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Apparatus and methods for studying artificial feedback-control of the plantarflexors in paraplegics without interference from the brain

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ABSTRACT

Apparatus has been built to explore the practical feasibility of using automatic control with electrical stimulation of paralysed legs to restore function. The experiments are performed with paraplegics with the aim of achieving a realistic postural task: to see whether the body may be maintained upright by stimulation of the plantarflexors when the other joints are braced. Significantly, the intact upper body, under natural control of the brain, cannot interfere with the automatic control. The "Wobbler" apparatus allows measurement of the ankle muscle properties in isometric conditions or in sinusoidal motion. Using the biomechanical properties of the body, which are also measured, controllers for stabilising the body can be designed. Controllers can be dynamically tested, imitating anterior–posterior sway, while the body is held upright, before "actual standing" is attempted. © 1997 IPEM. Published by Elsevier Science Ltd

1. INTRODUCTION

Experiments in which muscle stimulators are used with feedback from sensors to obtain automatic control of force or position have been performed for over 25 years, since Vodovnik *et al.*[1]. Chizeck has written an up-to-date review[2]. The advantages of using feedback control are well known and would be valuable in FES: the system is less affected by changes in the plant (e.g. muscle potentiation) and by external disturbances.

Much of the experimental work with feedback controllers has been done on animal preparations[3,4]. This allows many of the complications which occur when stimulating paralysed humans to be avoided. By stimulating with cuff electrodes on the motor nerve or with intramuscular electrodes, cutaneous reflexes are avoided. By attaching a force transducer to a transected tendon, the effect of contraction of other muscles at the same joint are neglected. If, in an acute experiment, the motor nerve itself is cut proximal to the electrodes, the muscle response becomes independent of activity in the spinal cord. The quantitative significance of these experiments is also difficult to assess since the animals are healthy, the muscles have not undergone the changes which follow upper motor neurone paralysis[5], and there is no real functional task with which to test the controller–stimulator muscles.

The difficulties, which occur when stimulating paraplegics with surface electrodes, include coactivation, sometimes of antagonist muscles, spasticity and spasms. The joints may not move freely, due to contractures and useful muscles may not be excitable due to loss of their lower motor neurones. There may also be interference from the cutaneous reflexes. Muscles which are stimulated continuously, so as to produce near maximal force, weaken, typically within a few minutes. This loss of force is accompanied by a reduction in the speed of response of the whole muscle, as the fast-acting motor units tire more rapidly. However, the influence of these effects on the motion are often not measured in clinical tests of controllers. Indeed, analysis of the behaviour of body as the system would, in general, require the dynamics of the whole body to be observed since the intact upper body can affect the paralysed limbs dynamically and statically[6].

Controllers for standing are of two types: those for supported and those for unsupported standing. In supported standing, the intact upper body is used for balance, as the hands hold crutches, a

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Walker, or any convenient handles. In this case, the essential purpose of stimulation is to cause sufficient knee extension to lift a significant part of the body weight. For example, Ewins *et al.*[7] developed such a system for clinical use. Their system maintains the knee near hyperextension during standing; feedback of the knee angle modulates the stimulation so that just sufficient extending moment is produced, and this minimises the rate of fatigue and prolongs standing.

Could some paraplegics be given controllers which, following a command, would provide unsupported standing so that the arms and shoulders, freed from the task of balancing the trunk, could perform manipulative tasks? This problem has been modelled by Khang and Zajac[8] and Bahrami *et al.*[9] However, to translate these studies into trials with human paralysed subjects, three difficulties must be faced:

- The stimulated-muscle and biomechanical parameters must be measured.
- Some means must be found to provide assurance that the controllers are bug-free before using them in a way which would expose the user to unacceptable risk.
- Finally, the ability of their controllers to cope with spasm, spasticity and fatigue must be assessed.

It seems therefore, that some means of dynamically testing the system under measurable conditions is required.

We decided to start by investigating a simpler controller than the Multi-input, Multi-output controllers needed to control the hips, knees and ankles, yet one which could still be used for a realistic purpose. There are several advantages in working with the ankle joints: when standing, they are not at an extreme position in their range of motion; their extensor muscles (plantarflexors) are easily accessible for surface stimulation; and the joint moments can be estimated with little error from the forces under the feet, because the inertia of the feet is small. Following Trnkoczy *et al.*[10] and Bajzac and Jaeger[11], we chose to see whether, by stimulation of these muscles, we could stabilise the subject upright, acting as a single inverted pendulum, with one rotational degree of freedom in the ankle joints and without hand support, while his other joints (knees, hips, spine) were braced.

However, unlike these previous investigators, we wished to be able to apply motion to the ankle joints. This allows the effects of motion to be measured; like Weiss *et al.*[12], Sinkjaer *et al.*[13] and Flaherty *et al.*[14], we describe the moments due to motion as joint stiffnesses. Also the responses of the controllers can be measured in realistic conditions. To do this we rock the feet while the body is held upright in its brace by ropes, to imitate the body swaying while the feet are stationary on the ground.

We named the apparatus the Wobbler. It has four modes of use: (1) fixed position, while measuring the moments at the two ankles in isometric conditions with the body fixed; (2) rocking the feet with the body fixed while the stimulation intensity is steady (including zero), to measure ankle stiffness at various stimulation levels; (3), during "imitation standing", again with the body fixed, sinusoidal displacement is imposed and the moments measured. In the latter mode, the controllers are active. (4) In the final mode, "real standing", the body has a restricted range to sway, the feet are fixed, the moment and position controllers are active and the angle of the body is measured with an inclinometer. In all these experiments, the behaviour of the ankle joints can be observed without interference from the subject's intact neuromuscular system (i.e. above the spinal cord lesion) since the upper body is braced and the arms are inactive.

The questions we wished to consider are as follows. The references are to papers that have been published or are in the process of review.

1. What is an adequate model of the electrically stimulated isometric plantarflexors in normal and paraplegic subjects[15]? What method of model identification is best?
2. How much do the model parameters vary for one person from day to day?
3. How much variation is there between individuals?
4. How does the stimulation intensity to joint moment transfer function for the ankle muscles change with ankle angle?
5. How does the muscle change with fatigue? How often must the plant be identified?
6. How well can we control the isometric joint moments[16]?
7. How should the ankle stiffness be modelled? For the small displacements of the ankle joint during standing, is it valid to design and test linear moment controllers (under isometric conditions), but then use these with motion, applying some correction for the joint stiffness[17]?
8. After we have designed an ankle angle controller and tested it by "imitation standing", can we achieve "real standing" of the paraplegic subjects[18–21]?

In this paper, we present full details of the Wobbler apparatus, hardware and software, and the methods used for measuring the biomechanical properties of the subjects.

2. APPARATUS

2.1. The Wobbler

The apparatus is shown in *Figure 1* (built by Gorrett Engineering, Bridgend, Mid-Glamorgan, UK). The power to rock the feet comes from a d.c. motor with a separate field winding (2.7 V, 3.5 A), and the speed set by a the voltage (0–300 V, up to 1 A) on the armature. This speed of rotation is reduced by three toothed 20 mm belts (Davall Ltd, Welham Green, Hatfield, Herts) which drive a flywheel with a moment of inertia
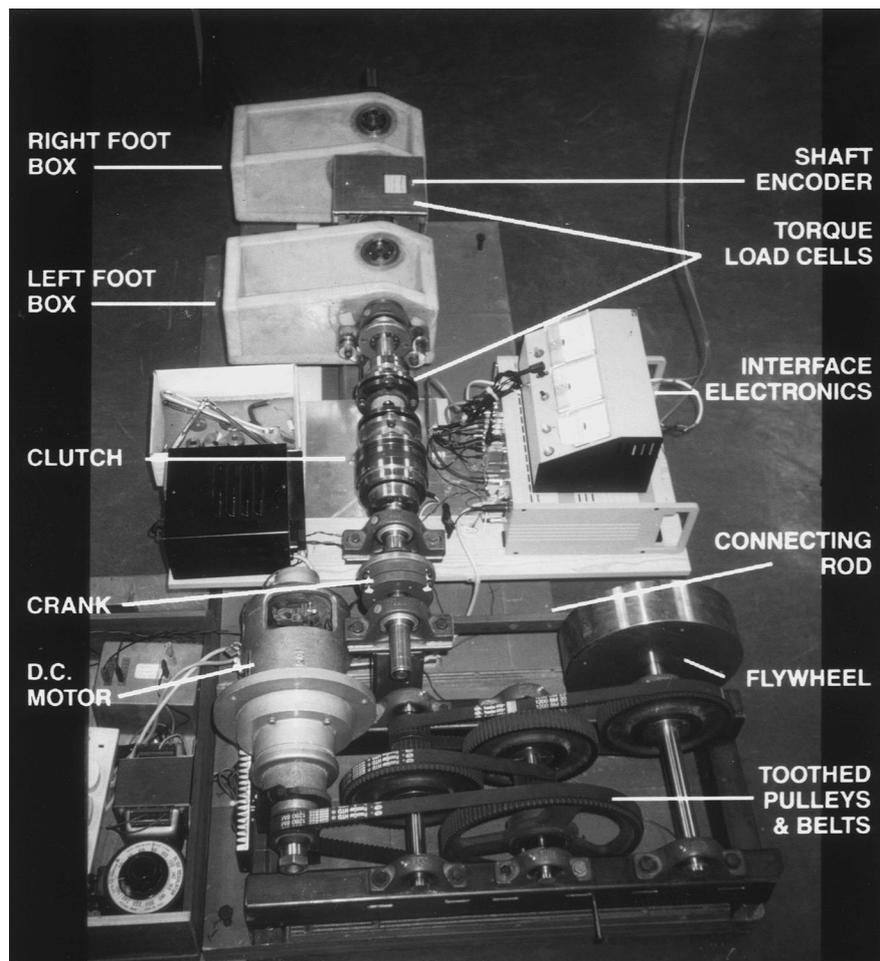
Figure 1 The Wobbler apparatus, looking along the main shaft.

of 0.33 kg.m². By changing one of the belts between pairs of pulleys, the speed reduction ratio may be chosen as 0.019 or 0.21, giving maximal wobble frequencies of 1 or 6 Hz, respectively. An eccentric spigot on the flywheel, joined through a connecting rod to a crank on the 30 mm diameter rocking shaft, allows the amplitude of the motion to be preset. There are open-top boxes for each foot so that the ankle joints can be aligned with the shaft. The rocking shaft drives the left box, then there is a short coaxial shaft between left and right boxes, and finally a stub of shaft to the right of the right box. The bearings are Plummer Blocks, on both sides of both boxes, to carry the weight of the subject. There is a 150 Nm torque load cell to the left of the left box (purpose-built by Datum Electronics Ltd, Newport, Isle of Wight), and one between the boxes, so that both ankle moments can be measured. There is one shaft encoder, also mounted between the foot boxes: the encoder disc is held on the shaft by a bush (ERO 1324, Heidenhain (GB) Ltd, Burgess Hill, West Sussex). The resolution of the encoder is 0.018° i.e. there are over 55 steps per degree of ankle rotation. A clutch, which is held engaged by an electromagnet, is in line with the rocking shaft, before the first load cell (Monninghoff Type 549, rated at 150 Nm, with Arcoflex couplings; Simpla-troll Ltd, Bedford MK41 0HT).

The foot boxes were designed to be torsionally stiff without having large angular inertia. The shape was defined by joining five pieces of Styro-foam (25 mm thick with a approximate density of 800 kg/m³). Each box was reinforced by covering with glass-fibre mat and impregnating with epoxy resin. To transmit the torque from the shaft to the box, 5 mm thick cast aluminium alloy flanges were embedded in the foam before reinforcement, and the epoxy bonded the glass to the alloy. Inside the flanges, expanding bushes clamp the box to the shaft. However, the first boxes, built like this, slipped on the shaft; this was due to the alloy creeping under the large hoop stress applied by the expanding bushes. In the second design, which overcame this problem, steel rings were glued inside the alloy flanges, so that the steel carried the hoop tension.

The moment of inertia of both boxes and the subjects Trainers and shoe plates (see below) was measured by wobbling these at four frequencies in the absence of the subject, while measuring the sinusoidal torque required. When peak moment was plotted against ω^2, the slope of the line was 0.0134 Nm/(rad/s)² with a standard error of 0.00014. As the amplitude was 8°, this gives the inertia as 0.096 kg.m² S.E. 0.001.

The torsional stiffness was measured from the left box, with the shaft clamped, as 79 Nm/deg. From the right box, with the shaft clamped, it is 53 Nm/deg. The stiffness between the two boxes is therefore $(79{*}53)/(79 - 53) = 161$ Nm/deg.

2.2. Inclinometer

The inclinometer comprises a high-resolution, 2% linearity potentiometer with a double pulley and an inextensible thread. The potentiometer, mounted on an arm which is attached to the framework on the ceiling, is at the height of the subject's chest and 1600 mm behind the axis of the ankle joints. One end of the thread is tied to the brace on the subjects' back (see below), at the same height as the potentiometer. The thread passes round a pulley and hangs down to a 60 g weight (small compared with the mass of the subject). The gain can be adjusted by selecting one of the pulleys. The angle resolution with the largest pulley is 0.014°, with worst case peak-to-peak noise 0.03° within an expected range of 14.5°.

2.3. Electrical interface

The computer receives signals from the two torque load cells and a tachometer on the motor through analogue-to-digital converters. Digital signals come from the shaft encoder; also to indicate that the clutch is engaged; and a synchronising pulse from the stimulator. The computer sends signals via serial link to the stimulator. The motor speed is set manually using the 500 VA autotransformer which feeds a bridge rectifier and thence the motor armature.

2.4. Safety features of the Wobbler

The experimenter can only engage the clutch by operating a switch manually, and this should only be done while the drive shaft is at rest. The hardware is arranged so that the program could disengage the clutch if excessive moments were detected but this has not been implemented. Actually it can be disengaged by the experimenter; or by the subject, should he wish, with a "quick release" button close to hand. This arrangement lets the subject avoid continuing discomfort if a test were to be painful.

The clutch allows the footboxes to be set at any angle, which is convenient during the procedure for getting the subject into the apparatus (see below) and also to accommodate ankles with plantargrade contractions. However, there is a danger that the clutch will be engaged near one end of the range of motion of the joint and subsequently, by mistake, when wobbling is started, the joint will be driven outside its normal range of motion. To protect the ankles from this risk, there are mechanical rotation stops between the clutch and the foot boxes. It is important to ensure that these are set at the beginning of a session of tests. If this mistake were made, and rotation of the boxes abruptly halted by one stop, an enormous torque would be exerted which would tend to damage the mechanical drive. To protect the apparatus, therefore, a "Wedgepin" torque limiter (S500, Howden Ltd, Stroud, Glos, UK) is included after the crank, which will shear if the torque exceeds 150 Nm (see *Figure 1*).
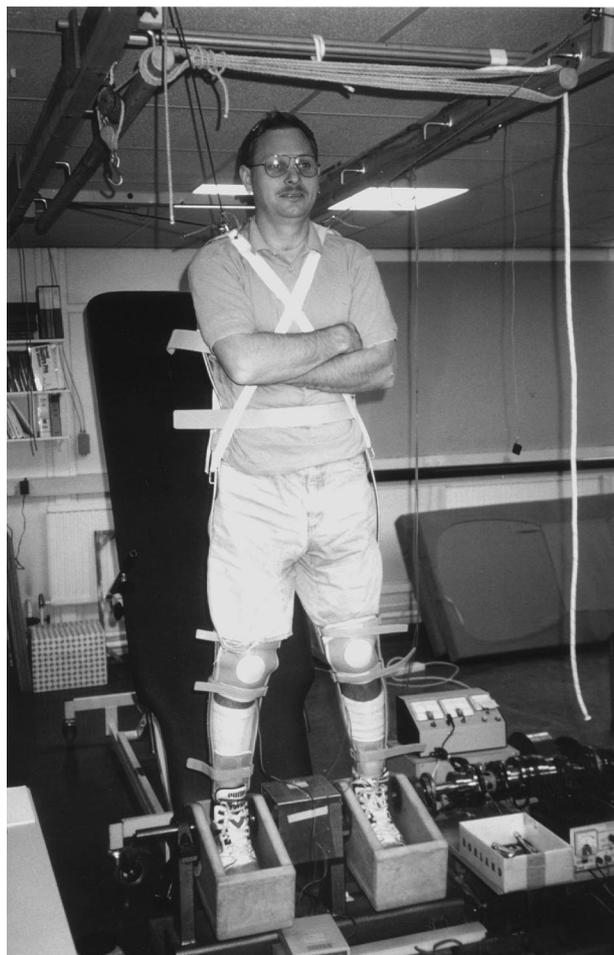

Figure 2   Subject in Wobbler wearing brace.

2.5. Holding the feet

Our original intention was to use the method of Weiss *et al.*[22] to support the feet in the foot boxes. In this method, plaster positives of the feet are made and suspended in moulds which are the size of the boxes. Polyurethane foam, of an orthotics grade, is then cast into the moulds. After solidifying, the foam and plaster feet are removed from the moulds; and each block of foam is cut into two halves, one on each side, which can be removed from the plaster. These should then hold the subject's feet snugly in the boxes. However, our first subject preferred us to sacrifice a pair of his Trainer shoes and we glued and bound those to 3 mm Duralumin plates which were bolted into the boxes instead. The subjects' feet are always inspected immediately after each session of experiments. With feet in Trainers, we have never thought that there had been significant danger of pressure sores forming.

2.6. Brace

We discussed what we wanted with the orthotists at the Royal National Orthopaedic Hospital and settled on a polythene half-shell which is strapped onto the back of the head, trunk and legs, down to the shanks (*Figure 3*). To form this shell, a whole-body cast is taken in the Plaster Theatre. The shell is made of three plastic parts, joined and
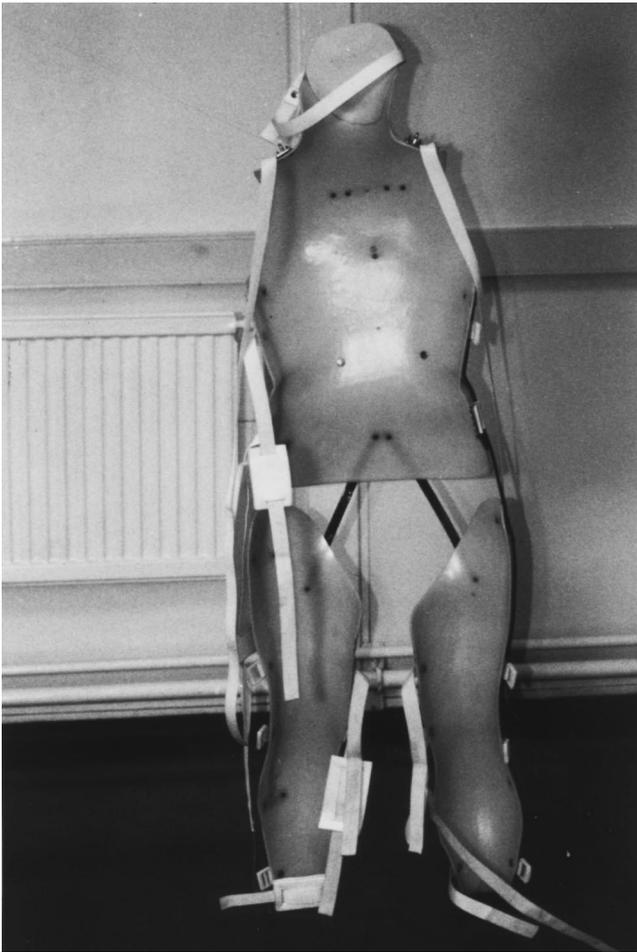
Figure 3 Body brace.

reinforced by steel strips. There are three straps round each leg, two round the trunk and diagonal straps across the shoulders (*Figure 2*).

2.7. Supports

Parallel 2 m long hand-rails are fixed below the ceiling of the laboratory, over the Wobbler (*Figure 2*). From cleats on the ends of these rails, light 6 mm diameter ropes run to snap hooks which clip to eyes on the shoulders of the body brace. These ropes may be cleated tight, to hold the body more-or-less upright, of with some slack to allow fore-and-aft sway to determined limits.

2.8. Stimulator

The stimulators used in this project are portable programmable 8-channel current-output surface electrode stimulators designed and built at Stanmore[23]. For exercise purposes, the patient has one unit at home. For laboratory use, the stimulator is driven directly in real time through a serial link optical isolator by the PC which is running the experiment. In this application, the stimulation pulse frequency is constant at 20 Hz, and muscle activation is varied by modulating pulse width from 0 to 500 μs, corresponding to activations from 0 to 1000 mAct[16].

3. SOFTWARE

Various experimental procedures allowed by the Wobbler hardware are supported with a custom-written real-time programs, display programs, data conversion programs and MATLAB scripts and functions (*Table 1*). All programs, except MATLAB, run under DOS operating system. The binary data, collected with the real-time programs, are checked immediately with graphical display programs, converted into an ASCII record, and further processed using the MATLAB package. To minimise the time that the subject must remain upright in the Wobbler, the sequence of tests is pre-programmed into a short sequence of a batch files. There is a consistent format for the names and contents of the data files and for the programs that act on these data. In some cases, the parameters used in one test are derived from the data yielded by a previous test in the same session. The various experiments' set-up data are passed through files which are themselves specified on the DOS command line. In addition, there are also supplementary files containing information about recruitment, defining the moment and position controller parameters, moment and position reference values, transducer offsets and calibration constants.

3.1. Real-Time program library

To maintain consistency, all the software is built around a central library of functions. The library provides facilities to undertake the following tasks:

- Establish a real time interrupt which is used for timing data acquisition at 200 Hz (or 20 Hz) and stimulation output at 20 Hz.
- Initiate automatic timed analogue-to-digital conversions on any arbitrary pattern of signal channels including the motor speed, left or right or left plus right shaft torques (subtraction is performed by an operational amplifier circuit), filtered torques, activation analogue or whole-body inclination, while using offset values stored in a disk file for adjusting the zero levels of all channels.
- Provide a clock output signal to drive the shaft angle encoder, and a routine to read the angle through the digital ports.
- Establish interrupt-driven buffered serial communications between the PC and the stimulator; put the stimulator into "laboratory mode" ready to receive pulse delivery instructions; deliver test pulses to check that the patient electrodes have the correct impedance, and report the status of these checks.
- Deliver individual pulses, comprising an error checking transmission protocol, a means of creating a voltage analogue of the stimulus activation level for hardware filtering, and a means of detecting the timing of the actual pulse.
- Store acquired data in memory during a test and automatically write these data to file on completion of the test.
- Read in the recruitment data from file, check

for multiple peaks or points of inflexion, convert this to an inverse recruitment, and provide a data structure from which the corresponding pulse width can be read for stimulation.

- Implement Linear Quadratic Gaussian (LQG) left and right moment controllers and the LQG position controller[16]. In earlier work[17], PID moment controllers and PD position controllers were used.
- General "housekeeping" functions such as opening and closing files, allocating memory, and restoring interrupt vectors.

3.2. Real-Time programs (tests)

These programs define the experiments which run in real-time for estimation of muscle parameters and closed-loop controller tests. In Tests A, B, C, G, PRBS and M (reference test), the ankle joints are isometric, with the body braced and fixed by the ropes. During Tests H, M (disturbance test) and P, the body is still fixed but the feet are rocked. In actual standing (Test R), the feet are fixed and the ropes are made slack to enable the body to sway as a single inverted pendulum in the sagittal plane.

A Current selection and maximum moment estimation. Stimulation is given for three seconds at constant predetermined current (5 mA to 120 mA), but steadily rising pulse width from 50 to 500 μs, and the resulting ankle moment value is immediately displayed versus time.

B Twitch response measurement. Prior to the experiment, the ten pulse widths in the range 50, 100,..., 500 μs are put in a random order to form a sub-group. Five sub-groups with different random orders are formed and then combined into a group of fifty. These pulses are delivered at intervals of 1.2 s while recording moment at 200 Hz.

C Muscle low frequency gain. A short burst of tetanic stimulation (20 Hz) is applied for 5 s, long enough to find the steady moment after the transient is complete.

G Sinusoidal activation. This is an open-loop test where muscle activation is varied sinusoidally at several predetermined frequencies, while recording the resulting moment at 200 Hz. The frequency response

Table 1

Real-time programs	Specified in command line	Screen	OutFile format	OutFile prefix	Sample freq. [Hz]	DOS plot	Conversion	MATLAB scripts
TESTA	⟨Channel⟩ ⟨Current⟩	graph	A	a1	200			testa
TESTB2	⟨Channel⟩ ⟨Current⟩ ⟨PWstep⟩ ⟨OutFile⟩	messages	B	b2	200		convb	testb2
TESTC	⟨Recruitment⟩ ⟨OutFile⟩ ⟨Activation⟩	messages	C	c1	200		convc	testc
TESTF	⟨RecFile⟩ ⟨OutFile⟩ ⟨ActFile⟩	messages	F	f1	200		convf	testf
TESTG	⟨RecFile⟩ ⟨OutFile⟩ ⟨ActFile⟩	messages, activations	G	g1	200		convg1, convg2	plotg, testg
TESTH3	⟨RecFile⟩ ⟨OutFile⟩	messages	H	h3	200		convh	ploth3
TESTPRBS	⟨RecFile⟩ ⟨PRBS File⟩ ⟨OutFile⟩	messages, frequency	M	e1	20	plotprbs	convm5	testprbs
TESTMK5	⟨RecFile⟩ ⟨L Par⟩ ⟨R Par⟩ ⟨TorqRefFile⟩ ⟨OutFile⟩	messages, activation bars	M	m5	20	plotm5 (l/r)	convm5	testm5, testm5d
TESTPK9	⟨RecFile⟩ ⟨Lpar⟩ ⟨Rpar⟩ ⟨POSpar⟩ ⟨RefPosFile⟩ ⟨OutFile⟩	messages, actual parameters	M	p5	20	plotp (l/r)	convm5	testp
TESTRK92	⟨RecFile⟩ ⟨Lpar⟩ ⟨Rpar⟩ ⟨POSpar⟩ ⟨PosRefFile⟩ ⟨OutFile⟩	messages, actual parameters	M	r5	20	plotr	convm5	testr

Nomenclature: ⟨Channel⟩1 or 2 for Left or right channel; ⟨Current⟩ Current in mA in 10 mA steps ⟨PWstep⟩ normally 50 μs; ⟨Activation⟩ from 0 to 1000 mAct, see text for explanation; ⟨RecFile⟩ Recruitment file, measured in testb2; ⟨ActFile⟩ Activation file; ⟨PRBS File⟩ PRBS sequence file; ⟨L Par⟩ ⟨R Par⟩ Left and right moment controller polynomials; ⟨TorqRefFile⟩ Torque reference file for moment controller; ⟨POSpar⟩ Position controller polynomials; ⟨PosRefFile⟩ Position reference file for position controller.

Note: All programs read zero.dat file with AD converter offsets and calib.dat with torque load cell calibration data. A few more constants are linked into the programs' code.

may be later used for muscle identification.

PRBS PRBS signals. This is an open-loop test where muscle activation is varied as a pseudo-random binary sequence. The PRBS sequence is generated before the experiment using MATLAB. The program takes the sequence from a file, uses the inverse recruitment curve to convert to pulse widths, applies the stimulation at 20 Hz and measures the moment response. This program may be used more generally for other excitation signals appropriate for system identification.

H Stiffness. The muscle is stimulated at constant activation levels of 0, 200, 400, 600, 800 and 1000 mAct, while wobbling the feet at speeds of 0.3, 0.5, 0.8 and 1.0 Hz and acquiring moments at 200 Hz. The inverse recruitment curve is included as a lookup table between the required activations and the delivered pulsewidths. The range of the angles may be preset mechanically to 1.7, 3.5, 5.2 or 7° peak-to-peak with the eccentric (2.1). The amplitude of the variation in moment gives the joint stiffness resulting from passive factors as well as from the muscle activity.

M Isometric moment controller. The LQG or PID controllers may be used in two different modes of operation: reference tracking and disturbance rejection. In both cases, the required moment values are read from a file and the moments and angle values are simultaneously stored at 20 Hz. Feedback moment signals are measured from load cells on the shaft. Reference tracking is a closed-loop version of test G, described above, for which any interesting reference input can be applied. In the disturbance test, the feet are wobbled, as in the stiffness test, while the reference moment is constant and activation levels are calculated by the controller. In this case, the moments due to ankle stiffness are a disturbance to the output and strong attenuation of the sinusoidal moment indicates good controller function. The other input data in this test include the recruitment curve and controller parameters.

P Imitation standing. Both left and right moment controllers are acting together, with the equal reference moment values being calculated from the position controller in the outer control loop. The position reference is prescribed in advance. During the test, the input is read from file and the actual ankle angle is acquired from the shaft encoder. Data acquisition and input files are similar to those in the moment controller tests, but now the position reference values are specified at 6.7 Hz intervals. The subject remains fixed vertically, while the ankles are slowly wobbled (0.1–0.3 Hz). This is analogous to "actual standing", when the feet remain horizontal and the subject sways.

R Actual standing. The restraints, which until now have held the subject upright, are slacked, allowing the subject free forward/backward sway, but preventing him from falling. The feet are fixed at a neutral position of the ankles. The program is the same as test P, but the measured input (feedback) angle is now obtained from the inclinometer, not the shaft encoder.

3.3. Auxiliary programs

Further programs have been written for displaying and converting measured data, and for testing the Wobbler hardware. These include:

- A program to detect the peak moments of test B, and to sort these according to pulse width. Five responses at each pulse width are averaged to yield the shape of the recruitment curve.
- Programs to display the results of the controller tests graphically immediately during tests. These are very much quicker than using the plot facilities in MATLAB.
- Programs to convert all the various types of a compact raw data files into the (verbose) ASCII files appropriate for MATLAB input.
- Programs to test analogue channel signals and the encoder in the Wobbler apparatus.
- Programs to create files of data, such as reference moments or positions.

3.4. MATLAB scripts

The ASCII data files are read into MATLAB for display and processing. This is hastened by having a library of several MATLAB scripts for each test. Data processing includes: transfer function identification based on twitch response (B), frequency response (G) or PRBS activation response (PRBS); estimation of stiffness; design of moment and position PD, PID and LQG controllers; and simulation scripts for moment loop and position loop with MATLAB/Simulink. These simulations provided options for reference signal tracking, disturbance signal and measurement noise interference studies by injecting signals similar to those arising in reality. From the MATLAB library, the System Identification, Polynomial and Signal Processing Toolboxes were used.

4. METHODS

4.1. Getting in and out of the apparatus

Before experiments in the Wobbler, the paraplegic subjects usually have a period of physiotherapy to reduce their spasticity. This is done on a low bed. 50 mm diameter electrodes (Axelgaard Manufacturing Company: Pals Plus) are attached over the midline of gastrocnemius-soleus with

centres about 100 mm apart. The subject, while lying down, then rolls on his side while his brace is brought up behind him. He then rolls back into the brace and the straps are fastened. A wheeled tilt bed is then brought alongside and the subject is lifted onto this bed. The tilt bed is pushed to the Wobbler, the subject aligned with the foot boxes, and his Trainers, with their metal plates, are laced to his feet. Tilting the bed then begins. When properly positioned longitudinally, the brakes on the bed are locked and, for extra security, a rope is tied between the bed and the frame of the Wobbler. As the inclination increases, the subject is helped to guide his feet into the boxes, as he slides down the bed. He can control this descent using handles at either side of the bed. When the feet are in place, the shoe plates are bolted into the boxes.

While the feet have been positioned, the clutch has held the foot boxes toe-up to meet the feet. When the subject is ready, the clutch is disengaged and the subject can then pull himself upright, using a rope from handles on the ceiling (*Figure 4*). When he is upright, he transfers his hands to these handles, to steady himself while the experimenters attach the rope stays.

The feet must next be lifted to a suitable position and the clutch re-engaged. At this time there is a lock on the drive to the crank so that the main shaft is fixed. Next the rotation stops are set (see

Figure 4 The subject lying on a tilt bed, with his shoes in the foot-boxes, ready to pull himself upright.

section 2.4), bearing in mind the future rotation of the ankle when the apparatus is put into motion. The physiotherapist can rotate the boxes, and feel the available range of motion of the ankles. She has, before her, meters which display the two ankle moments. She fixes the rotation stops to protect the ankles before engaging the clutch at a suitable position, which is assessed by the force needed to lift the boxes from the highly plantarflexed position they assumed when the subject pulled himself upright.

Getting out is the reverse of getting in: after the stays are released and the clutch disengaged the subject lowers himself back onto the tilt bed; the shoe plate bolts are removed; he lifts himself so his feet come out of the foot boxes; and the tilt bed is lowered to a comfortable attitude before the Trainers are removed.

4.2. Finding the moment of inertia

This measurement is made occasionally at the end of a session in the Wobbler. The subject is lowered, while still in his brace, and with the clutch disengaged, until he is horizontal. He is then suspended by a strap round his chest from a 1-m-long coil spring. The foot boxes are fixed. He is then "bounced", with a small amplitude. From the frequency, measured with a stopwatch over 20 cycles, the position of the strap and the spring constant, the angular moment of inertia about the ankles can be calculated (*Figure 5*). After taking the measurement, he is pulled upright with a block-and-tackle, attached to another chest strap. The usual procedure for getting out of the apparatus is then followed.

The moment of inertia about the ankles is calculated from the formula where p is the period (s), r is the distance to strap from the ankles (m), and k is the spring constant (N/m). Our spring was wound specially (Oswald Springs Ltd, Redditch, Worcestershire, UK) and has a spring constant of 1530 N/m.

$$I = \left(\frac{pr}{2\pi}\right)^2 . k$$

Our paraplegic subject has a moment of inertia of 95 kg.m^2.

4.3. Finding the centre of mass and the weight

With the subject in the brace, before or after a session in the Wobbler, he is lifted, while remaining horizontal, so that his weight is borne by four bathroom scales. He lies on two transverse wooden bars, arranged approximately with one under his shoulders and one under his knees. At the end of each bar is a scale. From the positions of the scales and the scale-readings, we calculate his weight and the position of his centre of gravity.

5. DISCUSSION

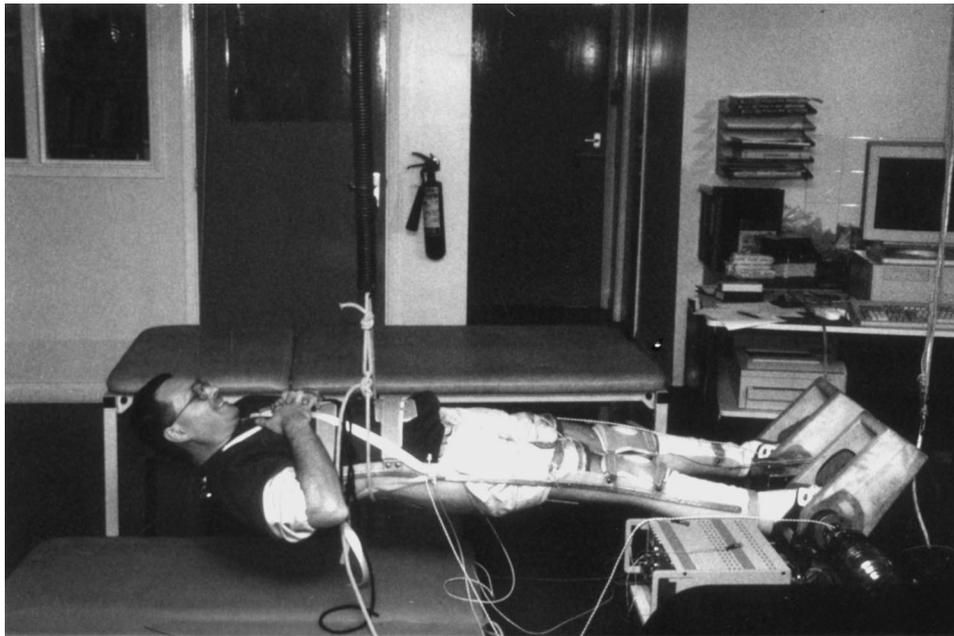When measuring joint stiffness, a displacement is imposed by the Wobbler; from the torque in the

Figure 5 Measuring moment of inertia.

shaft, one wants to estimate the elastic and viscous components of the joints' stiffnesses. The situation is shown in *Figure 6*. The inertia of the feet (M_{LF}, M_{RF}) and the inertia of the foot boxes, shoes, shoe plates and some shaft (which is between the torque load cells and the footboxes) (M_{LFB}, M_{RFB}), is in parallel to the stiffness one wishes to measure (k_{LA}, k_{RA} and the two viscosities). k_{LS} and k_{RS} represent the torsional stiffnesses of the shaft. Assuming that the inertia of the footboxes are equal, the measured values of these components are:

M_{LFB} 0.058 kg.m^2

M_{RFB} 0.058 kg.m^2

M_{LF}[20]   0.01 kg.m^2

M_{RF} 0.01 kg.m^2

k_{LA} ~ 2.5 Nm/deg

k_{RA} ~ 2.5 Nm/deg

k_{LS} 79 Nm/deg

Figure 6 Diagram for understanding the dynamics of the apparatus, showing the drive shaft to the footboxes, the masses of the feet, and the springs and dashpots representing the plantarflexors.

k_{RS} 161 Nm/deg

The apparatus alone is two stiff springs and two masses with seismic excitation. The eigenvalues of this system correspond to a high frequency mode, with the footboxes vibrating in antiphase, and a mode at 29.5 Hz, with the masses moving in phase. This is still a high frequency compared with the frequencies of measurement. The following paragraph considers the dynamics at lower frequencies than this resonance.

When in use at low frequencies, the torque transducers (which are part of the springs k_{LS} and k_{RS}) must measure the stiffnesses k_{LA} and k_{RA}. At very low frequencies, only these springs will be significant. At higher frequencies, the viscosities will cause some phase shift between the displacent drive and the ankle moments. The masses will create a torque which is in antiphase with the springs' torque, and there will be a frequency at which these cancel. For this calculation, we may assume that the shaft is infinitely stiff, and place the four masses and the two muscle-springs (k_{LA} and k_{RA}) in parallel. The frequency at which the torques cancel is given by $(1/2\pi)\sqrt{(k/m)} \sim 8$ Hz. We can conclude that so long as the frequency is well below 8 Hz, the torques measured by the transducers will be dominated by the elastic and viscous moments due to the ankle muscles. This shows that the inertias of the footboxes are adequately small. We may note in passing that for the normal body, held upright by the plantarflexors, the sway frequencies spectrum will be almost entirely below 1 Hz[24].

Ideally, during imitation and real standing, the stiffness of the footboxes would be infinite. This would mimic standing on a rigid floor which does not deflect due to ground contact forces. Actually the stiffnesses of the footboxes are finite (2.1). We can assess the significance of these figures (79 and 53 Nm/deg for left and right respectively: section
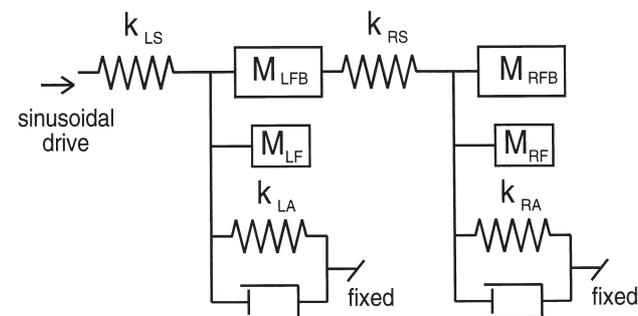
2.1), by considering the stiffness needed to stabilise a rigid body with the same mass distribution as a man. If the centre of gravity is 1 m above the floor and the mass is 70 kg, then to prevent the inverted pendulum falling, the stiffness at the base must exceed 12 Nm/deg (π.m.g.l/180). The control system, which is being tested, responds as a function of time and displacement, but at low frequencies it will approximate a restoring spring. The stiffness of the footboxes is bigger than that of this spring constant, but not much bigger. Most of the compliance of the footboxes is due to the left-hand torque load cell which might be made stiffer. The design of the foot boxes was a compromise to achieve low inertia, as described above, and large stiffness. A similar problem was faced by Robinson *et al.*[25].

Jaeger[26] simulated a controller for quiet standing using the stimulated plantarflexors. With only ankle angle fed back, he demonstrated that, following small disturbances, the inverted pendulum was stable with the parameters he was using. However, if the gain of the muscle is assumed to fall to half its initial value due to muscle fatigue, the system becomes unstable[18]. Such a reduction in gain is very likely to occur. This is a particular example of a general problem in FES; controllers have to be robust because of the changes in the neuromuscular system. One way to improve robustness, in this case, is to nest feedback controllers of the ankle joint moments within the feedback controller of the ankle angle. Design of such a nested controller using optimal methods will be the subject of forthcoming papers. In addition to this advantage, having a measure of the ankle moments, from the torque transducers in the shaft of the Wobbler, enables us to identify the muscles when their length is fixed (Tests B, G and PRBS) and also to test the moment-controllers (Test M) before adding motion to the system. We believe that this gives an incremental procedure which clearly shows what problems occur in practice.

The Wobbler apparatus has been used for many experiments in muscle model identification and feedback control which are described elsewhere[15–17,21,22]. Approval for its use with volunteer paraplegics was granted by the ethical committee at the Royal National Orthopaedic Hospital Trust (Stanmore). Three paraplegics have been tested in the apparatus and the procedures have been satisfactory, although the subjects have expressed amusement at what we ask of them.

Perhaps it is prudent to end with a note of caution. The transducers we are using to measure the ankle moments and the ankle positions are of high accuracy and resolution (0.1 Nm, 0.018°) and low noise. In our experiments with the Wobbler, we are interested in whether standing the paraplegics by feedback control is possible when these feedback signals are of high quality. However, to translate such a standing system to a purely clinical setting will, among other changes, require that the sensors are not conventional engineering transducers: they will either be implanted artificial sensors, the body's natural sensors or sensors worn on the body or part of the footwear. At present, none of these types of sensor are likely to be as accurate and, especially so low noise as the transducers on the Wobbler, so that satisfactory results from the Wobbler should not be taken as an indication that ankle-standing will be possible outside the laboratory.

6. CONCLUSIONS

The Wobbler apparatus was designed to allow feedback control of the plantarflexors to be studied in paraplegics. In this paper the hardware and software are described; also the methods used to measure the biomechanical variables of the inverted pendulum which represents the braced legs and trunk.

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