and C.-H. Yu, "FES standing: Control by handle reactions of leg muscle stimulation (CHRELMS)," *IEEE Trans. Rehab. Eng.*, vol. 4, Dec. 1996.
[19] Z. Matjacic, T. Bajd, M. Gregoric, H. Benko, and P. Obreza, "Trunk muscles strength evaluation in normal and paraplegic subject," in *Proc. 9th Int. Conf. Mechan. Med. Biol.*, Ljubljana, Slovenia, 1996, pp. 437–440.
[20] K. Lee, H. J. Chizeck, and W. Lin, "Trajectory tracking of multi-link systems with input delay via simultaneous modulation of torque and stiffness," in *Twenty-seventh Neural Prosthesis Workshop*, Washington, DC, Oct. 1996.
[21] J. J. Abbas and H. J. Chizeck, "Neural network control of functional stimulation systems: Computer simulation studies," *IEEE Trans. Biomed. Eng.*, vol. 42, pp. 1117–1127, 1995.
[22] G. Khang and F. E. Zajac, "Paraplegic standing controlled by functional neuromuscular stimulation: Part II—computer simulation studies," *IEEE Trans. Biomed. Eng.*, vol. 36, pp. 885–894, 1989.

Marko Munih (M'88), for photograph and biography, see this issue, p. 340.

Nick de N. Donaldson, for photograph and biography, see this issue, p. 340.

Kenneth J. Hunt, for photograph and biography, see this issue, p. 340.

Fiona M. D. Barr received the diploma from the Bristol School of Physiotherapy and the M.Sc. degree at Kings College London, London, U.K. She is currently working towards the Ph.D. degree with a dissertation on the quantification of spasticity.

She is a Research Physiotherapist at the Royal National Orthopaedic Hospital, London, U.K., and has had much experience in the application of FES to patients with spinal cord injury. Her particular interest is the effect of electrical stimulation on spasticity.