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Cycle-to-cycle control of swing phase of paraplegic gait induced by surface electrical stimulation

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Abstract—*Parameterised swing phase of gait in paraplegics was obtained using surface electrical stimulation of the hip flexors, hamstrings and quadriceps; the hip flexors were stimulated to obtain a desired hip angle range, the hamstrings to provide foot clearance in the forward swing, and the quadriceps to acquire knee extension at the end of the swing phase. We report on two main aspects; optimisation of the initial stimulation parameters, and parameter adaption (control). The initial stimulation patterns were experimentally optimised in two paraplegic subjects using a controlled stand device, resulting in an initial satisfactory swinging motion in both subjects. Intersubject differences appeared in the mechanical output (torque joint) per muscle group. During a prolonged open-loop controlled trial with the optimised but unregulated stimulation onsets and burst duration for the three muscle groups, the hip angle range per cycle initially increased above the desired value and subsequently decreased below it. The mechanical performance of the hamstrings and quadriceps remained relatively unaffected. A cycle-to-cycle controller was then designed, operating on the basis of the hip angle ranges obtained in previous swings. This controller successfully adapted the burst duration of the hip flexors to maintain the desired hip angle range.*

Keywords—*Cycle-to-cycle control, Functional electrical stimulation, Lower extremities, Paraplegia*

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1 Introduction

THE APPLICATION of cyclic functional electrical stimulation (FES) patterns to lower extremity muscles can, to a certain extent, result in paraplegic locomotion (ANDREWS *et al.*, 1989; CHIZECK *et al.*, 1988; KRALJ *et al.*, 1983; MARSOLAIS and KOBETIC 1987; PETROFSKY *et al.*, 1986; YAMAGUCHI and ZAJAC, 1989). The practical success of this technique is still seriously limited, owing to the occurrence of muscle potentiation and fatigue (BOOM *et al.*, 1993; FRANKEN *et al.*, 1993); inadequate timing of applied stimulation (YAMAGUCHI and ZAJAC, 1989); variability in response of stimulated muscle (TRNKOCZY, 1974); muscle spasms (STEFANOFSKA *et al.*, 1989); poor selectivity and electrical accessibility of muscles that generate gait propulsion; and sensitivity to external disturbances. These problems are most evident when stimulation is effected with electrodes on the skin (surface FES). In any case, there are two distinct problems to be resolved to restore functional movement. First, parameters for the cyclical stimulation patterns need to be chosen and optimized for each individual patient and situation. Secondly, these parameters, once selected, have to be continuously adapted to compensate for external disturbances and system performance deterioration such as fatigue.

Several methods to derive stimulation patterns for cyclical leg motions have been reported (CHIZECK *et al.*, 1988; MARSOLAIS and KOBETIC, 1987; MCNEAL *et al.*, 1989; STANIC *et al.*, 1978; YAMAGUCHI and ZAJAC, 1989). McNeal *et al.* used a trial-and-error method to determine the appropriate stimulation pattern of the hamstrings and the quadriceps in order to control the position of the lower leg in a cyclic motion (trajectory following). The resulting stimulation sequence was applied in open loop. Initial responses of the lower leg matched the desired trajectory very well. However, the response significantly deteriorated owing to fatigue when the stimulation patterns were applied for several minutes. Yamaguchi and Zajac computed stimulation sequences which, on the basis of a bio-mechanical simulation model, would restore unassisted paraplegic gait. Stimulation sequences were derived using dynamic programming and a trial-and-error adjusting method. The stimulation sequence was optimized by

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minimising a cost function, supposedly related to minimal energy consumption of the stimulated muscles. The resulting suboptimal set of stimulation patterns yielded a step motion in their computer model. However, this procedure was not validated experimentally. As such, these stimulation patterns were thought to be applied in open loop. Thus, in the method of Yamaguchi and Zajac the effect of fatigue was also not accounted for.

Stanic *et al.* derived open-loop multichannel stimulation patterns for the correction of hemiplegic gait, successfully preventing several anomalities (STANIC *et al.*, 1977; VODOVNIK *et al.*, 1981). The stimulation patterns were synchronised with the heel-on event, detected by a foot switch. No strategy was included to adapt the stimulation patterns when fatigue occurs. Marsolais and Kobetic and Chizeck *et al.* reported the walking capabilities of paraplegics with the FES system at their laboratory. Stimulation sequences were derived from trial-and-error experimentation. During functional use, the patient manually initiated the open-loop stimulation resulting in a step. The stimulation initially induced exaggerated joint movements to provide a margin of safety in the case of muscle fatigue. The walking performance deteriorated over time owing to fatigue. Thus, investigations indeed show that optimal stimulation patterns can be derived following several strategies, but do not remain optimal if not adapted to compensate for system performance deterioration.

Stimulation sequences can be derived and adapted by means of (closed-loop) control strategies. This can account for both system performance deterioration, such as muscle fatigue, and non-linear system characteristics, such as muscle dynamics. Hausdorff and Durfee demonstrated the use of an open-loop feedfoward position controller for the knee joint in healthy human subjects, stimulating the quadriceps and hamstrings (HAUSDORFF and DURFEE, 1991). They concluded that it was beneficial to account for the non-linear length dependence of stimulated muscle output. Muscle fatigue was still not compensated for. Hatwell *et al.* designed a model reference adaptive controller for the same system as Hausdorff and Durfee for paraplegics (HATWELL *et al.*, 1991). The stimulation pulse width for the quadriceps and hamstrings were computed on-line. The stimulation amplitude and frequency were kept constant. A reference knee angle signal needed to be tracked; the tracking performance of this position controller was poor.

In fact, it was questionable whether instantaneous control of the system's (joint angle) trajectory (WILHERE *et al.*, 1987; CHIZECK *et al.*, 1991; HATWELL *et al.*, 1991; LAN *et al.*, 1991) is possible or even essential for the lower extremity system. The limited mechanical output (joint torque) of the muscles in relation to the inertial properties of the lower extremity system limits the transient response. The trajectory of the leg is therefore mainly determined by its inertial constraints. For the same reasons, disturbances can hardly be compensated for. The decay of functional output due to muscle fatigue can best be compensated for by adapting the stimulation burst duration for subsequent swing cycles, in the basis of the performance in previous cycles.

Veltink experimentally tested this cycle-to-cycle concept in healthy human subjects in the control of a cyclical lower leg movement with quadriceps stimulation (VELTINK, 1991). The objective was that the lower leg should reach a certain maximal knee angle each swing cycle. The open-loop burst duration was adapted in each cycle on the basis of the error between the desired and actually obtained maximal knee angle in previous cycles, successfully maintaining the desired objective. Vodovnik *et al.* suggested the feasibility of cycle-to-cycle adaption of the duration of stimulation sequences to maintain symmetric gait in paretic patients (VODOVNIK *et al.*, 1981). Cycle-to-cycle approaches to inhibit habituation of afferent stimulation techniques (flexion withdrawal reflex) have also been reported (GRANAT *et al.*, 1991).

In this paper, we experimentally investigated the control of the swing phase of paraplegic gait by cyclically stimulating the hip flexors, hamstrings and quadriceps. The initial stimulation patterns were experimentally optimised to obtain the desired movement. The swing phase was parameterised on the level of the cycle. This furthermore enabled cycle-to-cycle adaptation of the applied stimulation as a means to compensate for slowly potentiation- and fatigue-induced time-varying muscle output. Thus, the desired swing phase objectives could not only be established, but also maintained.

2 Theory

2.1 *System description*

The considered lower extremity system with the stimulated muscle groups is shown in Fig. 1. The system was restricted to the sagittal plane by an orthosis. In these experiments, surface electrodes were used for stimulation. When stimulating hip flexors, we usually activated the iliopsoas, the rectus femoris, and hip abductors and adductors (guided by the orthosis). When stimulating the hamstrings, we also partly activated the biceps femoris (short head). Quadriceps stimulation resulted in activation of the rectus femoris, vastus lateralis and vastus medialis.

2.2 *Parameterised swing phase*

To derive the required stimulation patterns for the hip flexors, hamstrings and quadriceps, we observed the motion of the leg on the level of the swing cycle (Fig. 2). The generated movement was imposed to satisfy three parametrising conditions (Fig. 3) (described here as swing

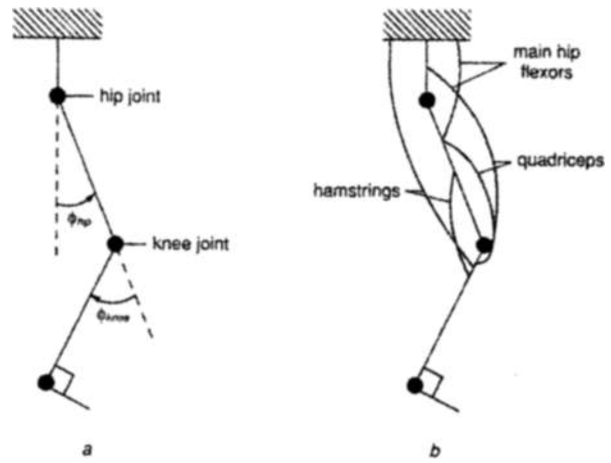


Fig. 1 *(a) Leg, freely swinging in sagittal plane; fixed hip position; orthosis restricted movement to sagittal plane and fixed ankle joint to constant angle; (b) when referring to stimulation of hip flexors, iliopsoas and rectus femoris were probably activated; hip ab- and adductors were probably also activated (not shown); orthosis restricted movement to sagittal plane; knee extension obtained by stimulating quadriceps, i.e. vasti and rectus femoris; foot clearance obtained by flexing knee joint by means of hamstring stimulation; biceps femoris (short head) probably also partly activated during hamstring stimulation*

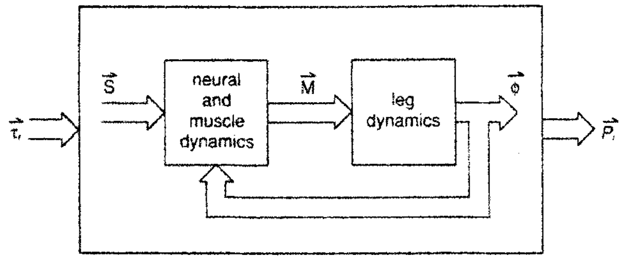


Fig. 2 *Freely swinging leg system with stimulation vector $\bar{\tau}_i$ and obtained swing phase objectives $\bar{P}_i$ for cycle number i; freely swinging leg system consists of neural, muscle and leg dynamics; input stimulation $\bar{\tau}_i$ activates neural and muscle dynamics, indicated by $\bar{S}$; activated muscles generate torque around hip and knee joints, depicted by $\bar{M}$; this induces joint movement $\bar{\varphi}$; in each cycle input vector $\bar{\tau}_i$ is applied to system and swing parameters are observed $\bar{P}_i$*

phase objectives), which are characteristic for natural human gait (INMAN *et al.*, 1981). First, a nominal hip angle range was desired that determines the step length during gait. The minimum foot-clearance in the forward swing then had to be sufficient to prevent contact with the ground. This is essential to prevent the subject from stumbling during gait. To ensure subsequent safe body weight bearing, knee extension with activated quadriceps was desired at the end of the forward swing, as a third condition.

2.3 *Cycle-to-cycle PID control strategy*

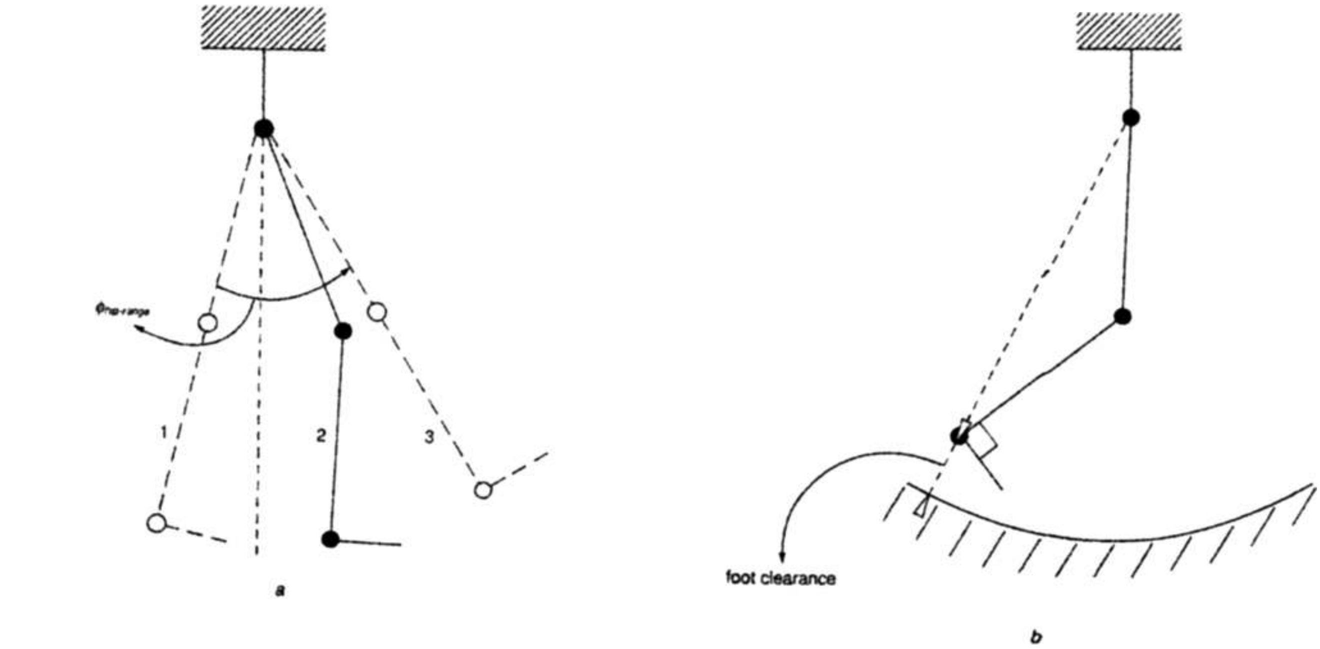
2.3.1 *Algorithm*: a discrete-time PID-controller (Fig. 4) adjusted the burst duration of the hip flexors from cycle to cycle to account for their time-varying mechanical output, thus maintaining the desired hip angle range. As the mechanical outputs of the quadriceps and hamstrings were relatively time-invariant, their stimulation patterns were not adjusted. The controller computed a new burst duration at the end of each swing cycle; the time step of the PID controller was thus the swing cycle time T_c. The controller did not instantaneously account for disturbances during a swing cycle, as the stimulation patterns were then applied in open loop. Adjustment was made at the very beginning of the cycle. In the z-domain, the controller is given (PHILLIPS and NAGLE, 1984) by

$$PID(z) = T_{burst\text{-}hip}(z)/\varepsilon(z) = \{G(z-z_1)(z-z_2)\}]/[\{z(z-1)\} \quad (1)$$

where $T_{burst\text{-}hip}$ is the burst duration for the hip flexors, ε is the difference between the desired and actually obtained hip angle range, G is the gain and $z_{1,2}$ are the zeros of the PID controller. The corresponding discrete-time algorithm is

$$T_{burst\text{-}hip,n} = T_{burst\text{-}hip,n-1} + G\{\varepsilon_{n-1} - (z_1 + z_2)\varepsilon_{n-2} + z_1 z_2 \varepsilon_{n-3}\} \quad (2)$$

where n denotes the cycle number.



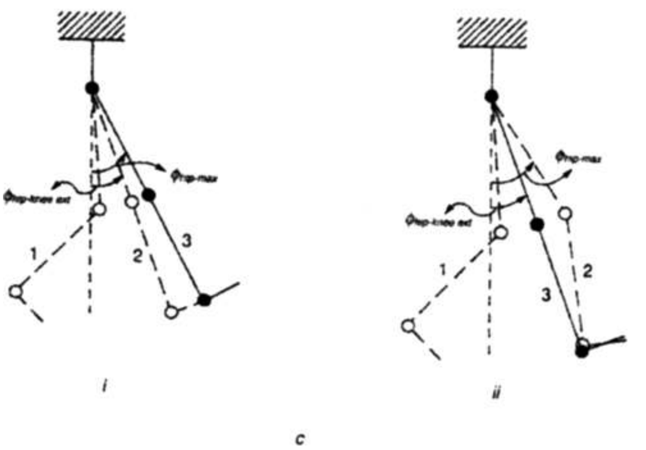


Fig. 3 *Swing phase objectives: (a) hip angle range defined as difference between minimum and maximum hip angle obtained during one cycle; (b) foot clearance computed as distance between heel and imaginary ground; imaginary ground surface assumed to be perfect circle with hip joint as origin and distance of hip joint to heel as radius; foot clearance only computed in hip range of 10° around fixed threshold angle (see methods), as foot contact was desired at end of swing phase; (c) knee extension expressed in terms of hip angle error; defined as absolute difference between hip angle at maximum hip flexion $\varphi_{\text{hip-max}}$ and hip angle at knee extension $\varphi_{\text{hip-knee-ext}}$; (i) and (ii) leg swings from state 1 via 2 to 3; discrimination between (i) and (ii) expressed in terms of sign of hip angle error; when knee extension occurs too early (i), sign is negative; when knee extension occurs too late (ii), sign is positive*

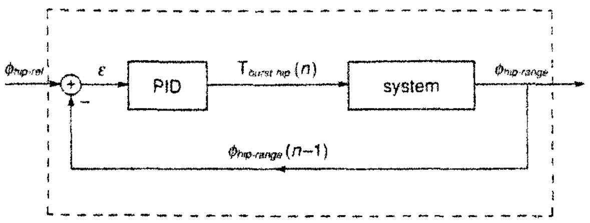


Fig. 4 *Discrete-time PID controller to adjust hip flexor stimulation burst duration for next cycle ($T_{burst\text{-}hip}(n)$) on basis of error ε between desired hip angle range $\varphi_{hip\text{-}ref}$ and hip angle range obtained in previous swing $\varphi_{hip\text{-}range}(n-1)$*

2.3.2 *Tuning*: the gain, poles and zeros of the combined system and controller determine the time-characteristics of a (linear) closed-loop system. The PID controller was tuned to an experimentally identified model of the total system on the level of a cycle. In this model, the leg was considered as a single pendulum, with the motion of the lower leg neglected. Model characteristics were obtained from a passive swing of the leg. The leg was raised to a certain hip angle and subsequently released. The hip angle range was damped subcritically in the form of a sine signal. This impulse-type decay was modelled as a linear second-order system. The ranges of subsequent cycles, which are the chronological peak-to-peak values in the impulse response, were expressed as a function of the cycle number (VELTINK, 1991)

$$\varphi_{n,hip\text{-}range} = ba^n \tag{3}$$

in the z-domain eqn. 3 becomes

$$\varphi_{hip\text{-}range}(z) = bz/(z-a) \tag{4}$$

Thus, the system can be modelled with a pole at $z = a$ and a zero at $z = 0$, which is a first-order model with time step T_c. The parameter a is determined by the damping (in the hip joint), as modelled in the linear second-order system, and b is determined by the release angle. The controller was to compensate for the slow time variation in mechanical output of the hip flexors caused by potentiation and fatigue (BOOM *et al.*, 1993). It should not become oscillatory (unstable) owing to high-frequency noise on, or stochastic variation in, the obtained hip angle range. The PID controller was tuned by cancelling system pole a with $z_1 = a$ (eqn. 1). The second zero was placed at $z_2 = 0{\cdot}2$ to obtain a small settling time in the oscilatory closed-loop response. The gain G was experimentally adjusted to obtain approximately 15% overshoot to a step response in the desired hip angle range.

3 Methods

3.1 *Subjects*

Two subjects participated in this study. They were complete T5-T6 level spinal cord injured patients (Table 1*a*). The degree of spasticity is given by the modified Ashworth Scale (SLOAN *et al.*, 1992) in Table 1*b*. Both subjects had been enrolled in the FES training programme of the Roessingh Rehabilitation Centre (Enschede, The Netherlands), hoping for the restoration of locomotor functions. The training programme initially concentrates on muscle strength training using surface FES (once a day, 30 min per muscle, external loads). Subsequently, standing up with quadriceps surface stimulation and ambulation with a reciprocating gait orthosis (RGO) are exercised. Locked knees and crutches ensure stability during locomotion. Manually controlled efferent surface stimulation of hip flexors and extensors induce propulsion.

Table 1 (a) Specific data on two participating paraplegic subjects; both subjects were complete T5-T6 level spinal cord injured patients

subject	sex	age	months after injury	FES training, months
A	M	18	40	15
B	M	42	28	6

(b) measures of spasticity of two participating paraplegic subjects according to modified Ashworth Scale (Sloan et al. 1992)

subject	movement	spasticity grade
A	hip flexion	2
	knee extension	1+
	knee flexion	2
	knee extension	1+
A	hip flexion	3
	hip extension	3
	knee flexion	3
	knee extension	3

the modified Ashworth Scale grades spasticity on basis of muscle activity during passive movements from 0 (no increase of muscle tone) to 4 (affected part(s) rigid in flexion or extension)

3.2 *Experimental set-up*

3.2.1 *Stand device:* to evaluate the effect of the stimulation patterns in paraplegic patients in a controlled situation, a standing frame with arm support was constructed (Fig. 5). It positioned the paraplegic subject, wearing a partly disassembled and modified advanced reciprocating gait orthosis (ARGO). The orthosis restricted the movement of the freely swinging leg in the sagittal plane; the ankle joint was locked. Without an orthosis, surface stimulation of the hip flexors generated hip ad- and abduction during the forward swing. The ankle, knee and hip joint of the standing leg were locked. A block supported the standing leg and elevated the entire body. Additionally, the subject was sitting on a bicycle saddle that provided additional support

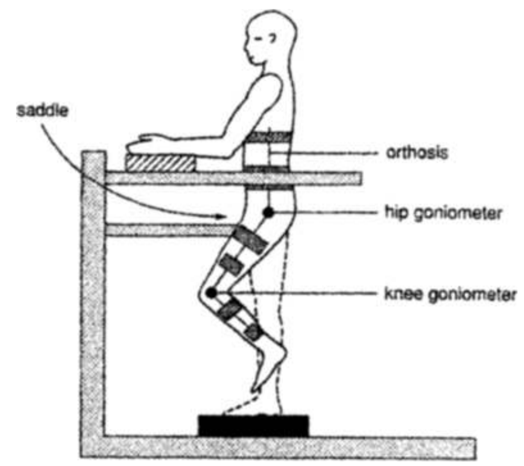


Fig. 5 *Paraplegic subject standing in stand device; subject wore self-fitting modular orthosis restricting motion of freely swinging leg with locked ankle joint to sagittal plane, and locking hip, knee and ankle joint of standing leg; standing leg elevated by block; bicycle saddle provided additional support; trunk and pelvis movements in sagittal and frontal plane prevented by set-up; hip and knee angles of freely swining leg measured by externally mounted goniometers*

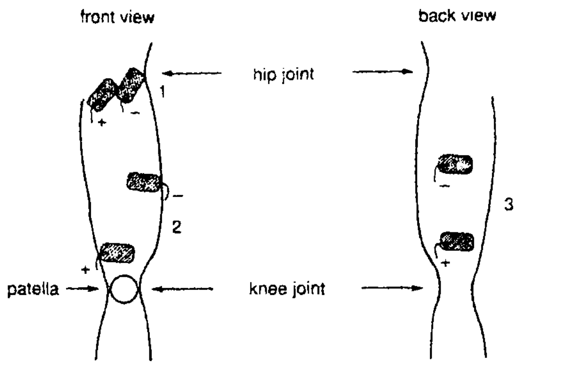


Fig. 6 *Electrode placement for stimulated muscle groups; anode (+) and cathode (−); electrodes for hip flexors (1) placed in groin fold; quadriceps (2) stimulated with electrodes placed over motor points of rectus femoris/vastus lateralis and vastus medialis, respectively; hamstrings (3) stimulated with two electrodes near knee joint*

in the frontal plane. The set-up prevented pelvis and trunk movements in the frontal and sagittal plane. Externally mounted goniometers (MCB pp 27c, 310°, non-linearity ≤1%) measured the hip and knee angle of the freely swinging leg.

3.2.2 *Stimulation, data recording and control*: adhesive surface electrodes* (5 × 9 cm) were used for stimulation, two electrodes (one anode and one cathode) for each muscle group. The placement of electrodes (Fig. 6) conforms to clinical usage for paraplegic standing and gait during the training programme. The PID control strategy was implemented on an IBM-AT compatible computer with AD facilities†, which also controlled a current stimulator§, generating monophasic rectangular pulses. The pulse width and pulse amplitude were fixed at maximal attainable recruitment without undesired spillover to other muscles (Durfee and Maclean, 1989; Franken *et al.*, 1993). This was necessary to obtain functional movement. The interpulse interval was chosen to be 40 ms for quadriceps and hamstrings and 20 ms for hip flexors (see discussion below). Fig. 7 illustrates the stimulation terminology used in this paper. The gonio signals were sampled at 100 Hz. These signals, together with the stimulus data, were stored on disk.

* Pals, Axelgaard Manufacting Co. Ltd., Fallbrook, California, USA
† Analog Devices, RTI-815, 12-bit
§ PLS 8, Roessingh R & D, Enschede, The Netherlands

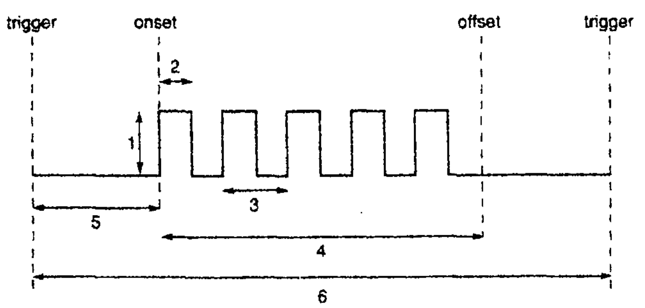


Fig. 7 *Stimulation nomenclature: 1 = pulse amplitude, 2 = pulse width, 3 = interpulse interval, 4 = burst duration, 5 = wait time, 6 = cycle time; after being triggered, onset of stimulation occurs when wait time has passed; offset of stimulation occurs exactly after duration of burst*

3.3 *Experimental optimisation*

The optimal stimulation onsets and burst durations of the three muscle groups were identified experimentally in three stages (Veltink *et al.*, 1992). Each muscle group was stimulated with a single burst per cycle. First, the onset time and burst duration of the hip flexors were determined without stimulating the quadriceps and hamstrings. The onset was adjusted such that hip flexor activation started at the beginning of the forward swing, similarly to natural human gait (Inham *et al.*, 1981). The burst duration was gradually incremented until the desired hip angle range was achieved. Hamstring stimulation was then added to obtain sufficient foot clearance in the forward swing. The onset of hamstring stimulation was optimised by studying the effects of an onset (with several fixed burst durations) prior to and after the onset of the hip flexors stimulation. Subsequently, the burst duration was adapted to obtain sufficient foot clearance. Quadriceps stimulation was then added to ensure knee extension. Quadriceps activation at the end of the forward swing was desired. Thus, when the mechanical output of the quadriceps was high, a short burst duration was sufficient, necessitating an onset late in the forward swing. If the applied stimulation resulted in a stable swinging motion, the swing phase objectives of the fifth cycle were examined.

Stimulation onsets were determined relative to a trigger instant (Fig. 7). The hip angle signal passing through a fixed threshold angle in the backwards swing triggered the stimulation of hamstrings and hip flexors (Fig. 8). Similarly, the forward swing triggered the quadriceps stimulation. The threshold angle was equal to the hip angle with the swinging leg at rest.

The desired values of the swing phase objectives (Section 2.2 and Fig. 3) were chosen as follows. The hip angle range was set to 45°, corresponding to the swing phase of natural human walking (Inman *et al.*, 1981). Sufficient foot clearance was assumed to be greater than 1 cm, to be obtained in a hip angle range of 10° centred around the threshold. Knee extension was considered to be reached when $\varphi_{knee} \leq 3°$.

3.4 *Protocol*

The experimental protocol was as follows.

(i) The three muscle groups were warmed up with a period of low-level stimulation in a sitting position. It has been reported that warming up prior to, for example, standing up or ambulation prevents the occurrence of spasticity (Stefanovska *et al.*, 1989).

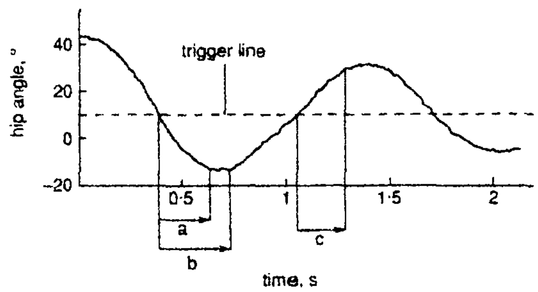


Fig. 8 *Derivation of onset timing for hamstrings, hip flexors and quadriceps; stimulation timing of hamstrings and hip flexors triggered on hip angle passing through threshold angle in backwards swing; quadriceps stimulation timing triggered identically in forward swing; wait time between trigger signal and actual onset of stimulation a = hamstring, b = hip flexors, c = quadriceps) facilitated easy adjustment of onset timing*

(ii) Stimulation onset and burst duration were optimized (Section 3.3).
(iii) In an open-loop controlled trial, the optimised stimulation sequences for the three muscle groups were applied for a period of 5 min.
(iv) After a rest period of 1 h, the cycle-to-cycle PID controller for burst duration adaptation of the hip flexors was tuned experimentally (Section 2.3.2).
(v) In a cycle-to-cycle controlled trial, the PID controller adapted the burst duration of the hip flexors to maintain a desired hip angle range for every swing cycle. The stimulation onset remained unchanged. The stimulation onsets and burst durations for hamstrings and quadriceps were identical to those of the open-loop experiment. The trial was terminated either after 5 min or when the burst duration of the hip flexors exceeded the duration of the forward swing (≥ 700 ms) for several successive cycles. The cycle-to-cycle controlled trial was performed twice on subject A (denoted as trials 1 and 2) without a significant rest period in between.

4 Results

Fig. 9 displays a typical recording of hip angle, knee angle and applied stimulation bursts during the open-loop controlled stimulation trial in subject A, with optimized stimulation onset and burst duration. The applied stimulation resulted in a stable swinging motion of the leg. The cycle time was approximately 1.3 s, which was mainly determined by the leg biomechanics. As pointed out, the hip flexors were stimulated at the beginning of the swing phase. Before the beginning of the swing phase, hamstring stimulation onset flexed the knee to obtain foot clearance. The quadriceps, activated at the end of the forward swing, realised knee extension with simultaneously activated quadriceps.

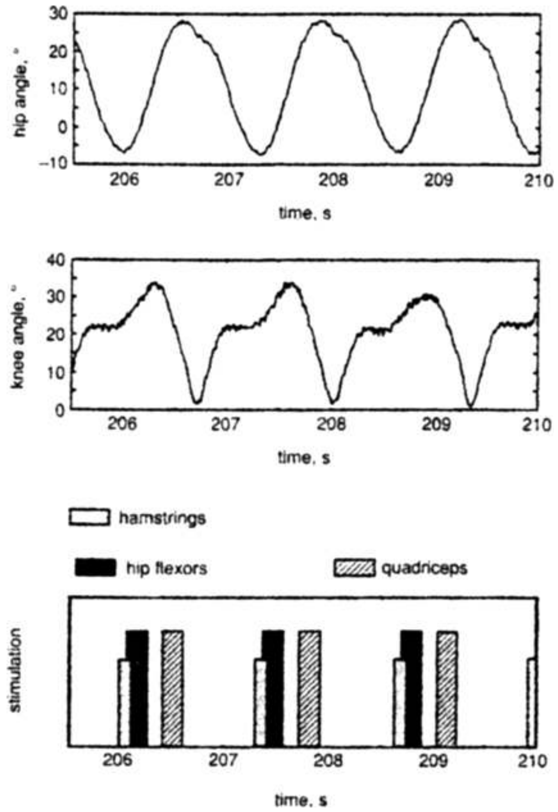


Fig. 9 *Hip and knee angle and stimulation onset and duration during open-loop stimulation trial in subject A with optimised stimulation for three muscle groups; applied stimulation timing was experimentally optimised, which resulted in desired parametrised swing in fifth swing cycle; hip flexors stimulated to obtain desired hip angle range; hamstrings provided foot clearance in forward swing; quadriceps stimulated to acquire knee extension at end of swing phase; stimulation sequences applied without adaptation; effect of each stimulation site can be detected from time course of hip and knee joint*

4.1 *Experimental optimisation*

Initially, we tried to obtain identical swing phase objectives in both subjects during the experimental optimisation. However, owing to the limited functional output of the stimulated hip flexors in subject B, a hip angle range of 25° was obtained. In subject A, a hip angle range of 45° could be obtained. The hip angle range at which foot clearance was to be obtained was therefore also smaller in subject B, at 5° centred around the threshold. Sufficient foot clearance and knee extension at the end of the forward swing were then obtained in both subjects. The wait times and burst durations for the three stimulated muscles groups and the threshold hip angle, which resulted in this parametrised swinging motion in the two subjects, are given in Table 2. Intersubject differences in the onset were caused by the different threshold angle at which the stimulation was triggered, which introduced a time shift in the onset (expressed in wait time). Generally, intersubject differences in burst duration were caused by differences in the mechanical output of the stimulated muscle.

A tentative significant increment of the burst duraction for the hip flexors in subject B barely improved the hip angle range. Furthermore, he exhibited significant spasm in his lower extremity muscles of the freely swinging leg while standing in the set-up (Table 1*b*). This limited the effect of

Table 2 Wait times and burst duration obtained in experimental optimisation procedure in subjects B and A

subject	threshold hip angle, °	muscle group	wait time, ms	burst duration, ms
		hamstrings	300	120
A	12	hip flexors	330	200
		quadriceps	190	160
		hamstrings	150	120
B	9	hip flexors	200	400
		quadriceps	50	200

stimulation of hamstrings and hip flexors triggered in backwards swing and stimulation of quadriceps in forwards swing on threshold hip angle: after being triggered, stimulation onset occurred when wait time had passed

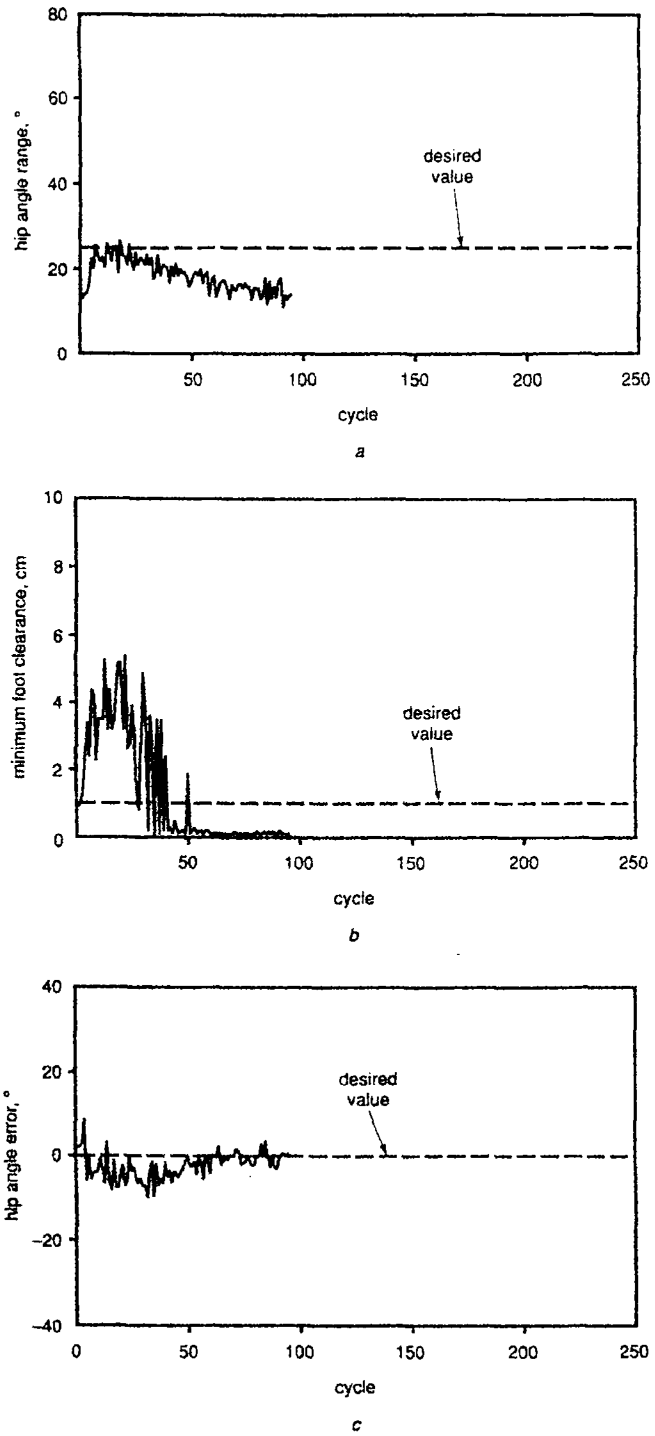


Fig. 10 *Swing phase objectives per swing cycle obtained in open-loop controlled stimulation trial with optimised stimulation timing for three muscle groups in subject B; (a) hip angle range; (b) minimum foot clearance; (c) hip angle error at knee extension; when knee extension occurs too early, sign of hip angle error is negative; when knee extension occurs too late, sign is positive (Fig. 3)*

the applied stimulation and deteriorated the swinging motion, complicating the optimisation of the stimulation onset and burst duration. Subject A exhibited only minor spasm in his lower extremity muscles (Table 1*b*). We also observed a control space in the burst duration of subject A's hip flexors. Incrementing it also caused significant increment of the hip angle range. Subject A's greater control space was possible available due to both the absence of spasm and the longer duration of training compared with subject B (Table 1*a*).

Adding stimulation of hamstrings provided foot clearance. Hoy *et al.* (HOY *et al.*, 1990) suggested that the hamstrings are mainly active over the knee joint when the hip is in extension. This was actually observed during our experiments (Fig. 9). Furthermore, the biarticular function of the hamstrings over the hip joint became more effective when the hip was also flexing. The hip angle range obtained was therefore significantly smaller when the hamstrings were stimulated simultaneously with the hip flexors, or when the leg had advanced significantly in forward swing. This also resulted in a less fluent or sometimes unstable motion of the leg. Only an onset of hamstring stimulation prior to the onset of the hip flexors stimulation resulted in sufficient foot clearance.

Quadriceps stimulation locked the knee at the end of the swing phase. When the onset of the quadriceps stimulation was too early, foot clearance was insufficient. This also necessitated a longer burst duration, as quadriceps activation was required at the end of the forward swing. When the hip flexors fatigued significantly, quadriceps stimulation provided a noticeable hip flexion impulse.

4.2 *Open-loop control*

4.2.1 *Hip angle range*: the time development of the hip angle range during the prolonged open-loop stimulation trial, with experimentally optimised stimulation patterns, is displayed in Figs. 10*a* and 11*a*. For both subjects, the hip angle range initially increased above the value obtained in the experimental optimisation, probably due to the effects of potentiation in the hip flexors, but also due to additional impulse due to quadriceps stimulation. Subsequently, the hip angle range decreased below this value due to hip flexor muscle fatigue. This decay is characteristic of the open-loop experiments in both subjects. The open-loop trial in subject B was terminated after approximately 95 cycles by subject request, due to continuous spasm.

4.2.2 *Minimum foot clearance*: in subject B, the obtained foot clearance was negligible after approximately 40 cycles (Fig. 10*b*). This was again caused by spasm in the lower extremity muscles, which sometimes stiffened the knee joint in full extension and limited the effect of the applied stimulation. Subject A's foot clearance stabilised within a few cycles to approximately 1·5 cm (Fig. 11*b*).

4.2.3 *Knee extension*: in subject B, knee extension at the end of the swing phase was obtained throughout the entire open-loop trial (Fig. 10*c*). However, owing to spasm in the lower extremity muscles, the knee was almost in extension during the entire swing from approximately cycle 40 onwards, explaining the time development of the obtained foot clearance and knee extension during the final part of the trial (Fig. 10*b* and *c*). In subject A, knee extension occurred at various hip angles, i.e. too early or too late, during the first 100 cycles; this was partly caused by the excessive magnitude of and variation in the hip angle range during this initial part of the open-loop trial (Fig. 11*a*). In addition, the relatively short burst duration of subject A's quadriceps, generating a high mechanical impulse on the lower leg, caused an oscillatory movement of the lower leg. Knee extension stabilised after approximately 100 cycles (Fig. 11*c*) when the hip angle range also stabilised. Simultaneous control of recruitment and burst duration, i.e. spreading the mechanical impulse of the quadriceps on the lower leg in time, might avoid the oscillatory response in subject A.

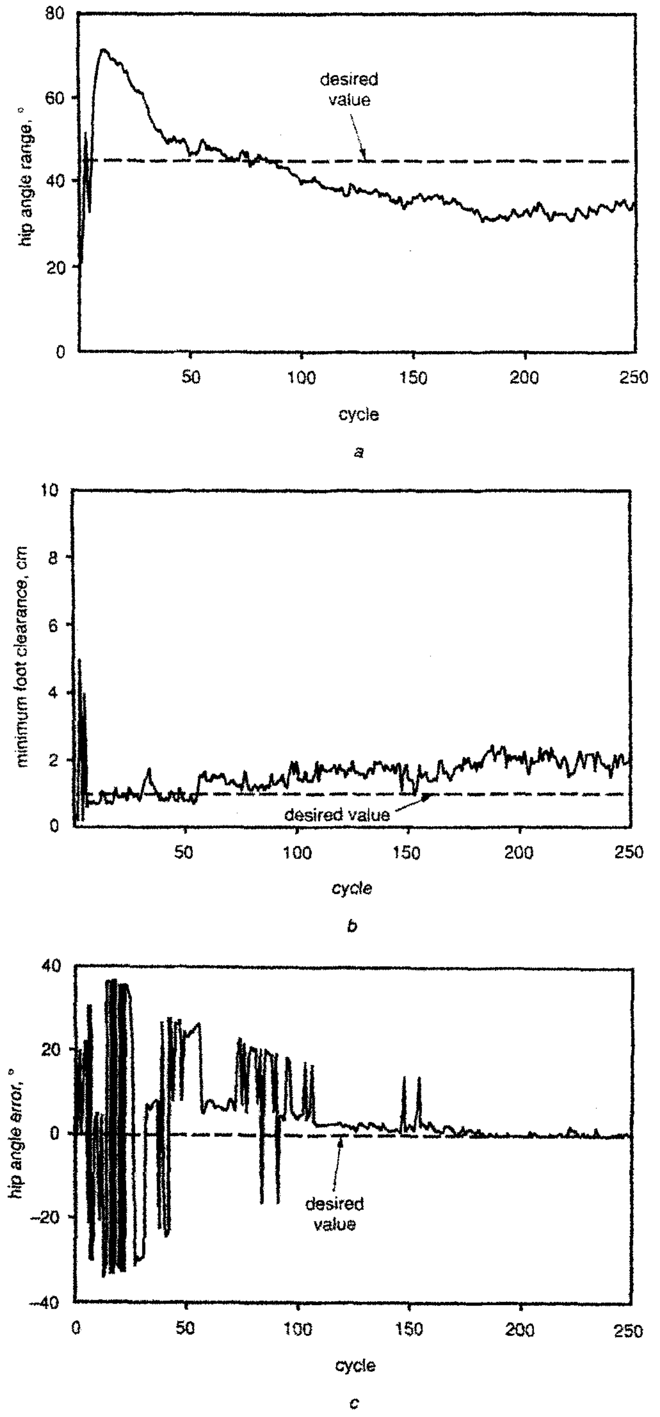


Fig. 11 *Swing phase parameter values per swing cycle obtained in open-loop controlled stimulation trial with optimised stimulation timing for three muscle groups in subject A; (a) hip angle range; (b) minimum foot clearance; (c) hip angle error at knee extension; when knee extension occurs too early, sign of hip angle error is negative; when knee extension occurs too late, sign is positive (Fig. 3)*

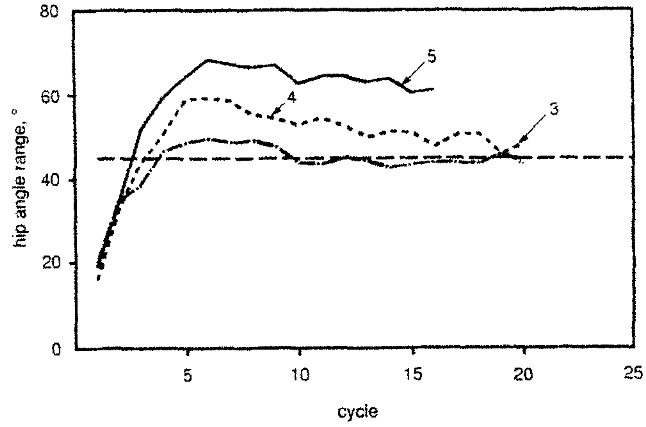


Fig. 12 *Step responses in hip angle range in cycle-to-cycle controlled stimulation trial in subject A, with gain G of PID controller equal to 3, 4 and 5 ms/°; hip flexor stimulation burst duraction adapted from cycle to cycle; controller output initialised to 100 ms to start first cycle; onset of hip flexors remained unchanged; hamstrings and quadriceps not stimulated; each step response started with leg at rest*

4.3 *Tuning of PID controller*

The hip angle range varied significantly throughout the entire open-loop trial in subject A (Fig. 11). A large hip angle range complicated obtaining the knee extension at the end of the forward swing, because quadriceps stimulation was triggered on the hip angle passing through a threshold (Fig. 11). The knee extension was correct when the obtained hip angle range was close to desired. Subject A's foot clearance was satisfactory throughout the open-loop trial. As noted, the open-loop trial in subject B was severely influenced by spasm. Based on the open-loop results, we attempted to adapt the hip flexor stimulation burst duration from cycle to cycle in order to maintain a stable swinging motion of the leg with stationary swing phase objectives.

The desired hip angle ranges of the two subjects, used as a reference for the PID controller, were those obtained in the experimental optimisation. The second zero z_2 (eqn. 1) of the PID controller, i.e. the pole in eqn. 4, was estimated at 0·7 for both subjects. The gain G was estimated to be 4 ms/° for both subjects, on the basis of examining step responses. Fig. 12 depictes typical step responses in subject A. The step response with the gain G of the PID controller equal to 4 ms/° had the desired overshoot. The settling time with G equal to 4 ms/° was, however, rather long, which was probably caused by the simultaneous time variation in the mechanical output of the hip flexors due to potentiation.

4.4 *Cycle-to-cycle control*

4.4.1 *Hip angle range*: Figs. 13*a* and 14*a* and *b* display the hip angle ranges and the burst durations applied to the hip flexors. The initial burst duration for the hip flexors followed from the experimental optimisation. Evidently, the desired hip range could not be maintained in subject B, although the hip flexor burst duration was incremented rapidly (Fig. 13*a*), showing the limited control space in subject B. In subject A (Figs. 14*a* and *b*), the PID controller initially decremented the hip flexor burst duration, to compensate for the effects of potentiation in the hip flexors. It then gradually incremented to compensate for the effects of hip flexor muscle fatigue. In the subsequent trial (Fig. 14*b*), the effect of muscle fatigue became more notable, resulting in a fast increment of the computed burst duration after approximately 100 swing cycles.

4.4.2 *Minimum foot clearance*: Figs. 13*b* and 14*c* and *d* depict the foot clearance during the cycle-to-cycle control of the hip flexors. Owing to spasm in the lower extremity muscles, stiffening the knee joint in full extension, subject B's foot clearance was again negligible. Subject A's foot clearance (Figs. 14*c* and *c*) was sufficient and approximately stable thoughout both trials.

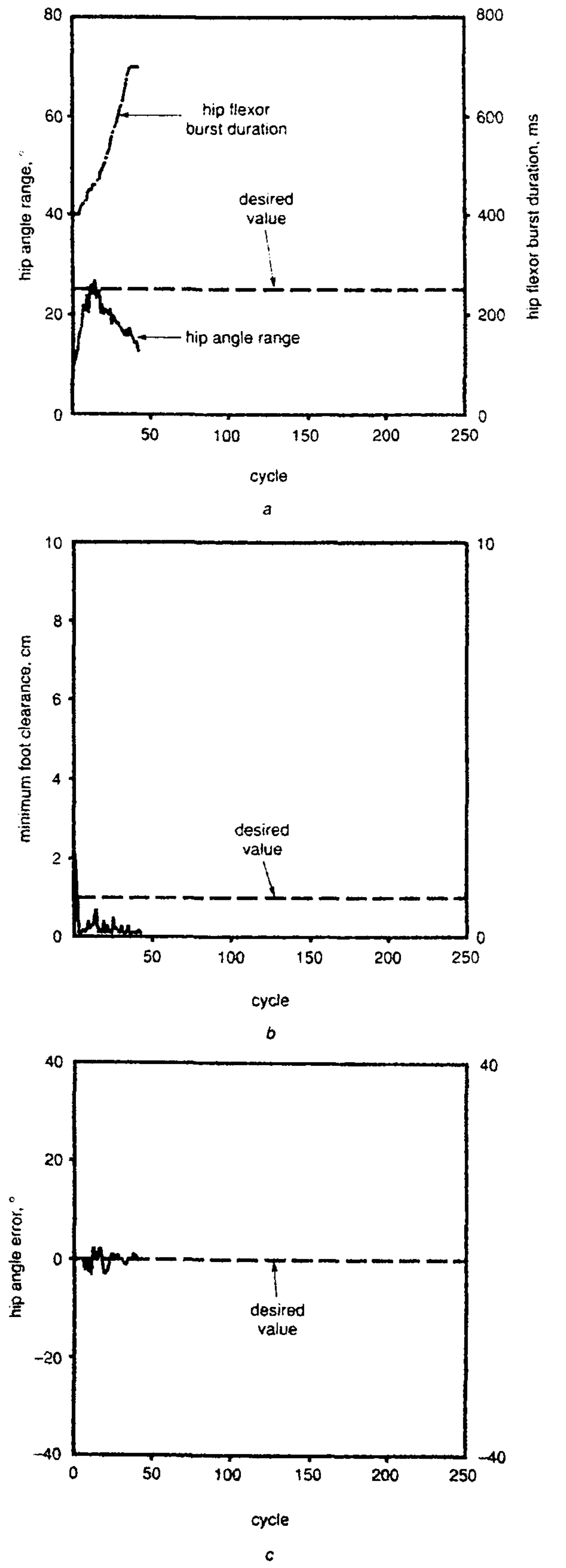


Fig. 13 *Swing phase objectives per swing cycle obtained in cycle-to-cycle controlled stimulation trial in subject B; hip flexor stimulation burst duration adapted from cycle to cycle; onset remained unchanged; stimulation onset and burst duration for hamstrings and quadriceps identical to open-loop experiment; (a) hip angle range and hip flexor burst duration; (b) minimum foot clearance; (c) hip angle error at knee extension; when knee extension occurs too early, sign of hip angle error is negative; when knee extension occurs too late, sign is positive (Fig. 3)*

4.4.3 *Knee extension*: as spasm stiffened subject B's knee in full extension, knee extension was always obtained at the maximum hip angle. In subject A (Figs 14*e* and *f*), knee extension occurred at the end of the forward swing throughout both trials. Only during the initial part of both trials, when the swinging motion had not stabilised, knee extension occurred too late.

5 Discussion

FES has the potential to develop into clinically feasible systems for the restoration of lower extremity functions in paraplegics (KRALJ *et al.*, 1983; PETROFSKY, *et al.*, 1986; MARSOLAIS and KOBETIC, 1987; CHIZECK *et al.*, 1988, ANDREWS *et al.*, 1989; GRAUPE, 1989; SOLOMONOW *et al.*, 1989; PECKHAM, 1989). Kralj *et al.* demonstrated ambulation assisted by parallel bars in selected paraplegics using multichannel surface FES with preprogrammed parameters. Marsolais and Kobetic and Chizeck *et al.* reported the achievement of reciprocal walking in 11 selected paraplegics using percutaneous electrodes with walker support or crutches. The stimulation sequences were again preprogrammed. Andrews *et al.* used sensory feedback to trigger preprogrammed stimulation sequences in predetermined phases of the gait cycle, described by a finite state model. Chizeck *et al.* also reported that such an approach is also under development in their laboratory.

The above systems do not, however contain strategies to adapt the stimulation sequences in case of muscle fatigue. Therefore, the systems demand preprogrammed stimulation patterns which provide a margin of safety in the case of FES-induced muscle fatigue or when unexpected disturbances perturb the system (MARSOLAIS and KOBETIC, 1987). Such patterns result in unnaturally exaggerated joint angles during the initial phase of a gait trial and very small joint angles (and step lengths) during the final phase. the results presented in this paper demonstrate that stimulation patterns for the major swing phase muscles can be derived so that swing phase objectives, characteristic of natural human gait, can be achieved in paraplegics. Moreover, slowly time-varying system characteristics, such as the effects of muscle fatigue and potentiation, can be compensated for by using a relatively simple cycle-to-cycle control strategy, successfully maintaining the desired swing phase objectives for a prolonged period of cyclic stimulation. When using the derived patterns during paraplegic gait, the influence of hip joint and voluntary upper body movement must be investigated. In the current study, we used a stand set-up to prevent these movements.

The PID controller could not adapt fast enough for the effect of potentiation at the initial part of the trials in subject A (Figs. 14*a* and *b*). Boom *et al.* (BOOM *et al.*, 1993) reported that the effect of potentiation on the mechanical output of intermittently stimulated muscle has a significantly smaller time constant than the effect of fatigue. The controller could thus have been tuned to react faster. Its initial output (hip fluxor burst duration) could also have been smaller. However, quadriceps stimulation at the end of the forward swing to obtain knee extension also contributed to a higher hip angle range. This was not compensated for by the PID controller, which only adapted the hip flexor stimulation.

The control space for hip flexor stimulation was determined by the mechanical output of the hip flexor muscles. We found this control space to be remarkably different for the two subjects. Training can enlarge the maximal mechanical output of the stimulated muscle groups, thus ensuring a reasonable control space to adapt for muscle fatigue. The extensive sessions of training,

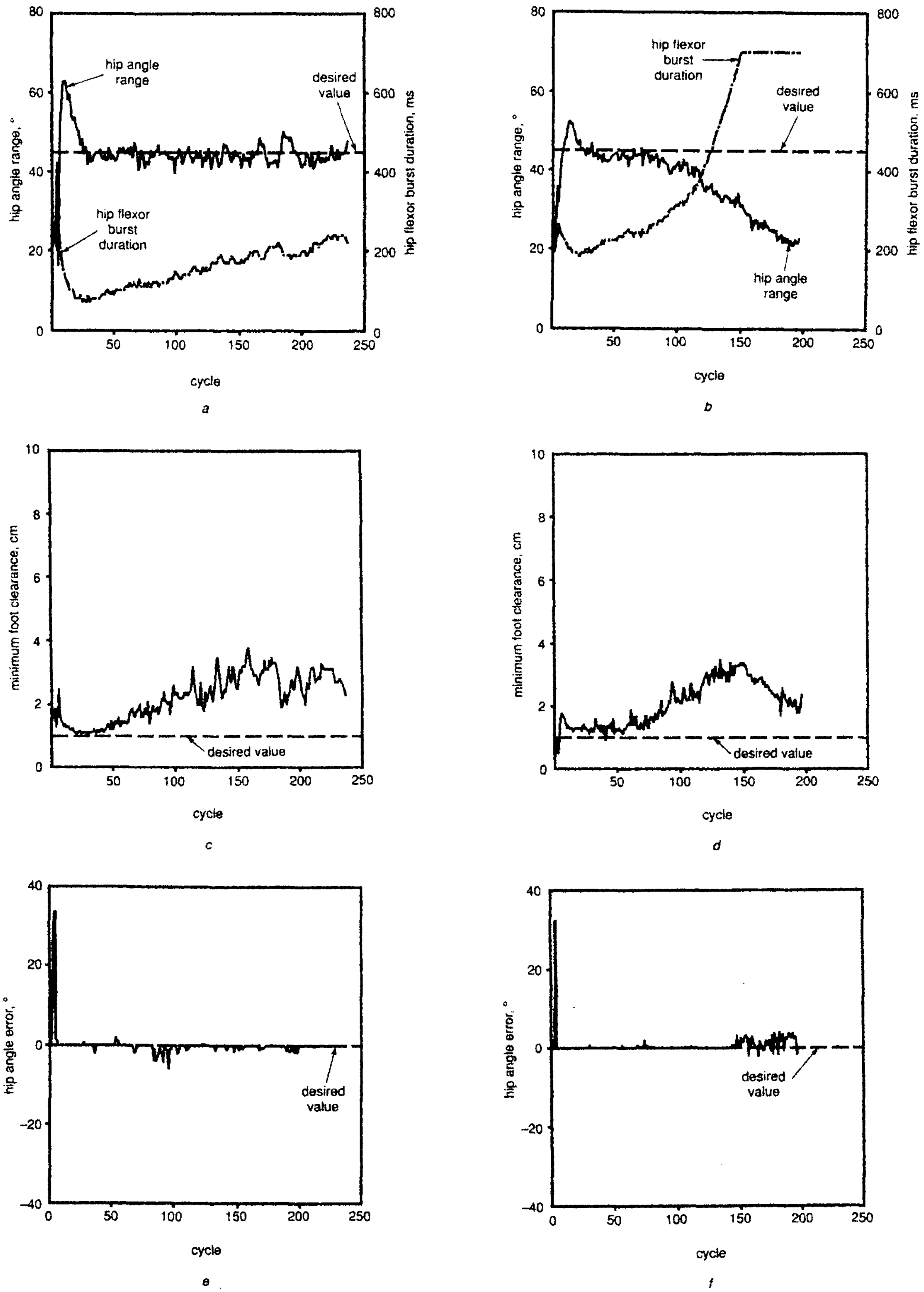


Fig. 14 *Swing phase objectives per swing cycle obtained in two cycle-to-cycle controlled stimulation trials in subject A; trail 2 performed directly after trial 1; hip flexor stimulation burst duration adapted from cycle to cycle; onset remained unchanged; stimulation onset and burst duration for hamstrings and quadriceps identical to open-loop experiment; when knee extension occurs too early, sign of hip angle error is negative; when knee extension occurs too late, sign is positive (Fig. 3); (a) hip angle range and hip flexor burst duration of trial 1; (b) hip angle range and hip flexor burst duration of trial 2; (c) minimum foot clearance of trial 1; (d) minimum foot clearance of trial 2; (e) hip angle error at knee extension of trial 1; (f) hip angle error at knee extension of trial 2*

however, are a significant burden on the participating paraplegic subjects. For this reason, the number of participating subjects was small, which limited the general applicability of the achieved results.

The stimulation frequency employed in the current study for the hip flexors was significantly higher than that used in FES-induces paraplegic standing and ambulation feasibility studies. Whereas a typical stimulation frequency between 20 and 30 Hz has appeared sufficient to generate movement (KRALJ *et al.*, 1983; PETROFSKY *et al.*, 1986; MARSOALIS and KOBETIC, 1987; CHIZECK *et al.*, 1988, ANDREWS, *et al.*, 1989; DURFEE and MACLEAN, 1989; GRAUPE *et al.*, 1989), in the current study the hip flexors were stimulated at 50 Hz. This was chosen to ensure a reasonable control space, in terms of burst duration, in case of muscle fatigue. Hamstrings and quadriceps were stimulated at 25 Hz, comparable to previously published studies. Recruitment modulation appeared insufficient as the control parameter, as the desired swing phase objectives could not be obtained at partial recruitment levels. These controls space limitations are typical for surface FES, mainly due to poor muscle selectivity and accessibility and changing electrode position during contraction. Percutaneous and implanted systems will have better muscle selectivity and accessibility and a higher total muscle torque output (MARSOLAIS and KOBETIC, 1987). However, dislocation, infection, and breakage, especially near the hip joint where muscle selectivity and accessibility improvement are most vital, still limit the robust applicability of these systems.

Our results show that rate of fatigue in the hip flexors to be much higher than in the hamstrings and quadriceps. To maintain the desired swing characteristics, the hip flexor burst duration needed to be incremented in time, whereas the mechanical output of the quadriceps and hamstrings was relatively time-invariant. The condition of hip flexors thus determined how long the swinging motion could be performed. This higher rate of fatigue can be caused by a number of reasons. The hip flexors had to initiate the movement of the entire leg, and the hamstrings and quadriceps mainly acted on the lower leg. In addition the hip flexors were stimulated at a higher stimulation frequency than the quadriceps and hamstrings, to ensure a reasonable control space for the hip angle range, which induces fatigue at a higher rate (FRANKEN *et al.*, 1993). Finally, the hip flexors were probably less well trained.

Parameterisation of FES-induced paraplegic gait on the level of a cycle facilitates adaptation of the applied stimulation from cycle to cycle on the basis of the deviation between desired and obtained functional objectives. Slowly time-varying system characteristics, caused by muscle fatigue for example, can be then adapted for using this methodology. However, several questions on the optimisation and control of stimulation remain.

(*a*) How can biarticular muscles be used in an optimal way?
(*b*) What are the optimal adaptation rules for stimulation onsets and burst durations for individually fatiguing muscles?
(*c*) Can unexpected disturbances be compensated for, or should the patient himself undertake intervention in such cases?
(*d*) How can the voluntary actions of the patient be modelled?

Active hip flexion is a prerequisite for a successful swing phase in natural human gait (INMAN *et al.*, 1981). In FES-induced paraplegic gait using surface electrodes, active hip flexion is usually generated afferently, by means of flexion withdrawal reflex (FWR) activation (KRALJ *et al.*, 1983, ANDREWS *et al.*, 1989; GRAUPE, 1989; GRANAT *et al.*, 1991). However, this reflex habituates to the applied stimulation, resulting in unpredictable and consequently poorly controllable hip flexion (GRANAT *et al.*, 1991). Solomonow *et al.* (SOLOMONOW *et al.*, 1989) reported the use of a reciprocating gait orthosis (RGO), which transfers hip extension in the standing leg to contralateral hip flexion, successfully avoiding the problems of FWR habitation. Chong *et al.* (CHONG *et al.*, 1989) reorted the feasibility of efferent electrical stimulatio of paralysed hip flexors under isometric condition using surface electrodes, placed over various motor point positions of muscles known to generate hip flexion. They concluded that sufficient hip torques could be generated to assist the swing phase of paraplegic gait. The results achieved in this paper show that sufficient hip flexion could be obtained using efferent hip flexor (surface) stimulation, and also reveal its controllability.

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