

Artificial-reflex stimulation for FES-induced standing with minimum quadriceps force

A. J. Mulder H. B. K. Boom H. J. Hermens G. Zilvold

University of Twente, PO Box 217, 7500 AE Enschede, The Netherlands and
Roessingh Rehabilitation Centre, Department of Research & Clinical Engineering,
PO Box 310, 7500 AH Enschede, The Netherlands

Abstract—A control strategy is proposed to decrease quadriceps activity during standing. Modified on/off (or artificial reflex) control is used: a non-numerical or finite-state control scheme based on feedback of knee angle and angular velocity. The control strategy is evaluated in paraplegic patients in an experimental setup using transcutaneous stimulation. The stability of the system and its sensitivity to various control parameters are determined. It is concluded that the control scheme will enable reduction of muscle force independent of additional mechanical bracing or specific posture, and may result in continuous dynamic activation of muscle.

Keywords—Artificial reflex control, Functional electrical stimulation, Paraplegia, Stabilisation of knee joint, Standing

Med. & Biol. Eng. & Comput., 1990, 28, 483-488

1 Introduction

FUNCTIONAL ELECTRICAL stimulation (FES) systems for restoring locomotion in paraplegic patients still have many basic problems preventing their everyday use (SOLOMONOW, 1984a). In general the complexity of inducing motions by FES, whether in upper or lower extremities, requires control strategies at levels ranging from muscle force regulation (CRAGO *et al.*, 1980) and joint position control (ALLIN and INBAR, 1986) up to the co-ordination of synergistic and multi-joint muscle contractions (ALLIN and INBAR, 1986; SOLOMONOW, 1984b).

A typical example of low-level control is the control of knee angle during standing: guaranteeing reliable locking of the knee (during quiet standing or in the stand-phase during walking), without rapidly inducing local muscle fatigue. Current clinical FES systems are based mainly on open-loop control of stimulation (PECKHAM, 1987; JAEGER *et al.*, 1989). Locking of the knee is realised by continuous supramaximal stimulation of knee extensor muscles. Although providing safe knee locking this method easily results in ischaemia due to the maintained fused contraction (MERTON, 1954; MORTIMER *et al.*, 1971) and therefore induces early quadriceps fatigue and limited standtime. Minimisation of quadriceps activation through feedback of kinematical information from the lower limb to the stimulation equipment potentially reduces these problems.

Closed-loop feedback control has been studied extensively to achieve and maintain a desired joint angle or

muscle force (STANIC and TRNKOCZY, 1974; PETROFSKY and PHILLIPS, 1979; CRAGO, 1983; JAEGER, 1986; WILHERE *et al.*, 1985). However, for application in the control of standing (especially when using surface stimulation) poor recruitment stability of the stimulated quadriceps muscle is a problem (TRNKOCZY, 1974). Besides, the mechanical lock state in the knee joint together with the absence of direct muscle force information hinders a proper control of contraction (see also Section 4: Discussion). PETROFSKY *et al.* (1984) report on one of the few studies concerning closed-loop multijoint control for standing and walking in paraplegics. The system was tested in paraplegic subjects and angle sensor stability is reported to be crucial.

In the current study a control strategy is proposed to achieve decreased quadriceps force during standing, using non-numerical or finite-state control. Modified on/off (or ramp-down) control is used based on feedback of knee angle and angular velocity. A procedure for repeated calibration of the locked position is incorporated. The control strategy was applied in paraplegic patients in an experimental setup, in which external parameters (leg load and movement) could be kept constant. The goal of the experiments was to explore the control scheme for its ability in stabilising the knee joint by investigating the systems dynamics and its sensitivity to important control parameters.

2 Methods

2.1 Control scheme

Fig. 1 shows the state diagram of the knee joint used in the control algorithm. Two physical states are considered: knee-lock and knee-unlock.

Correspondence should be addressed to Professor Mulder at the Roessingh Rehabilitation Centre.

First received 2nd May and in final form 30th October 1989

© IFMBE: 1990

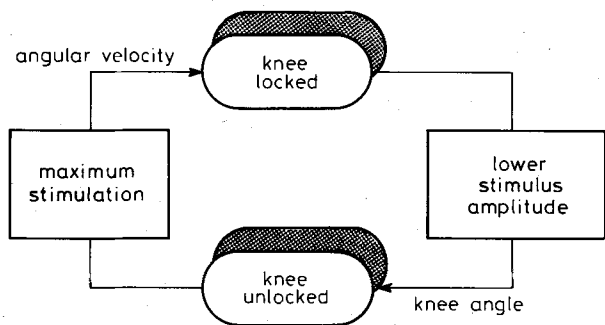


Fig. 1 State diagram of the control scheme

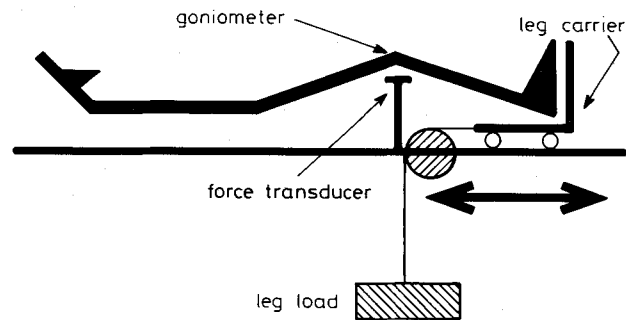


Fig. 2 The stand simulator. The patient is in supine position with his feet on a moveable carrier

2.1.1 *The knee-lock state:* The lock state is defined as the individually and mechanically determined joint end point of the knee. This includes hyperextended positions or end points determined by protecting braces. When the knee joint is locked muscle force cannot be controlled using feedback of knee angle information. Therefore muscle activation is minimised by controlling stimulus pulse amplitude open loop, with the interpulse interval set at a fixed maximum to avoid fatigue. Amplitude and force are supposed to have positive correlation. Linear ramp-down of pulse amplitude (with time-derivative $R \text{ mA s}^{-1}$) is used to lower muscle activation. As a result, the muscle force will drop and the knee joint may finally bend. The dynamics of this bending process depend on external factors: the degree of (hyper-) extension in the knee joint, extending or flexing knee moments resulting from the upper body, passive elasticity and damping of hamstring or iliopsoas muscles, etc. We call this bending 'breakdown'. The stimulation amplitude at which it occurs is the breakdown amplitude. Ramp-down always starts at a defined step level (increment step $S \text{ mA}$) above the previous breakdown amplitude. In this way the average stimulus amplitude adapts to changing breakdown amplitudes due to load and position changes of the leg or to muscle fatigue.

2.1.2 *The knee-unlock state:* When the knee joint comes out of the locked position, further bending must be stopped and the knee joint must be returned to the locked position. From different possible strategies, we have chosen to simply switch stimulation to a high level (MULDER *et al.* 1988).

2.1.3 *State transitions:* An accurate detection of transitions from lock to unlock and vice versa is important for good controller performance. To signal the transition from lock to unlock we use knee angle data obtained from a goniometer. Knee flexion over a defined minimum (the angular deadband A°) will switch stimulation on.

To determine the transition from unlock to lock, angular velocity data are used. Using knee angle data would make the system highly sensitive to goniometer drift. Additionally, using angular velocity data allows automatic calibration of the locked position after each unlock. Lock will be indicated when angular velocity equals zero for a minimum time interval ($T \text{ ms}$) to eliminate errors due to sign transition. A velocity deadband ($V^\circ \text{ s}^{-1}$) is used to eliminate possible velocity noise.

2.2 Experimental set up

2.2.1 *The stand simulator:* To test the control strategy in paraplegic patients under reproducible conditions, we devised a special stand simulator (Fig. 2). The patient is in the supine position on a bench with one foot on a horizon-

tally moveable carrier. A mass under the bench provides an artificial inertial leg load. Leg load and movements are reproducible in this way. It can be shown that for small movements the dynamics of this system equal that during real standing. A force transducer under the knee joint both serves as an artificial kneelock (to obtain an overall flexing knee moment) and measures the locking force for evaluation. The force transducer consists of a bridge configuration of four strain gauges on a thin steel plate connected to a bridge amplifier.

2.2.2 *Subjects:* The experiments were carried out on two complete T5-T6 level spinal cord injured patients. A surface cathode and anode (carbon rubber, $4 \times 7 \text{ cm}$) placed over the rectus femoris motor point and near the patella transcutaneously stimulated the right leg quadriceps muscle. The patients had normal excitability of quadriceps muscles and were well trained during at least a 6 month FES exercise programme. The programme consisted of weight lifting twice a day (30 min, ankle load up to 5 kg) and low-load exercise cycling twice a week (30–60 min). Owing to interpatient differences in training performance and amount of spasticity the muscle contractile properties may differ between subjects.

2.2.3 *Control of stimulation:* To control stimulation we used an IBM-XT computer with AD facilities (Analog Devices, 12-bit), an electrogoniometer to measure knee angle (MCB pp27c, 310° , nonlinearity 1 per cent), and a high output impedance current stimulator (monophasic, rectangular current pulses). The pulse parameters could be controlled by the computer. Pulse duration was fixed at $300 \mu\text{s}$, pulse rate was fixed at the minimum frequency for a fused contraction: 20 Hz. Pulse amplitude was used as control parameter. The bandwidth of the gonio-signal was found to be 10 Hz. To determine the knee angle, the gonio-signal was sampled at 100 Hz and low-pass filtered with a digital first-order filter at 15 Hz. Angular velocity was calculated digitally from the knee angle intersample difference and smoothed by a digital third-order low-pass filter at 15 Hz.

2.2.4 *The experiments:* During the experiments knee angle was locked at 30° . This was chosen to create a high joint load situation to evaluate system dynamics under what were considered the 'worst-case' circumstances to occur during real standing. The external leg load (20 kg) equalled half the trunkload of an average subject. One experimental session lasted about 60 min and consisted of a series of about 10 stimulation experiments of 30 s each. Each session started with a short warming-up of the quadriceps muscle. During this phase the maximum stimulation level, defined as the value sufficient to stretch the leg against the external leg load, was determined. Each stimulation experiment was followed by at least 5 min rest.

During the experiments knee angle data, angular velocity data, stimulation amplitude and locking force data were recorded and stored on disk for offline evaluation.

The most important parameters expected to affect the performance of the controller were: negative slope of the stimulus amplitude ramp-down ($R \text{ mA s}^{-1}$), minimum time interval after which the knee joint is declared stable after switch-on of the stimulation ($T \text{ ms}$), amplitude increment step ($S \text{ mA}$), angular deadband (A°) and angular velocity deadband ($V^\circ \text{ s}^{-1}$). To evaluate the controller, the influence of these parameters on knee angle stability and extending quadriceps force during knee-lock, as well as the response of the knee angle to a positive and negative step change in the leg load, were determined.

Therefore average maximum knee flexion during breakdown (KF), average maximum locking velocity of the knee joint (LV) and average knee extension force (EF) were calculated from the measurements. This was done after a session was finished. KF and LV were calculated over the full 30 s of an experiment. EF was calculated over a 3–10 s interval to eliminate artefacts from start-up or load changes.

3 Results

Fig. 3 shows a typical result of a stimulation experiment. The knee joint continuously switches between the lock and unlock state. After each unlock stimulation is switched on, and after the locked position is regained it is recalibrated and stimulation ramps down again. At $t = 11 \text{ s}$ the leg load was increased with 5 kg. The extra load was removed at

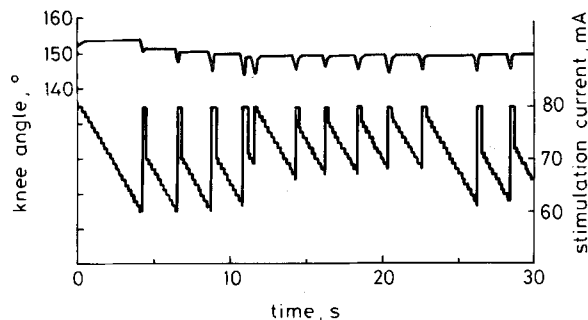


Fig. 3 Typical example of a stimulation experiment. Upper curve: knee angle; lower curve: stimulus amplitude. Essential control parameters: $R = 5 \text{ mA s}^{-1}$, $T = 200 \text{ ms}$, $S = 10 \text{ mA}$, $A = 1.8^\circ$, $V = 13.2^\circ \text{ s}^{-1}$

$t = 24 \text{ s}$. The stimulator responds by instantaneously terminating ramp-down and switching to full stimulation level. After that, average stimulation is at an increased level. The first breakdown at 4 s is due to goniometer disturbances. The locked position is recalibrated and the system remains stable.

3.1 Knee angle

Fig. 4 shows in more detail one stimulus cycle for each of the two patients studied. After stimulation is switched to a high level flexing is slowed down. Then the knee returns to the locked position. The new constant knee angle does not necessarily equal the angle before bending (Fig. 4b). This may be due to movements of the goniometer frame, the hip joint not completely being fixed, or friction in the mechanical parts of the stand simulator. This emphasises the need for repeated calibration of the locked position.

During the experiments average knee flexion during one stimulation cycle depends on the parameter setting of the controller and the mechanical parameters of the leg. Figs. 4a and 4b show characteristic angle responses for the two patients at equal controller setting. Patient 2 exhibits more

knee bending than patient 1. Obviously, mechanical or neural response including muscle dynamics are significantly different for both patients. During the tests the average maximum breakdown knee flexion was 3.6° (standard deviation 1.6°) for patient 1 and 4.8° (standard deviation 1.2°) for patient 2. The time for one stimulus cycle (knee flexion-extension) ranged from 200 to 400 ms.

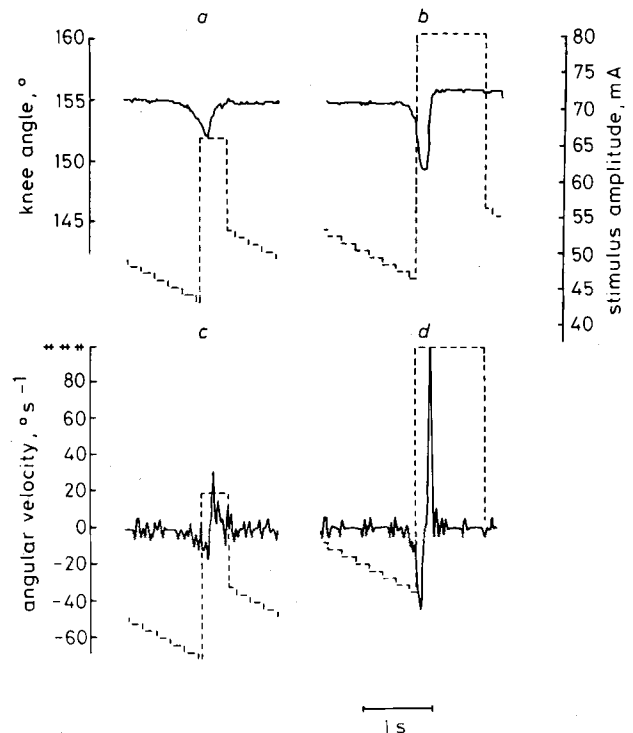


Fig. 4 One detailed stimulus cycle for two patients. Solid line: knee angle (a, b) and angular velocity (c, d) for patient 1 (a, c) and patient 2 (b, d). Dotted: stimulus amplitude. R , S , A and V : see Fig. 3, $T = 400 \text{ ms}$ (a, c) or $T = 1 \text{ s}$ (b, d)

Angular deadband A must be adjusted to cover the remaining signal noise due to insufficient filtering of electrical or mechanical disturbances. If this is not performed, repeated false detection of knee flexion will occur and stimulation will be continuously at high level. On the other hand, angular deadband must be minimised to minimise average knee angle flexion. During the experiments the remaining angle noise was 0.5° peak-to-peak.

3.2 Angular velocity

During the experiments the average locking velocity was 42° s^{-1} (standard deviation 15° s^{-1}) for patient 1 and 54° s^{-1} (standard deviation 15° s^{-1}) for patient 2. Differences between patients may be caused by different muscle force levels, or muscle dynamics (Figs. 4c and 4d).

To achieve accurate detection of knee lock both adjustment of angular velocity deadband V and minimum stimulus on-time T is important. During unlock angular velocity may not exceed the level of noise (Fig. 4c). The use of T has therefore proven to be useful to avoid false detection of knee lock due to low unlock velocities. On the other hand T must be minimal to avoid stimulation at high level longer than required for locking. Optimally T equals the maximum time required to reverse the knee movement from flexion to extension. From velocity data it was concluded that values of T under 100 ms make the detection procedure critical.

V must be set to cover the remaining velocity noise after filtering, which was 15° s^{-1} peak-to-peak. Although V can be reduced when the order of the velocity filter is

increased, this is not recommendable. It would increase the time delay between angle and velocity data as well, which would increase the minimum T which is required, thus deteriorating the dynamic detection of knee lock.

3.3 Stimulus amplitude

The maximum stimulus amplitude was found to be not critical with respect to stabilising the knee joint, although it determines the ultimate level of leg load or allowable muscle force decrease.

When the leg load increases or the muscle force decreases the amplitude increment step S increases the average level of stimulation. When the load increase is high, repeated switching of the stimulation (as demonstrated in Fig. 3) may occur until the mean stimulus level is sufficiently increased. The increment step size S is determined by the desired maximum number of stimulus cycles which are needed to increase the muscle force from minimum to maximum.

For a given leg load, the average interval between 'switch-on stimulation' events is dependent on the stimulus increment step S , the stimulus amplitude ramp-down R and the time needed to stabilise the knee after break-down. This time is dependent on the dynamics of leg and muscles and was found to be 200–400 ms during the experiments. This means that R and S determine the contraction/relaxation ratio of the quadriceps muscle. Besides, R and S determine the response of the muscle to changes in leg load. Therefore, optimum values depend on both the requirements for blood support to the muscle and the patient's individual dynamic load characteristics during actual standing.

During the experiment of Fig. 5 early fatigue of the quadriceps was forced using a stimulus frequency of 60 Hz, and a stimulus on-time $T = 1$ s. The average stimulus level is quickly adapted.

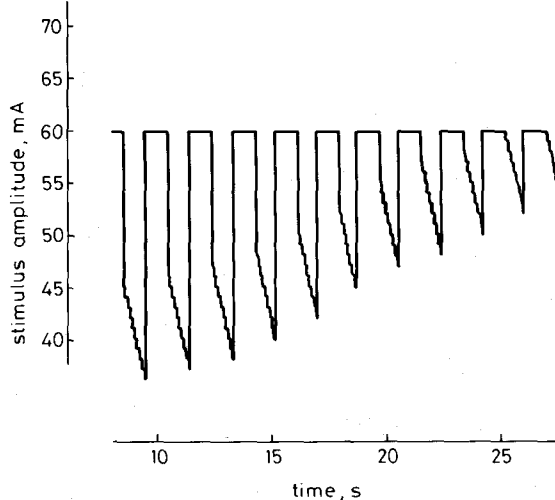


Fig. 5. Response of stimulus amplitude to forced fatigue. Refer to the text for details

The dynamic detection of knee lock from angular velocity data works satisfactorily when the minimum stimulus on-time T does not exceed the 200–400 ms needed to stabilise the knee after breakdown (Fig. 6). When T exceeds this value actual stimulus on-time equals T in most cases, so muscle activation is not optimal.

3.4 Locking force

From several experiments nonlinear and nonstationary recruitment could be seen. Fig. 7a shows inpatient variability in muscle gain during linear amplitude ramp-

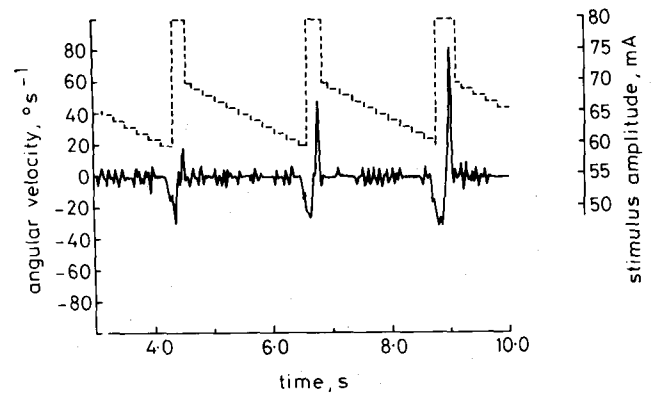


Fig. 6 Dynamic stimulus on-time (dotted line) resulting from lock detection by angular velocity (solid line)

down. An explanation may be the movement of the muscle under the electrode. Fig. 7b shows how both subjects exhibit different force build-up after switch-on of stimulation. This may be due to interpatient differences in muscle and leg responses. Obviously during the recordings of Fig. 7b the minimum on-time T was high with respect to minimising muscle activity: the locking force stabilises at a constant level.

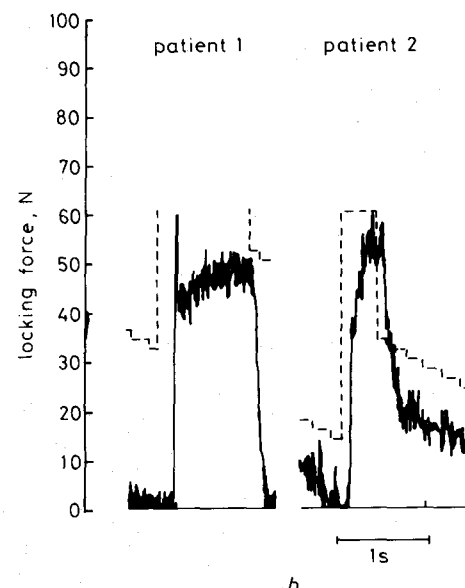
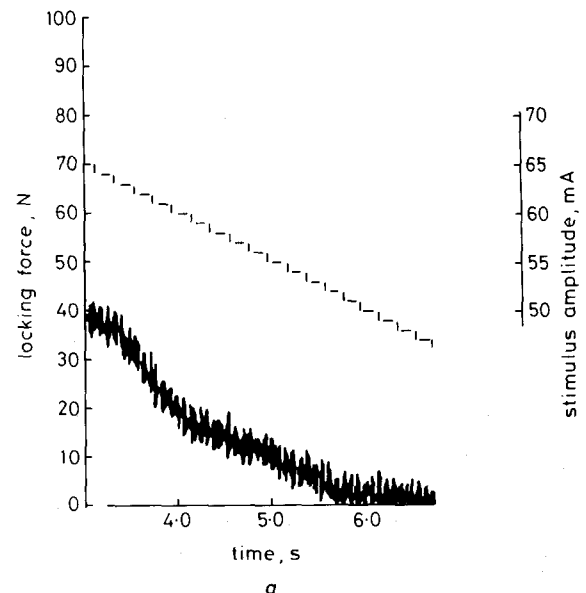


Fig. 7 Locking force for evaluation of muscle response and system behaviour. (a) Nonlinear recruitment; (b) different force build-up for both patients

3.5 Sensitivity

After the exploratory sessions, three sessions of 10 experiments were performed to evaluate the influence of the control parameters in more detail. Specifically, the sensitivity of the system's output parameters KF , LV and EF to changes in the control parameters was investigated.

The experiments were matched to obtain intraindividual sets of experiments performed under equal environmental conditions. Within one set only one control parameter differed (Table 1). The output parameters KF , LV and EF were calculated for each experiment. Using the simple sign test ($p = 0.5$), significance of increase or decrease with respect to changes in the control parameters was determined. The results are shown in Table 2. The effect of S could not be determined due to an insufficient number of valid experiments.

Table 1 Parameters used in the sensitivity analysis. Refer to the text for details

Parameter	Test values	Number of matched experiments
A , °	0.9, 1.8, 3.5	2, 2, 3, 3
R , mA s ⁻¹	5, 10, 20, 40	2, 2, 3, 4
S , mA	5, 10, 20	3
V , ° s ⁻¹	7.8, 13.2	2, 2, 2, 2
T , ms	100, 200, 400 600, 1000	2, 3, 3, 4, 4

Table 2 Correlation between control and output parameters. Correlation can be positive (+), negative (-) or non-significant (0). (α : level of significance)

Parameter	KF	LV	EF
A , °	+ ($\alpha = 0.05$)	+ ($\alpha = 0.125$)	0
R , mA s ⁻¹	+ ($\alpha = 0.01$)	0	- ($\alpha = 0.05$)
V , ° s ⁻¹	0	0	0
T , ms	0	0	+ ($\alpha = 0.05$)

When controller performance is defined to be optimal for minimum KF , LV and EF , Table 2 shows that A and T must be minimised (positive correlation). R shows both positive (KF) and negative (EF) correlation. Therefore its optimal value depends on the relative weight given to KF and EF . Besides optimal R depends on the required system response to dynamic load changes as stated in Section 3.3.

4 Discussion

Important in paraplegic standing and walking is counteracting gravity without undue stressing of the upper body. To improve on the effect of walking with long leg braces stabilising the knee joints additionally requires stabilisation of the hip (KRALJ *et al.*, 1980; ISAKOV *et al.*, 1986). For keeping balance crutches or a walking frame may be necessary.

Traditionally paraplegic standing requires mechanical bracing (ROSMAN and SPIRA, 1974). In conjunction with FES bracing may still be useful, especially in stabilising the hip (PETROFSKY *et al.*, 1985; MCCLELLAND *et al.*, 1987) as well as for the protection of joints. Considering knee stabilisation, besides being a passive solution mechanical bracing usually leads to walking with locked knees requiring increased arm push-up during swing phase. Knee stabilisation by FES allows walking with unlocked knees giving more natural gait and limiting upper body loading.

ANDREWS *et al.* (1989) apply finite state control in a so-called hybrid orthosis, combining low-leg bracing and stimulation of quadriceps to lock the knee. Stimulation is on or off depending on force signals derived from the

brace. The system shows similarity to the work of KRALJ *et al.* (1986) who describe a system which switches between quadriceps stimulation and soleus stimulation to lock the knee. This requires feedback of knee and ankle angle. Both methods offer the opportunity to release the quadriceps temporarily by taking a specific posture. The method proposed in the present study basically has the advantage to reduce average quadriceps force even if the knee is not hyperextended or otherwise locked from ground reaction force. It is therefore independent of mechanical bracing or posture. However, when needed (e.g. in patients with hypermobile knee joints) external bracing may be added to protect the knee joint or prevent it from hyperextending, without affecting the control strategy.

From a model study previously carried out (MULDER *et al.*, 1987; 1988) we concluded that theoretically a proportional derivative (PD) controller is able to stabilise the knee joint during standing. However, it was concluded that the gain robustness of continuous (linear) feedback control is insufficient to counteract the poor recruitment stability of the quadriceps muscle when using surface stimulation (TRNKOCZY, 1974). Although adaptive techniques may improve the performance of such a system (BERNOTAS *et al.*, 1987), its application is still limited to slowly changing parameters. Besides, continuous feedback would be highly sensitive to setpoint drifting as this setpoint coincides with the joint end point of the knee, being a highly nonlinear system element. The present approach incorporating finite state control has been shown to be robust with respect to these problems and may be useful when reliability of system response is more important than accuracy of (position) control. It therefore is considered appropriate for joint locking during standing.

The presence of a limit cycle (i.e. a system oscillation resulting from discontinuity in the control function) enables repeated calibration of the locked position from angular velocity data. This makes the requirements for stability of angle measurement low. Our results indicate that even large goniometer disturbances do not deteriorate overall system stability, i.e. the dynamic system response is not affected, the knee joint does not collapse and muscle force continues to be minimised.

During the experiments average breakdown knee flexion rarely exceeded 8°, which for an average person would correspond to a vertical hip movement of 2 mm during standing. In real standing the load situation may be more favourable than the experimental situation. Therefore generated knee jerks are expected to be within an acceptable range of motion, and the prospects for clinical applicability are good.

In the present study system dynamics were evaluated under high load conditions compared with during standing. However, the proposed stimulation strategy clearly reduces the average muscle force and offers a dynamic muscle activation which is expected to improve muscle blood flow over using high level stimulation continuously. Although in real standing the relative force reduction will depend on the individual knee load situation, it will be better than the experimental situation and fatigue is expected to be reduced significantly (MARSOLAIS *et al.*, 1988). First results with the artificial reflex controller being used in real paraplegic standing have shown a standtime increase of 3.5 times compared with constant supra-maximal stimulation (MULDER *et al.*, 1989). A special situation occurs during longer periods of no flexing knee moment: the level of stimulation will decrease to zero, and no muscle fatigue will be induced. The stimulator then acts as a safety device preventing the knee from inadvertently buckling. This 'no flexing' situation will certainly occur e.g.

in patients who use hyperextension of the knee joint for standing.

For clinical application two controllers can be used in parallel to stabilise both knee joints independently. This offers to the patient the opportunity of transferring weight to one leg to release the other, like healthy subjects during standing.

Acknowledgments—Our work on FES is supported by the Dutch foundations: Stichting Technische Wetenschappen, IOP-HG and St. Joris Stichting.

References

- ALLIN, J. and INBAR, G. F. (1986) FNS control schemes for the upper limb. *IEEE Trans.*, **BME-33**, 818–828.
- ANDREWS, B. J. *et al.* (1989) Rule-based control of a hybrid FES orthosis for assisting locomotion. *Automedica*, **11**, 175–199.
- BERNOTAS, L. A., CRAGO, P. E. and CHIZECK, H. J. (1987) Adaptive control of electrically stimulated muscle. *IEEE Trans.*, **BME-34**, 140–147.
- CRAGO, P. E., MORTIMER, J. T. and PECKHAM, P. H. (1980) Closed loop control of muscle force during electrical stimulation of muscle. *Ibid.*, **27**, 306–311.
- CRAGO, P. E. (1983) Control of movements by functional neuromuscular stimulation. *Eng. in Med. & Biol. Mag.*, **1**, 32–36.
- ISAKOV, E., MIZRAHI, J. and NAJENSON, T. (1986) Biomechanical and physiological evaluation of FES-activated paraplegic patients. *J. Rehab. Res. & Dev.*, **23**, 9–19.
- JAEGER, R. J. (1986) Design and simulation of closed-loop electrical stimulation orthoses for restoration of quiet standing in paraplegia. *J. Biomech.*, **19**, 825–835.
- JAEGER, R. J., YARKONY, G. M. and ROTH, E. J. (1989) Rehabilitation technology for standing and walking after spinal cord injury. *Am. J. Phys. Med. & Rehab.*, **68**, 128–133.
- KRALJ, A., BAJD, T. and TURK, R. (1980) Electrical stimulation providing functional use of paraplegic patient muscles. *Med. Progr. through Technol.*, **7**, 3–9.
- KRALJ, A., BAJD, T., TURK, R. and BENKO, M. (1986) Posture switching for prolonging functional electrical stimulation standing in paraplegic patients. *Paraplegia*, **12**, 221–230.
- MARSOLAIS, E. B., BENNETT, G. and EDWARDS, C. C. T. (1988) Energy cost of walking and standing with functional neuromuscular stimulation and long leg braces. *Arch. Phys. Med. & Rehab.*, **69**, 243–249.
- MCCLELLAND, M., ANDREWS, B. J., PATRICK, J. H., FREEMAN, P. A. and MASRI, W. S. (1987) Augmentation of the Oswestry parawalker aorthosis by means of surface electrical stimulation: gait analyses of three patients. *Paraplegia*, **25**, 32–38.
- MERTON, P. A. (1954) Voluntary strength and fatigue. *J. Physiol.*, **123**, 553–564.
- MORTIMER, J. T., KERSTEIN, M. D. and MAGNUSON, R. (1971) Muscle blood flow in the human biceps as a function of developed muscle force. *Arch. Surg.*, **103**, 376–377.
- MULDER, A. J., VERHEIJEN, J. M. E., HERMENS, H. J. and VAN ALSTE, J. A. (1987) Control of standing using functional electrical stimulation. In *Int. Ser. on Biomech. XI-A*. GROOT, G. de HOLLANDER, A. P., HUIJING, P. A. and VAN INGEN SCHENAU, G. J. (Eds.), Free University Press, Amsterdam, 208–213.
- MULDER, A. J., HERMENS, H. J., VAN ALSTÉ, J. A. and ZILVOLD, G. (1988) An improved strategy to minimize muscle fatigue during FES-induced standing. In *Electrophysiological kinesiology*. WALLINGA, W., BOOM, H. B. K. and DE VRIES, J. (Eds.), Excerpta Medica ICS 804, Elsevier, Amsterdam, New York, 109–112.
- MULDER, A. J., BOOM, H. B. K., HERMENS, H. J. and ZILVOLD, G. (1989) Artificial-reflex stimulation: clinical possibilities in paraplegic standing. Proc. Vienna 3rd Int. Conf. on FES, Baden/Vienna, 47–50.
- PECKHAM, P. H. (1987) Functional electrical stimulation: current status and future prospects of applications to the neuromuscular system in spinal cord injury. *Paraplegia*, **25**, 279–288.
- PETROFSKY, J. S. and PHILLIPS, C. A. (1979) Constant-velocity contractions in skeletal muscle by sequential stimulation of muscle efferents. *Med. & Biol. Eng. & Comput.*, **17**, 583–592.

- PETROFSKY, J. S., PHILLIPS, C. A. and STAFFORD, D. E. (1984) Closed-loop control for restoration of movement in paralyzed muscle. *Orthopedics*, **7**, 1289–1302.
- PETROFSKY, J. S., PHILLIPS, C. A., LARSON, L. and DOUGLAS, R. (1985) Computer synthesized walking: an application of orthosis and functional electrical stimulation (FES). *J. Neur. & Orth. Med. & Surg.*, **6**, 219–230.
- ROSMAN, N. and SPIRA, E. (1974) Paraplegic use of walking braces: survey *Arch. Phys. Med. & Rehab.*, **55**, 310–314.
- SOLOMONOW, M. (1984a) Restoration of movement by electrical stimulation: a contemporary view of the basic problems. *Orthopedics*, **7**, 245–250.
- SOLOMONOW, M. (1984b) External control of the neuromuscular system. *IEEE Trans.*, **BME-31**, 752–763.
- STANIC, U. and TRNKOCZY, A. (1974) Closed-loop positioning of hemiplegic patient's joint by means of functional electrical stimulation. *Ibid.*, **BME-21**, 365–370.
- TRNKOCZY, A. (1974) Variability of electrically evoked muscle contractions with special regard to closed-loop controlled orthosis. *Ann. Biomed. Eng.*, **2**, 226–238.
- WILHERE, G. F., CRAGO, P. E. and CHIZECK, H. J. (1985) Design and evaluation of a digital closed-loop controller for the regulation of muscle force by recruitment modulation. *IEEE Trans.*, **BME-32**, 668–676.

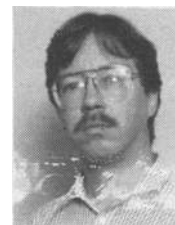
Authors' biographies



Arjan Mulder was born in Nieuwiesden, The Netherlands, in 1958. He received his MS degree in Electrical Engineering from the University of Twente in 1984. He specialised in medical electronics. In 1985 and 1986 he worked at the Research & Clinical Engineering Department of the Roessingh Rehabilitation Centre. Since 1986 he has been working as a part of the rehabilitation engineering collaboration between the University of Twente and the Roessingh Rehabilitation Centre. His main research interests are development and clinical implementation of control systems for functional electrical stimulation and rehabilitation engineering in general.



Herman Boom was trained as a Medical Physicist at the University of Utrecht, The Netherlands, where he received his Ph.D. degree in 1971. He joined the Departments of Medical Physics and Medical Physiology where he was engaged in research in the field of cardiac mechanics and taught Physiology and Biophysics. Since 1976 he has occupied the Chair of Medical Electronics in the Department of Electrical Engineering, University of Twente, The Netherlands. His research interests are cardiovascular systems dynamics, bioelectricity and rehabilitation technology.



Hermie Hermens was born in Nijmegen, The Netherlands in 1953. He received his masters degree in Electrical Engineering from the University of Twente in 1981. Since then he has worked at the Roessingh Rehabilitation Center in Enschede, as head of the Research & Clinical Engineering Department from 1984. His main research is directed towards the development of methods for analysis and simulation of surface EMG. Further interests include the field of functional electrical stimulation.



Gerrit Zilvold was born in Roermond, The Netherlands, in 1939. He received his MD degree in Medicine from the Vrije Universiteit in Amsterdam in 1968. In 1976 he was awarded the D.Sc. degree on the subject of functional electrical stimulation of the n. peroneus. Since 1978 he has been medical director of the Roessingh Rehabilitation Center. In 1983 he was appointed part-time Professor for Rehabilitation Engineering at the University of Twente.