Rehabilitation engineering

On/off control in FES-induced standing up: a model study and experiments

A. J. Mulder P. H. Veltink H. B. K. Boom

Twente University, PO Box 217, 7500 AE Enschede, The Netherlands

Abstract—*Control of paraplegic standing up was studied with respect to the limitation of the end-velocity of the knee joint when the patient reaches the upright position. Closed-loop on/off control of knee extensor muscle was compared with open-loop controlled standing up both in a model study and in paraplegic patients in a controlled model situation. Criteria were knee-end-velocity and knee extensor muscle activation time. Sensitivity of the system to additional arm support and (in the model study) to the dynamics of knee extensor muscle was studied. It is concluded that the control scheme may reduce knee-end velocity to about 40 per cent and knee extensor activation time to near 70 per cent of the respective open-loop values.*

Keywords—*Functional electrical stimulation, On/off control, Paraplegic, Standing up*

Med. & Biol. Eng. & Comput., 1992, **30**, 205–212

1 Introduction

CONTROL OF stimulation is an important aspect in the restoration of paraplegic standing and walking by means of functional electrical stimulation (FES). In literature, different control strategies are described, usually depending on the task to be performed. This involves detection of patient intention (GRAUPE and KOHN, 1988; ANDREWS *et al.*, 1989), maintaining or sequencing desired joint positions (STANIC and TRNKOCZY, 1974; ALLIN and INBAR, 1986; MULDER *et al.*, 1990) and the overall co-ordination of joint movements (POPOVIC *et al.*, 1989; ANDREWS *et al.*, 1989).

One of the tasks to be performed in the application of FES is to guarantee a safe transition from sitting to standing. In literature closed-loop control of standing up is reported by PETROFSKY *et al.* (1984). Fast open-loop ramp-up of knee extensor activation stretches the knee joint to bring the patient to a standing position. Knee angle is measured to detect full leg extension. This strategy shows minor differences compared with the traditional (open-loop) method of FES induced paraplegic standing up: traditionally standing up is established by constant supramaximal stimulation of at least the quadriceps femoris muscles, sometimes in conjunction with the gluteus maximus muscles (KRALJ *et al.*, 1980; PHILLIPS, 1989; PECKHAM, 1987). This may lead to high knee end-velocities because during standing up the knee torque required to counteract gravity decreases from an initial maximum to a low value during stable standing (BAJD *et al.*, 1982). This will especially be the case in patients with hypermobile knee joints. In general high joint end-velocities may raise the risk of joint damage in the long term (see Discussion).

In the current study a control strategy was developed to limit knee end-velocity during FES-induced standing up. In designing the control strategy two considerations were of major importance. First, during a major part of the stand-up movement maximum knee extensor activity is required. BAJD *et al.* (1982) report how the paralysed quadriceps muscle is capable of generating only part of the peak knee joint moment required during standing up. Consequently, assistance of nonparalysed upper-body musculature is required, at least in an initial phase of movement. This may deteriorate controllability of the system from a FES point of view. Secondly, nonstationary muscle behaviour is still considered an important problem in FES (TRNKOCZY, 1974; CRAGO *et al.*, 1980; BERNOTAS *et al.*, 1987). This is especially true in transcutaneous applications, where the motor point of the muscle may move relative to the electrode. It may therefore be advantageous to use only the most reproducible levels of recruitment, which are maximum and no muscle activation. This incorporates the boundary condition of maximum initial knee extensor moment.

In the present study on/off controlled standing up is compared with open-loop controlled standing up in both a model study and paraplegic patients in a controlled model situation. Comparison was performed against two criteria: knee end-velocity and the estimated metabolic energy consumption of the stimulated muscle. In general energy consumption is related to local muscle fatigue, which is considered to be a major factor determining the clinical possibilities of FES systems (PECKHAM, 1987). To estimate metabolic energy consumption (under the condition of fixed stimulus rate and stimulus amplitude) muscle activation time was used (KHANG, 1988).

Correspondence should be addressed to Professor P. H. Veltink, Department of Electrical Engineering, University of Twente, PO Box 217, 7500 AE Enschede, The Netherlands.

Dr A. J. Mulder is now at the Roessingh Rehab Centre, Department of Research & Clinical Engineering, PO Box 310, 7500 AH Enschede, The Netherlands.

Received 9th August 1990

2 Theory

2.1 *Biomechanical model*

The biomechanical model used to describe standing up was based on a movement study of NUZIK *et al.* (1986) concerning standing up of normal subjects and a study of BAJD *et al.* (1982) on both normal and paraplegic standing up. From these studies standing up was concluded to consist of sagittal rotations of trunk and upper leg segments, the lower leg in the first approximation remaining in a vertical position. Trunk and upper leg rotation are in opposite directions, where the absolute trunk rotation equals approximately half the upper leg rotation. Initial knee and hip flexion are $\frac{1}{2}\pi$ and $\frac{1}{4}\pi$ rad, respectively.

Based on this a model was set up consisting of three rigid bodies (trunk, upper and lower leg) interconnected by two one-dimensional ideal joints (Fig. 1). For each segment the centre of gravity is assumed to be positioned halfway

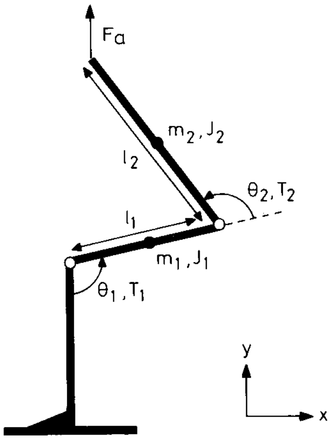


Fig. 1 *The segmental model of standing up*
m_1, J_1 : mass and moment of inertia of upper leg
m_2, J_2 : mass and moment of inertia of trunk
θ_1, T_1 : knee angle and external knee torque
θ_2, T_2 : hip angle and external hip torque
l_1, l_2 : length of upperleg and trunk
F_a : vertical arm force

along the segment. To match the kinematic constraints θ_2 was defined as

$$\theta_2 = -(3/2)\theta_1 + (3/2)\pi$$

In this way knee angle completely determines the system.

To find the dynamic equation of the system the Lagrangian formalism was used (KARNOPP, 1977). The equation is derived in the Appendix. The result is

$$\begin{aligned} T_1 = \ddot{\theta}_1(&-(1/8)m_2 l_2^2 + (1/4)m_1 l_1^2 + m_2 l_1^2 \\ &-(1/4)m_2 l_1 l_2 \sin((3/2)\theta_1) + J_1 - \tfrac{1}{2}J_2) \\ &-(3/8)m_2 l_1 l_2 \cos((3/2)\theta_1)\dot{\theta}_1^2 \\ &+\tfrac{1}{2}m_1 g l_1 \sin(\theta_1) + m_2 g l_1 \sin(\theta_1) \\ &-\tfrac{1}{2}m_2 g l_2 \cos(\tfrac{1}{2}\theta_1) \end{aligned} \qquad (1)$$

where T_1 is the knee torque, θ_1 the knee angle and g is the acceleration of gravitation. The remaining parameters are explained in Fig. 1. Eqn. 1 is the differential equation describing a nonlinear second-order system.

Knee moment T_1 is composed of the two model inputs: knee extensor moment T_e resulting from stimulation according to the muscle dynamics and additional knee moment T_a which is generated from voluntary arm force F_a. Arm force F_a was modelled as a vertically directed constant force input. Horizontal arm force components which in real standing up are necessary for horizontally stabilising the body are not incorporated in the model (see also Discussion). This leads to

$$\begin{aligned} T_1 &= T_e + T_a = T_e + F_a(l_1 \sin(\theta_1) + l_2 \sin(\theta_1 + \theta_2)) \\ &= T_e + F_a(l_1 \sin(\theta_1) - l_2 \cos(\tfrac{1}{2}\theta_1)) \end{aligned} \qquad (2)$$

2.2 *Controller*

To obtain zero knee end-velocity stimulation must be switched off at the right instant during movement. In general, when both system and optimisation criteria would be known an open-loop optimal control strategy could be derived. However, in standing up system response may be unpredictable due to arm force influence and deviations in stand-up movement. This indicates applying closed-loop control, to limit the sensitivity of the system for such disturbances.

For closed-loop on/off control a switching function can be defined (LUENBERGER, 1979; RUBIN, 1986) which divides the state space of a system into regions of on and off actuator command.

When the order of the system is N, and X is the N-dimensional state vector of the system, the switching function can be written as:

$$f(X) = 0 \qquad (3)$$

This requires feedback of N state variables. For standing up the minimum value of N will be three when considering the order of the mechanical system to be two and knee extensor muscle to have at least a first-order nature (see below). State variables then are θ, $\dot{\theta}$ and T_e.

In clinical practice from these only θ and $\dot{\theta}$ can be measured and used for feedback. Therefore control of standing up has to be confined to the use of a two-dimensional projection of the switching function on the state plane (θ, $\dot{\theta}$), which projection is then dependent on the third state variable, i.e. instantaneous knee extensor output.

The present study investigates the possibilities of replacing this projected switching function by a static predetermined switching curve in state space (θ, $\dot{\theta}$). To find this switching curve the following procedure was applied. First knee extensor dynamics was considered to be absent. The corresponding switching curve (see below) is still arm force dependent. Then to compensate for knee extensor dynamics this curve was experimentally corrected to obtain an earlier switching moment.

The switching curve for the situation that knee extensor dynamics is neglected can be found by solving eqns. 1 and 2 for $T_e = 0$, under the initial condition ($\theta = \pi$, $\dot{\theta} = 0$). Because eqn. 1 is invariant for the sign of $\dot{\theta}$, thus the switching curve has an equivalent for inverted $\dot{\theta}$. This describes the situation where the system is free falling from the state ($\theta = \pi$, $\dot{\theta} = 0$).

3 Methods

On/off controlled standing up using a predetermined switching curve in state space (θ, $\dot{\theta}$) was compared with open-loop controlled constantly stimulated standing up both in a model study and experiments on paraplegic patients. The criteria were knee end-velocity KEV (rad s^{-1}) and knee extensor muscle activation time MAT (s). KEV is defined as the angular velocity for $\theta = \pi$ rad, and MAT was used to estimate metabolic energy consumption of knee extensor muscles under the condition of fixed stimulus rate and amplitude.

3.1 *Model simulations*

Model simulations were used to determine the sensitivity of the system to muscle dynamics and (unknown) arm force input F_a (eqn. 2). The systems equations were implemented on an IBM-XT computer. Values for the mechanical parameters of the model were based on regression equations of Chandler *et al.* (1975): $l_1 = l_2 = 0{\cdot}5\,m$, $m_1 = 10{\cdot}0$ kg, $m_2 = 50{\cdot}0$ kg, $J_1 = 0{\cdot}18\,\mathrm{N\,m\,s^2}$, $J_2 = 1{\cdot}4\,\mathrm{N\,m\,s^2}$, where m_1 represents both upper leg segments. Maximum knee extensor moment was set to 75 N m per leg (Bajd *et al.*, 1982). According to eqns. 1 and 2 this requires at least 250 N of vertical arm force F_a to start any upward movement. In simulations a range of 300–500 N was used. Based on literature (Durfee and MacLean, 1989; Carroll *et al.*, 1989; Baratta *et al.*, 1989) and our own findings muscle dynamics was modelled as a linear second-order critically damped system determined by a single time constant t_m (s) corresponding to both system poles. t_m was set to 50 ms nominal value, range 25–100 ms.

3.2 *Experimental setup*

To test the control strategy on paraplegic patients in a controlled model situation, a stand-up apparatus was constructed (Fig. 2). The patient is in the supine position on a bench with one foot on a carrier, moving the leg according to the stand-up trajectory. A mass under the bench and an adjustable compensation weight simulate the forces and inertial load as it results from the trunk during real standing up. During experiments leg loads of 10 and 15 kg were applied.

Experiments were performed on two complete T5–T6 level paraplegic subjects. The right leg quadriceps muscle was transcutaneously stimulated using carbon rubber cathode and anode (4 × 7 cm) placed over the rectus femoris motor point and near the patella. The patients had normal excitability of quadriceps muscles and were well trained during at least a 6 month FES exercise programme. Maximum isometric knee torque ranged from 40 to 50 N m for both patients. Subject 1 was free from spasticity. Subject 2 showed slight flexion and extension spasms and had obstructions near full knee extension.

To control stimulation we used an IBM-XT computer with AD facilities (Analog Devices, 12-bit). Knee-angle was measured using an electrogoniometer (MCB pp 27c, 310°, nonlinearity 1 per cent). For stimulation a digitally controlled high-output impedance current stimulator was applied (monophasic rectangular current pulses, duration 300 μs, pulse rate fixed at the minimum frequency for a fused contraction: 20 Hz). Pulse amplitude was set supramaximal. To determine knee angle (bandwidth 10 Hz), the goniosignal was sampled at 100 Hz and digitally first-order low-pass filtered with a cutoff frequency of 15 Hz. Angular velocity was calculated from the knee angle intersample difference and smoothed by a digital third-order low-pass filter with a cutoff frequency of 15 Hz. The goniometer system was calibrated before each experiment. To avoid the occurrence of a limit cycle, stimulation was on when knee angle was within 10° of extension.

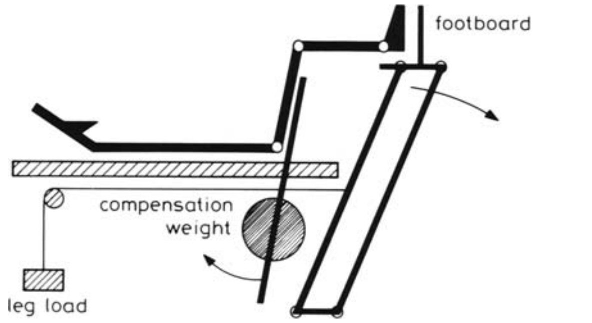


Fig. 2 *Schematic drawing of a paraplegic subject in supine position on the stand-up apparatus*

3.3 *Protocol*

Three series of experiments were performed for both the model simulations and stimulation experiments.

(*a*) Open-loop controlled standing up. Applying continuous activation of knee extensor muscles, *KEV* and *MAT* were determined and used as reference for closed-loop experiments.

(*b*) The stand-to-sit movement for zero knee extensor moment. Being the (mirrored) switching curve for the situation that knee extensor dynamics is absent, this is used as a reference for closed-loop controlled standing up (see Theory). A small initial knee flexion is used to start the movement.

(*c*) Closed-loop on/off controlled standing up. The switching curve found from series (*b*) was corrected to obtain an earlier switching moment to compensate for knee extensor dynamics. This was realised by rotating the switching curve in state space around the target position ($\theta = \pi$, $\dot{\theta} = 0$) (Nuzik *et al.*, 1986). In the model simulations first automatic hill-search was applied to find the optimum rotation of the switching curve with respect to F_a and t_m. Then sensitivity of *KEV* and *MAT* to F_a and t_m was determined when using a predetermined average switching curve.

During stimulation experiments on paraplegics the average switching curve found from the model study was used and the sensitivity of *KEV* and *MAT* for rotation of the curve was studied.

4 Results

4.1 *Model simulations*

4.1.1 *Open-loop stimulation:* Fig. 3 shows the open-loop state-space trajectories of the system for different values of arm force. For $F_a = 0$ the movement was initiated at the knee angle where knee extensor moment starts exceeding negative gravitation moment (2·3 rad). When F_a exceeds the minimum value to start upward movement from sitting (curves 2–5) *KEV* linearly increases with F_a ($r > 0{\cdot}99$) from 3·3 ($F_a = 250$ N) to 4·1 rad s^{-1} ($F_a = 500$ N). For curve

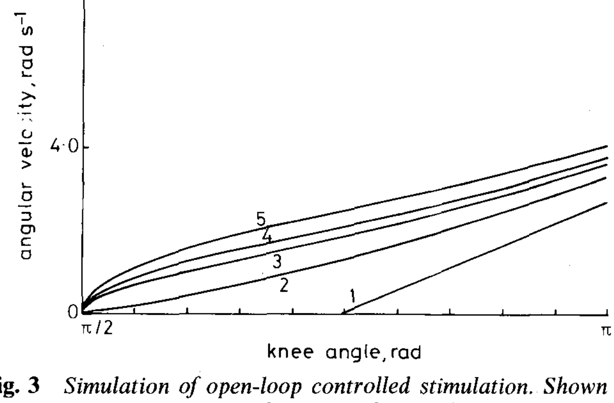


Fig. 3 *Simulation of open-loop controlled stimulation. Shown are state trajectories of knee joint for F_a: (1) 0 N; (2) 250 N; (3) 350 N; (4) 400 N; and (5) 500 N. Curve 1 ($F_a = 0$) results in minimum KEV (2·8 rad s^{-1}). The curve starts at $\theta = 2{\cdot}3$ rad, from which knee extensor moment exceeds gravitation. For the other curves starting angle is $\pi/2$ rad. Respective KEV and MAT are (2) 3·3 rad s^{-1} and 4·3 s; (3) 3·7 rad s^{-1} and 1·5 s; (4) 3·8 rad s^{-1} and 1·3 s; (5) 4·1 rad s^{-1} and 1·0 s*

1 MAT is undefined because the movement is undefined for $0 < \theta < 2{\cdot}3$ rad. MAT decreases with F_a from 4·3 s ($F_a = 250$ N) to 1·0 s ($F_a = 500$ N). So maximum KEV corresponds to minimum MAT.

4.1.2 *The stand-to-sit trajectory:* Fig. 4 shows the stand-to-sit trajectory for knee torque zero and different arm forces. Negating $\dot{\theta}$ gives the switching curves that would result in zero KEV during standing up when knee extensor dynamics is neglected. The trajectories approach linear

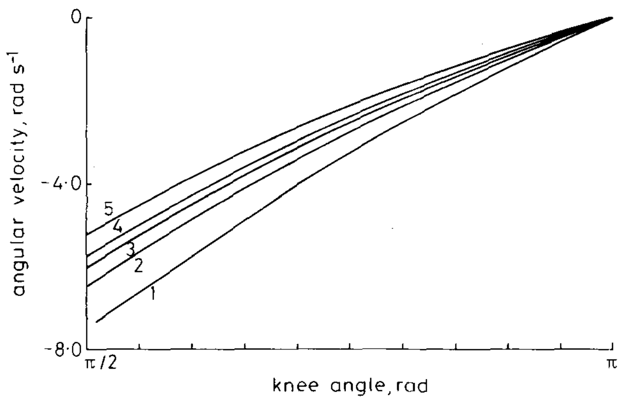


Fig. 4 *Simulation of stand-to-sit movement. Negating $\dot{\theta}$ gives the required switching curves to obtain zero KEV during on/off controlled standing up, when knee extensor dynamics is neglected. Shown are state trajectories of knee joint for F_a: (1) 0 N; (2) 250 N; (3) 350 N; (4) 400 N; and (5) 500 N. To start the movement of 0·01 rad initial flexion is used*

functions ($r > 0{\cdot}99$) although they have slight negative curvature. Clearly the nonlinear character of the biomechanical system results in overall linear phase behaviour as the curvature would have been positive for a linear second-order inertial system. The slope linearly decreases ($r > 0{\cdot}99$) from 4·11 to 3·31 s^{-1} for F_a of 250 to 500 N, respectively (Fig. 5).

According to the near linearity observed in the stand-to-sit trajectories the closed-loop switching curves (next section) were approximated and implemented as the linear function $\dot{\theta} = S(\pi - \theta)$. S is the slope of the linearised switching line, and was subjected to optimisation and sensitivity analysis with respect to KEV and MAT. Thus the applied control algorithm was

$$\underline{\text{IF}}\ ((\theta + \dot{\theta}/S) \geqslant \pi)\ \underline{\text{THEN}}\ (\text{stimulation OFF})$$
$$\underline{\text{ELSE}}\ (\text{stimulation ON}) \qquad (4)$$

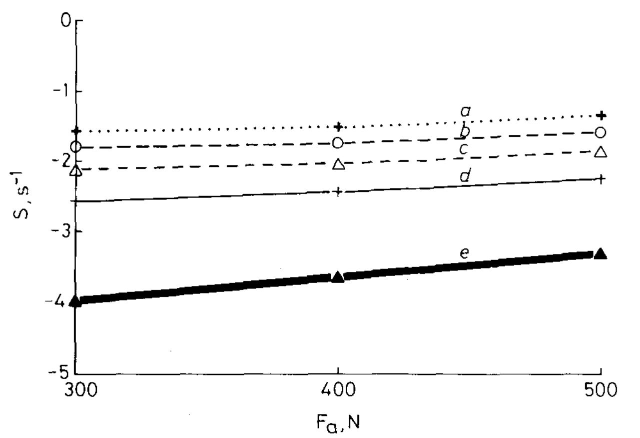


Fig. 5 *(a–d) Slope S which results in zero KEV during closed-loop on/off controlled standing up when using a linear switching curve. (e) As a reference the (negated) slope of the stand-to-sit trajectories of Fig. 4 is indicated. S is plotted against F_a for different values of t_m: (a) t_m = 100 ms; (b) 75 ms; (c) 50 ms; and (d) 25 ms*

4.1.3 *Closed-loop on/off controlled standing up:* During on/off controlled standing up those values of S that resulted in zero KEV were substantially lower then the (negated) slope of the stand-to-sit trajectory, due to the compensation required for knee extensor dynamics (Fig. 5). MAT is sensitive mainly to F_a, whereas sensitivity to t_m is low. Averaged over all test values of t_m, MAT is $1{\cdot}6 \pm 0{\cdot}1$ s ($F_a = 300$ N), $0{\cdot}99 \pm 0{\cdot}07$ s ($F_a = 400$ N) and $0{\cdot}72 \pm 0{\cdot}06$ s ($F_a = 500$ N), which are 84, 76 and 72 per cent of the respective open-loop values.

As an example Fig. 6*a* shows the behaviour of the system during standing up when applying the switching function of eqn. 4 for a fixed value of S and different arm forces. A fixed switching line did not completely compensate for arm force alteration with respect to obtaining zero KEV. For low arm force extra switch-on of stimulation was required. Figs. 6*b*–6*d* show the influence of F_a and t_m on KEV and MAT in more detail.

Figs. 6*b* and 6*c* show the influence of F_a on KEV and MAT, respectively. Fig. 6*b* shows minimum, average and peak KEV over the observed range of F_a when S is adjusted to obtain minimum KEV at $F_a = 500$ N (curve 3). This requires extra switch-on of knee extensor muscle over all F_a except for maximum F_a (compare Fig. 6*a*). Fig. 6*c* shows the influence of F_a on MAT for the controller settings of Fig. 6*b*. The impact of S was minimal. MAT decreased from $1{\cdot}6 \pm 0{\cdot}01$ ($F_a = 300$ N) to $0{\cdot}71 \pm 0{\cdot}01$ s ($F_a = 500$ N). Fig. 6*d* gives the influence of t_m on KEV for different F_a, showing positive correlation and high sensitivity for t_m exceeding the nominal value of 50 ms. The effect of t_m on MAT is not shown and was found to be low. Compared with the values in Fig. 6*c* MAT increases by 23 ± 3 per cent over the total range of F_a when t_m increases from 50 to 100 ms.

4.1.4 *Comparison of KEV and MAT in open- and closed-loop simulations:* When using a fixed switching-line, on/off controlled standing up reduced KEV to 36 per cent or less of the minimum open-loop value over the observed range of F_a (Fig. 3). With S not properly tuned to knee extensor dynamics KEV still was reduced to 54 per cent or less of the open-loop value over the observed range of t_m. Closed-loop control reduced MAT to an average of 77 per cent of the open-loop value for given F_a.

4.2 *Experimental results*

4.2.1 *Subject 1:* Fig. 7*a* shows a typical state trajectory of the knee joint as recorded during open-loop controlled stimulation. Compared with the simulation results of Fig. 3 some conspicuous deviations occurred. Near full knee extension angular velocity was decreasing rapidly despite stimulation being on. This may be due to compliance and/or damping in the knee joint, where also nonlinearities and friction of the stand-up apparatus may be involved. The knee joint appeared to be hyperextended for the patient involved. Based on these observations during experiments KEV was defined as maximum angular velocity during open-loop experiments and angular velocity at the knee angle corresponding to this maximum during closed-loop experiments. Open-loop KEV was $9{\cdot}0 \pm 0{\cdot}7$ rad s^{-1} for the subject involved. This is high compared with the simulation results and may be caused by the relative low inertial load (see Discussion). The average MAT was $0{\cdot}74 \pm 0{\cdot}05$ s.

Fig. 7*b* gives a typical example of the stand-to-sit trajectory for knee torque zero. The irregular movements around full extension may be due to the movement manually being started, whereas a limited range of motion of the

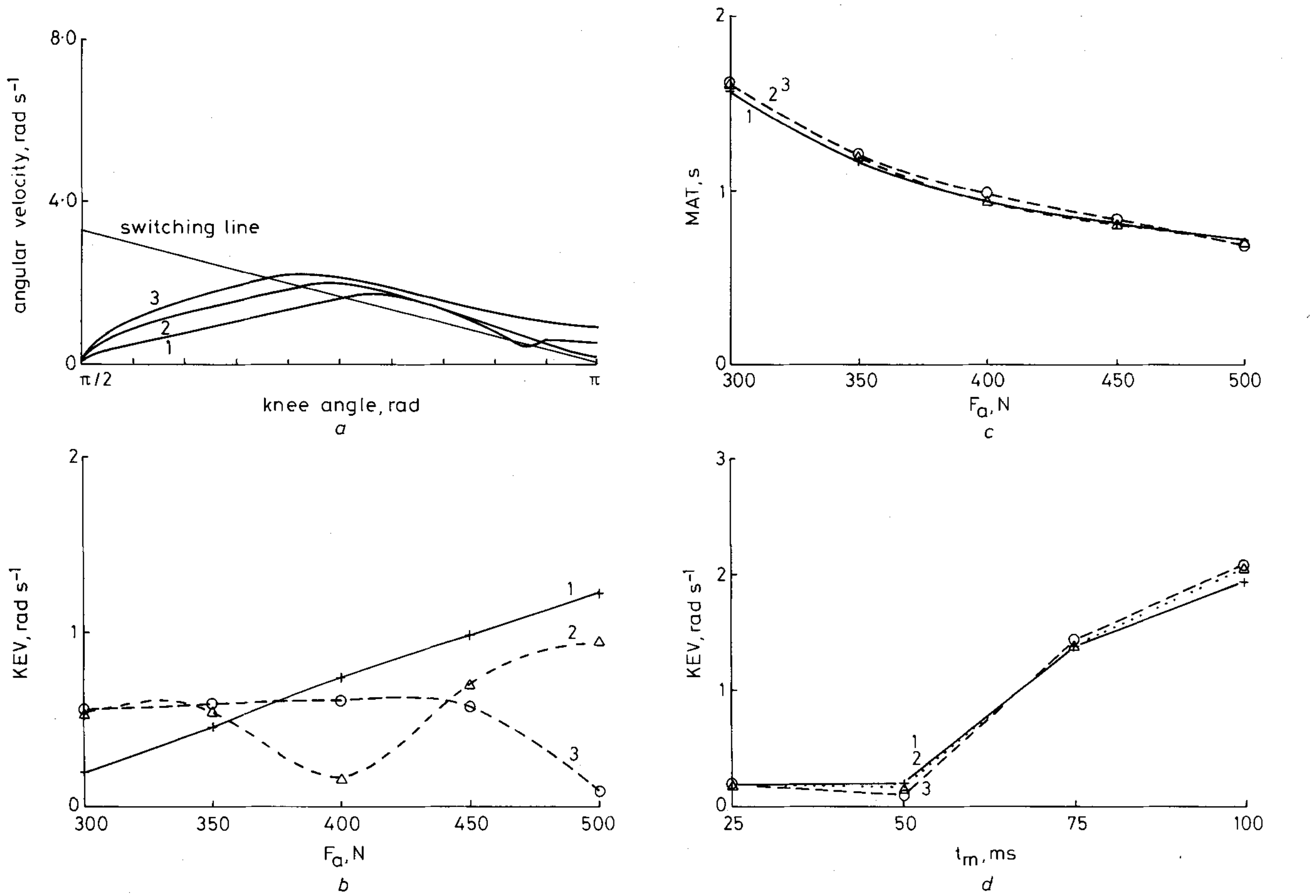


Fig. 6 *Simulation of closed-loop on/off controlled standing up. (a) State trajectories of knee joint when using the linear switching function of eqn. 4 for $S = -2{\cdot}10\,s^{-1}$. F_a is (1) 300 N; (2) 400 N; or (3) 500 N. $t_m = 50$ ms. Starting angle is $\pi/2$ rad. For low arm force (curve 1) extra switch-on of stimulation is required. For values of KEV and MAT see (b) and (c). (b) KEV against F_a for three values of S, such that S has the required value to obtain minimum KEV at (1) $F_a = 300$ N; (2) 400 N; or (3) 500 N. ((1) $S = -2{\cdot}12\,s^{-1}$; (2) $-2{\cdot}03\,s^{-1}$; and (3) $-1{\cdot}85\,s^{-1}$). $t_m = 50$ ms. Average and peak values of KEV are: (1) 0·85, 1·22; (2) 0·69, 0·95; and (3) 0·58, 0·61 rad s^{-1}. (c) MAT against F_a. S and t_m as in (b). (d) KEV against t_m for (1) $F_a = 300$ N; (2) 400 N; and (3) 500 N. For each curve S was adjusted to obtain minimum KEV at $t_m = 50$ ms as in (b)*

apparatus decelerates the knee before full flexion. From the intermediate part of the trajectory the average slope was estimated using a least square method, and was found to be $2{\cdot}5 \pm 0{\cdot}1\,s^{-1}$ ($r > 0{\cdot}97$).

During closed-loop control of stimulation (using the linear switching function of eqn. 4) the sensitivity of *KEV* with respect to *S* was determined. Fig. 7*c* shows the typical state trajectories for three value of *S*. Compared with the model simulations of Fig. 6*a* the trajectories show deviations after switching off stimulation. Initially angular velocity rapidly increased and decreased corresponding to the model study. Then it increased without stimulation being switched on. A possible explanation may be switching stimulation off to cause reflex activity, although this

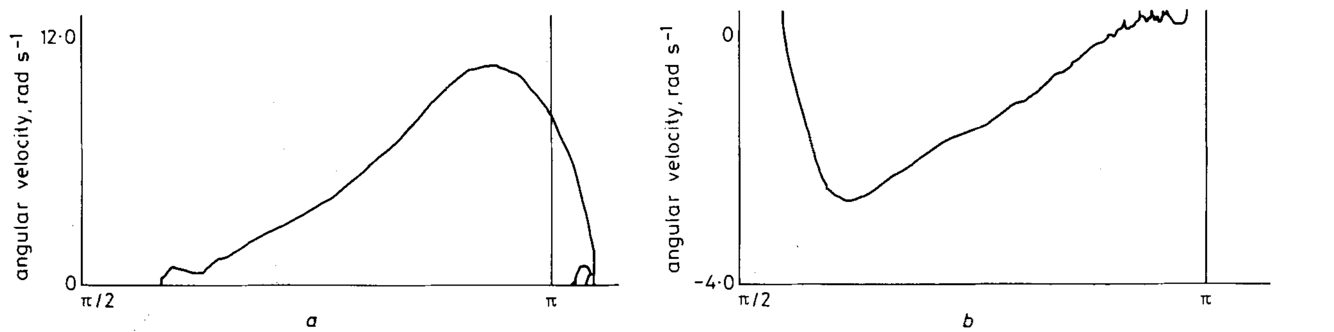
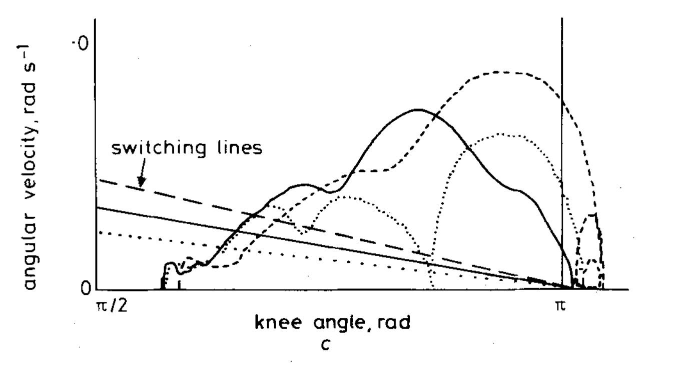


Fig. 7 *Typical examples of the experimental results of subject 1. Shown are state trajectories of knee joint for: (a) open-loop controlled stimulation ($KEV = 9{\cdot}0 \pm 0{\cdot}7$ rad s^{-1}, $MAT = 0{\cdot}74 \pm 0{\cdot}05$ s); (b) stand-to-fit movement and (c) closed-loop control of stimulation for $S = -1\,s^{-1}$ (dotted, $KEV = 4{\cdot}5 \pm 0{\cdot}1$ rad s^{-1}, $MAT = 0{\cdot}58 \pm 0{\cdot}1$ s), $S = -1{\cdot}5\,s^{-1}$ (solid, $KEV = 3{\cdot}7 \pm 1{\cdot}8$ rad s^{-1}, $MAT = 0{\cdot}49 \pm 0{\cdot}06$ s) and $S = -2\,s^{-1}$ (broken, $KEV = 6{\cdot}3 \pm 0{\cdot}1$ rad s^{-1}, $MAT = 0{\cdot}50 \pm 0{\cdot}01$ s). Corresponding switching lines are indicated*



was not validated. Compared with during open-loop stimulation average KEV was reduced to 70 ± 1 per cent ($S = -2{\cdot}0$), 41 ± 18 per cent ($S = -1{\cdot}5$) and 50 ± 1 per cent ($S = -1{\cdot}0\,s^{-1}$). For $S = -1{\cdot}0\,s^{-1}$ extra switch-on of stimulation was required. The optimum value of S will be between 1·0 and $1{\cdot}5\,s^{-1}$. Compared with during open-loop stimulation MAT was reduced to 68 ± 1 per cent ($S = -2{\cdot}0\,s^{-1}$), 66 ± 8 per cent ($S = -1{\cdot}5\,s^{-1}$) and 78 ± 13 per cent ($S = -1{\cdot}0\,s^{-1}$), respectively. For $S = -1{\cdot}5$ and $S = -2{\cdot}0\,s^{-1}$ no significant difference in MAT was found. For $S = -1{\cdot}0\,s^{-1}$ extra switch-on of stimulation significantly increased MAT compared with the other values of S (sign test, level of significance $<0{\cdot}05$).

4.2.2 *Subject 2:* The stimulation experiments for subject 2 showed high nonreproducibility. This is illustrated in Fig. 8. Fig. 8*a* shows the results of two consecutive open-loop experiments. There was an increased overall angular velocity in the second experiment. As also the second experiment showed a more regular movement, non-reproducibility is expected to be caused by spasticity of knee flexors. KEV ranges from 0·6 to $6{\cdot}2\,rad\,s^{-1}$. In Fig. 8*b* the stand-to-sit movement of subject 2 is shown. Compared with Fig. 7*b* there was an increased acceleration at the start of movement, near knee extension. This may be caused by contractures of hamstring muscle and/or obstructions in the knee joint which were clearly present. The closed-loop results of this patient are not presented, because spasticity deteriorated any controller effect on KEV or MAT. Although spasticity tends to cease during prolonged stimulation, it eliminated the possibilities for control of standing up in this patient.

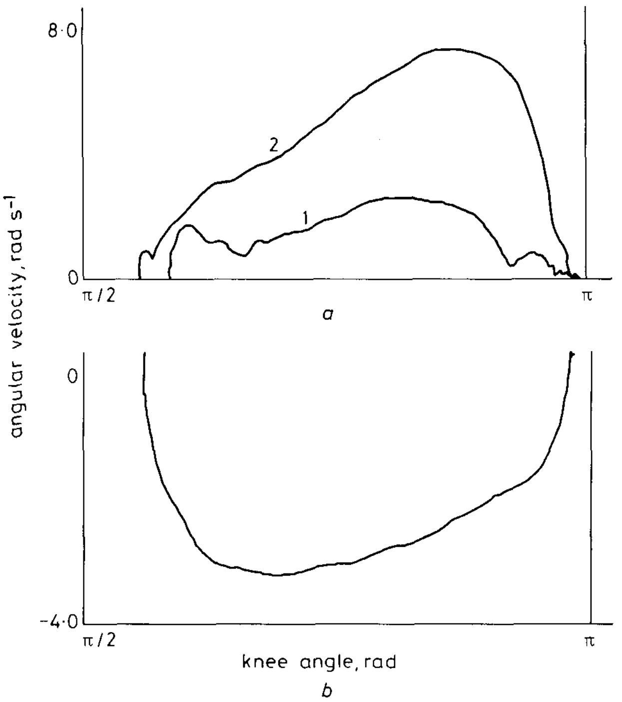


Fig. 8 *Typical examples of the experimental results of subject 2. (a) State trajectories of two consecutive open-loop experiments (1: first experiment, 2: second experiment). KEV varies from 0·6 to 6·2 rad s⁻¹, which is probably due to spasticity of the knee flexors in curve 1. (b) State trajectory of stand-to-sit movement*

5 Discussion

The question whether mobility restoration by FES may damage paraplegic patients joints is still not answered (Solomonow, 1989*a*). However, it may be an important consideration when designing strategies for FES controlled motion thus stressing e.g. limitation of joint end-velocities. In traditional orthopaedics, paraplegic joints are stabilised and protected by applying mechanical bracing (Rosman and Spira, 1974). Combined with FES bracing is still considered useful, especially in stabilising the hip (Petrofsky *et al.*, 1985; McClelland *et al.*, 1987; Solomonow *et al.*, 1989*b*) but also for stabilising the knee (Andrews *et al.*, 1989). However, for spinal cord injured patients (with good excitability of muscle) bracing as yet is a passive and bulky solution.

In current FES-induced open-loop controlled paraplegic standing up, voluntary arm force (needed for antigravity support and for keeping balance (Bajd *et al.*, 1982; Kralj *et al.*, 1980; Isakov *et al.*, 1986)) offers the only possibility to actively control knee end-velocity KEV. Our model study predicts only limited effect of arm force on reducing KEV: maximum reduction of arm force is predicted to reduce KEV to 67 per cent of the end-velocity obtained with twice the minimum arm force required for standing up. Further reduction requires closed-loop control of knee extensor muscle activation.

Most of the closed-loop control schemes proposed for FES are based on classical methods for linear systems, controlling force, position or velocity according to a predetermined trajectory in time (Stanic and Trnkoczky, 1974; Petrofsky and Phillips, 1979; Crago, 1983; Jaeger, 1986; Wilhere *et al.*, 1985). Some exceptions are found in the work of Petrofsky *et al.* (1984) who used 'successive windowing' to control joint angle in paraplegic standing and walking, whereas Andrews *et al.* (1989) report on rule-based (on/off) control in hybrid-assisted locomotion. However, none of the proposed strategies is aiming for reduced knee end-velocity during standing up. In the present study feedback of a limited number of state variables clearly resulted in limited knee end-velocity during standing up. Control was based on the use of a limited number of criteria and boundary conditions, not defining predetermined joint trajectories in time.

Several assumptions applied in our model may have bearing on the interpretation of the results. The assumption of the lower leg being vertical implies the presence of dorsal flexing ankle moment. As active ankle moment is not present in paraplegic patients, in clinical practice it is usually generated by slight backward positioning of the patients' feet before standing up (Kralj *et al.*, 1980; Isakov *et al.*, 1986). However, as the lower leg is not completely fixed, angular velocity may exceed model values, especially near full knee extension. Another aspect affecting angular velocity may be neglecting the effect of muscle length/velocity on muscle tension (Hill, 1938). This will reduce actual angular velocity compared with the model situation. The above effects may partially counteract each other but, more importantly, these deviations between model output and actual standing up will be limited in on/off controlled stimulation where knee extensors are off during a significant part of the movement, especially when reaching the upright position. When comparing the model predictions with data on normal standing up, our model study predicts transfer times in the same range as found for normals (1·3–2·5 s for normals (Nuzik *et al.*, 1986) against 1·0–2·7 s in model simulations).

Our study indicates introduction of closed-loop control to have only minor effect on metabolic energy consumption of knee extensor muscle (as estimated from MAT). On the other hand closed-loop control increases stand-up time significantly and energy consumption of upper body musculature is therefore expected to considerably increase. However, considering the incidental character of the stand-up movement (in contrast with maintained stable standing) priority was given to reduction of *KEV* over energy criteria.

The experimental situation represented real standing up but was not identical to it. In the experimental setup with the patient in the supine position no arm force was present for input. This was compensated by applying low external leg load compared with the trunk mass used in the model. This can explain both the relative high open-loop end-velocity and low acceleration during stand-to-sit in the experimental results of subject 1 compared with the model results. In subject 2 knee obstructions counteracted this phenomenon. During the experiments the variability of *KEV* and *MAT* exceeded that found in the simulations. This may be caused by the limited activation frequency (20 Hz) as applied during the experiments (corresponding to an interpulse interval of 50 ms), whereas during simulation a calculation step of 5 ms was applied.

In the current study only gravity was used to decelerate the body. The use of hamstrings was not considered. In our opinion this would make the switching criterion sensitive to the maximum torque and dynamics of the hamstrings, which sensitivity is expected to be high considering the direction of the force vector at the knee joint. This may correspond to the observed sensitivity of *KEV* for spasticity of upper leg musculature.

The results indicate optimal controller performance in the case of hypermobility of the knee joint, and the absence of spasticity. An assessment of the paraplegic subjects who finished the walking programme of the Roessingh Rehabilitation Centre over the last four years shows that 40 per cent might benefit from the proposed control strategy. This percentage may increase as spasticity can usually be decreased by a period of (low level) stimulation (STEFANOVSKA *et al.*, 1989), which could be considered prior to standing up.

Real-time adjustment of the switching curve to actual arm force and to the estimated knee extensor output during movement may further improve on the system performance in terms of reduction of *KEV*. However, the expected higher number of switching actions may not necessarily improve overall convenience for the patient as well.

Acknowledgment—This work was supported by the Netherlands Technology Foundation, St. Joris Foundation and IOP-HG.

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Appendix

Dynamic equation of the mechanical system

Fig. 1 gives the definition of variables and parameters used to determine the dynamic system equations. Using the Lagrangian formalism the moment of force T_i for joint i can be calculated from:

$$T_i = \frac{d}{dt}\left(\frac{\delta L}{\delta \dot{\theta}_i}\right) - \frac{\delta L}{\delta \theta_i} \tag{5}$$

where θ is the joint angle, $\dot{\theta}$ the angular velocity, $L = E_k - E_p$ where E_k is the kinetic and E_p the potential energy of the system. In general

$$E_k = \tfrac{1}{2}\left(\sum_{i=1}^{N} (m_i v_i^2 + J_i \dot{\theta}_i^2)\right) \text{ and } E_p = \sum_{i=1}^{n} m_i g y_i \tag{6}$$

where N is the total number of segments, m_i the mass, J_i the moment of inertia, v_i the velocity of the centre of mass, $\dot{\theta}_i$ the angular velocity and y_i the vertical displacement of the centre of mass for segment i, and g the acceleration of gravitation.

Under the assumption of a fixed lower leg we can write for segment 1

$$E_{p,1} = -\tfrac{1}{2} m_1 g l_1 \cos(\theta_1) \qquad E_{k,1} = (1/8) m_1 l_1^2 \dot{\theta}_1^2 + \tfrac{1}{2} J_1 \dot{\theta}_1^2$$

and for segment 2

$$E_{p,2} = -m_2 g(l_1 \cos(\theta_1) + \tfrac{1}{2} l_2 \cos(\theta_1 + \theta_2))$$

$$E_{k,2} = \tfrac{1}{2} m_2 v_2^2 + \tfrac{1}{2} J_2 (\dot{\theta}_1 + \dot{\theta}_2)^2$$

$$= \tfrac{1}{2} m_2 \left(\left(\frac{d}{dt} x_2\right)^2 + \left(\frac{d}{dt} y_2\right)^2\right) + \tfrac{1}{2} J_2 (\dot{\theta}_1 + \dot{\theta}_2)^2$$

where

$$x_2 = l_1 \sin(\theta_1) + \tfrac{1}{2} l_2 \sin(\theta_1 + \theta_2)$$

$$y_2 = l_1 \cos(\theta_1) + \tfrac{1}{2} l_2 \cos(\theta_1 + \theta_2)$$

The knee joint moment T_1 can now be derived from substitution in eqn. 5, using the kinematic constrain $\theta_2 = -(3/2)\theta_1 + (3/2)\pi$ as found in Section 2.1:

$$\begin{aligned} T_1 = \ddot{\theta}_1(&-(1/8) m_2 l_2^2 + (1/4) m_1 l_1^2 + m_2 l_1^2 \\ &- (1/4) m_2 l_1 l_2 \sin((3/2)\theta_1) + J_1 - \tfrac{1}{2} J_2) \\ &- (3/8) m_2 l_1 l_2 \cos((3/2)\theta_1)\dot{\theta}_1^2 + \tfrac{1}{2} m_1 g l_1 \sin(\theta_1) \\ &+ m_2 g l_1 \sin(\theta_1) - \tfrac{1}{2} m_2 g l_2 \cos(\tfrac{1}{2}\theta_1) \end{aligned} \tag{7}$$

Authors' biographies

Arjan Mulder was born in Nieuwleusen, The Netherlands, in 1958. He received his M.Sc. degree in Electrical Engineering from the University of Twente in 1984. He received his Ph.D. in Biomedical Engineering from the same university in 1991, the topic of his dissertation being finite state control in functional neuromuscular stimulation. Since 1984 he has been working as a part of the rehabilitation engineering collaboration between the University of Twente and the Roessingh Rehabilitation Centre. His current activities are in the field of neuromuscular implant technology.

Peter H. Veltink studied Electrical Engineering at the University of Twente, Enschede, The Netherlands. He received the M.Sc. degree in 1984 and the Ph.D. degree in 1988 (on recruitment of myelinated nerve fibres during artificial nerve stimulation). In 1989 he visited Case Western Reserve University in Cleveland, where he researched nonlinear control of muscle contraction. Currently, he is a member of the Biomedical Engineering Division of the Department of Electrical Engineering, University of Twente. His research is in control of functional neuromuscular stimulation, which is carried out in co-operation with the Roessingh Rehabilitation Centre in Enschede.

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.