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Swing-through gait with free-knees produced by surface functional electrical stimulation

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The swing-through gait is often the gait of choice for those crutch walkers who can perform it. However, a practical (sufficiently low energy and sufficiently fast) gait is usually not achievable by paraplegic individuals with thoracic lesions. Functional electrical stimulation (FES) was used to assist three spinal cord injured (SCI) subjects with complete thoracic lesions at T11, T11 and T6 to ambulate with a swing-through gait patten. Eight channels of surface stimulation were used to bilaterally stimulate knee extensors, knee flexors, hip extensors and hip flexors. The stimulation sequence was controlled by a computer that implemented a finite-state, rule-based control strategy according to sensor inputs. Over a long, level walkway, the T11 subjects averaged 0.40 m/s and 0.38 m/s for distances of 56 m and 51 m; the T6 subject averaged 0.30 m/s for 43 m. Using a motion analysis system, the gait patterns of two of the subjects were compared to those of a trained, non-impaired subject. The SCI subjects spent more time in both double support phases (when both crutches and both feet contact the floor) than did the non-impaired subject, leading to a loss of momentum and hence a slower and less efficient gait. In conclusion, an FES assisted swing-through gait is shown to be a potentially useful mode of FES gait.

Keywords: electrical stimulation; gait; paraplegia

Introduction

Despite environmental modifications such as improved wheelchair access to buildings, the inability to stand and walk often constitutes a severe handicap for a mid-thoracic spinal cord injured person who wishes to play an active part in his/her community. The technique of using functional electrical stimulation (FES) to reinstate the lost neural input to paralysed muscle has been available for some time. A relatively fast and efficient gait can be obtained by the use of many channels of percutaneous or implanted electrodes (eg speeds of up to 0.8 m/s have been reported;[1]) however, the invasive nature of this approach may be inappropriate for many patients. The alternative technique of using surface electrodes to stimulate neural structures is less invasive, but produces cruder control and thus for paraplegic subjects with a complete thoracic lesion the gait obtained is slow (typically 0.04 m/S to 0.35 m/s).[2,3] The reciprocal walking pattern usually adopted is quasi-static, with much of the cycle being spent in the double support phase and very little or no mechanical energy being conserved from one stride to another. This leads to a gait which has a high energy cost[4] and is better described as stepping rather than walking. The low speed and high energy cost of this gait in part explains its lack of widespread use outside research laboratories. In contrast, swing-through gait is dynamic and fast, it has been described as:

. . . the fastest and most useful gait pattern of the completely paralysed person using long-leg braces and crutches.[5]

. . . the fastest and most useful gait, though requiring skilled balance.[6]

. . . [it] offers the paraplegic patient the fastest and most graceful type of mobility. Clinical experience has shown that many paraplegics can be taught a skill of application and an economy of motion that make crutch walking very practical.[7]

There are many variations of this gait, depending on the subject's strength, skill, level of lesion and degree of orthotic support. The type of gait that can be performed by a paraplegic person who has no control at and distal to the hip, and who is using knee-ankle-foot-orthoses (KAFOs) will be described (see Figure 1). In this gait the crutches are placed on the ground in front of the body (the first period of double support), and weight is transferred on to them from

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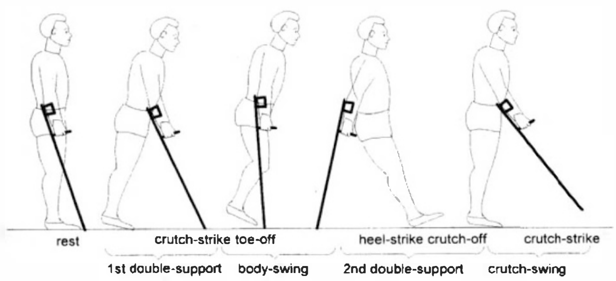


Figure 1 Phases and events in the swing-through gait cycle

the feet. The body is then lifted by depressing the shoulders and extending the elbows, raising the feet off the ground (toe off), and the body is allowed to swing through and beyond the crutches (body swing phase). The feet land in front of the crutches (heel strike) initiating the body stance phase with a second period of double support. The body has sufficient momentum to allow it to continue moving forwards, pivoting about the foot-ground contact point and passing through the vertical position (mid stance). During this period the crutches are lifted (crutch off) and brought to a position in front of the body (crutch strike). Having returned to the double support position, the ambulator can either stop, or can continue with the next stride whilst the body is still moving forwards, thus conserving some of the kinetic energy gained. The ratio of the time spent in both double support periods to the total stride time is called the double-support ratio.

Swing-through gait is characterised by lateral symmetry and stability, whereas reciprocal walking (as opposed to stepping) is unstable in both the anterior-posterior and medial-lateral planes.[5] This lateral stability reduces the number of muscles that need to be stimulated and controlled by preventing the need for ab- or adduction.

FES has been previously used to synthesise swing-through gait.[8,9] However, in these applications FES was used to maintain the knees braced in extension throughout the gait cycle (emulating the action of knee-ankle-foot orthoses). Braced knees require a greater energy input from the upper body to provide the extra lift required for foot clearance during the body-swing phase.[10] If the knees were actively flexed during the swing phase of the gait cycle, the work done by the upper body could be reduced, and the quadriceps could be temporarily rested. Thus if either upper body fatigue or quadriceps fatigue is a constraint on the duration of the gait, flexing the knees should increase the range of the gait.

The aim of the research programme described in this paper was to assess the feasibility of producing free-knee crutch assisted swing-through gait in spinal cord injured subjects with mid and low thoracic level lesions; thus providing an alternative, higher speed, gait mode to complement the FES reciprocal gait that many subjects working with us can already perform.

Methods

Subject selection
The selection criteria were as follows:

1 Mid- to low- thoracic, motor-complete lesion.
2 At least 2 years post-injury.
3 Lack of joint contracture.
4 Subject had previously undergone an FES strengthening programme for the quadriceps muscle group, and was able to stand using electrical stimulation and a walking aid for a period of 5 min.
5 It was possible to elicit a flexion reflex from one of the five sites suggested by Kralj *et al.*[11]
6 The subject was trained in the use of knee-ankle-foot orthoses for swing-to or swing-through gait (but did not necessarily have to walk regularly).
7 The subject was able to attend weekly training sessions.

Three subjects (A,B,C) were selected for training in swing-through gait; subject C withdrew from the programme (for personal reasons) before completing all the tests. The subject details (at the time of selection) are given below:

Subject A: 28 years, female, complete T11 lesion.
Subject B: 22 years, male, complete T6 lesion.
Subject C: 27 years, male, complete T11 lesion.

Subject training
All the subjects underwent a standardised training regime, supervised by a physiotherapist. Initially, a subject was trained to perform four point (reciprocal) FES augmented gait as described by Kralj *et al* [2] using a rolling walker as a support device. Once the subject was walking confidently, s/he proceeded to swing-to gait using mobile parallel bars, followed by swing-to gait in a rollator; once the subject was judged to be adept at this gait, the training proceeded to the use of crutches.

Initial crutch training involved the subject standing with crutches. Crutch height was adjusted for each subject, in order to produce the most stable standing position. The subject then advanced to taking single swing-through steps using crutches, and finally to continuous gait. Whenever a subject was standing, a nominated 'catcher' stood immediately behind him or her. Additionally, whenever a subject stood with crutches (which are less stable than a rolling walker), s/he wore an upper body harness (LecSave Ltd, Avon, UK) which was attached by a loop of 9 mm diameter mountaineering rope and a karabine (Wild Country Ltd, UK) (breaking force 24 000 N) to a specially developed mobile overhead support consisting of a pyramidal frame with self-locking castors attached to the base. These precautions ensured that when a subject fell, it was impossible for her/him to strike the ground. The karabiner was strain-gauged to determine if it was loaded (i.e. if the subject was using the harness to support her/his body-weight) during gait.

Stimulation strategies
All stimulation was performed using self-adhesive surface electrodes (Pals-Plus electrodes, Axelgaard manufacturing Co Ltd, Fallbrook, CA). The stimulator was an eight channel, current regulated, programmable device, capable of producing output pulses up to a maximum of 150 mA (described in Phillips *et al*).[12] The stimulator was controlled by an IBM PC compatible computer (Compaq Portable type II, Compaq, Houston, Texas) via a digital input/output board (PC-14A, Amplicon Ltd, Brighton, UK). Analogue data were sampled via a 12 bit analogue to digital (A/D) converter board (PC-26A, Amplicon Ltd, Brighton, UK). A *Turbo Pascal* (version 5, Borland International, Scotts Valley, CA) program controlled the stimulation parameters. The low-level routines that allowed this program to communicate with the stimulator and A/D board were provided by the *Turbo Pascal* unit *Stimdriv*.[13]

The following muscle groups were used for the production of swing-through gait:

- *Knee extensors* (quadriceps group): to prevent buckling of the knees during the body-stance period of the gait, and to produce knee extension during the late body-swing phase, immediately prior to heel-strike.
- *Knee flexors* (hamstrings group): to generate knee flexion during the early bodyswing phase of the gait, ensuring ground clearance.
- *Hip extensors* (gluteus maximus): to prevent buckling of the hip ('jack-knifing') during stance, especially at heel-strike.
- *Hip flexors* to actively flex the hip during body-swing, thus helping to produce ballistic knee flexion, and increase stride length. Due to the difficulty of recruiting the deep iliopsoas (hip flexing) muscle directly with surface stimulation, hip flexion was produced indirectly by the flexion reflex. The excitation of the flexion reflex also produced knee flexion, although this had to be augmented by direct stimulation of the knee flexors. For some subjects, at some sites, the inappropriate knee extension response described by Rudel *et al*[14] was observed. To avoid the bilateral inhibition effect reported by Granat *et al*[15] the onset of stimulation was delayed by 100 ms in one leg compared to the other.

The locations of the electrodes were as follows:

Electrode 1 Right flexor reflex: usually at the right peroneus superficialis site.[11] Small electrodes (circular with a diameter of 3.2 cm) were used. The active electrode was sited distally to the indifferent electrode, with a separation of approximately 5 cm. The precise location of the electrodes was determined by using a motor-point locator (a 2.5 cm diameter electrode that could be held and moved against the subject's skin to find the best stimulation site).

Electrode 3 Right quadriceps: the active electrode was placed over the estimated location of the femoral nerve near the top of the thigh, the indifferent electrode was located medially, roughly 5 cm superior to the patella. Both electrodes were large (circular with a diameter of 7.6 cm).

Electrode 5 Right hamstrings: the large active electrode was placed medially on the right posterior thigh, just inferior to the right buttock. The indifferent electrode (also large) was placed just superior to the right popliteal hollow, in an attempt to additionally recruit the gastrocnemius muscle. The exact position was determined by trial and error to maximise knee flexion and minimise hip extension.

Electrode 7 Right gluteus maximus: the large active electrode was placed near the right posterior-superior-iliac-spine. The large indifferent electrode was placed inferior to the buttock, just medial to the proximal hamstring electrode. Again, the exact position was determined by trial and error until the best visible and palpable contraction was obtained.

Electrodes 2, 4, 6, 8 were placed in similar, contra lateral positions to electrodes 1, 3, 5, 7 above. Each subject was fitted with polypropylene ankle-foot orthoses (AFOs), set in a neutral angle, which were worn inside their shoes. These prevented dorsi-flexion at the ankle during stance and foot-drop during swing. However, their use precluded the generation of push-off using active plantar-flexion.

Other equipment
The subjects used elbow crutches which had been strain-gauged to measure axial force. These crutches were calibrated using a force plate (Kistler Instruments Ltd, Hartley Witney, RG27 8RN, UK) at the start of each session. The crutch length was adjusted to the optimum previously found for each subject.

Two switches were used for manual control of the gait. One was attached to a crutch hand-grip, in a position which allowed it to be easily operated by the subject. The other was held by an experimenter, who could operate it if a subject preferred not to control the gait.

All gait trials were recorded on to video-tape (using a Panasonic NV-MS90 cam-corder recording onto a Panasonic NV FS90 HQ video cassette recorder); the computer output was superimposed on the picture at the time of recording by a 'gen-lock' card (EGA-lock, Vine Micros Ltd, UK). This facility allowed the active state of each stimulation channel, and the value of each sensor, to be overlaid onto the video picture.

The computer, amplifiers, camera, video-recorder and monitor were mounted on a specially designed mobile trolley which was wheeled alongside the ambulating subject.

Control of stimulation
The gait control program was written to allow simple development, execution and modification of finite-state controller strategies. All states were implemented as procedures which were written in a uniform manner. Transition between states was controlled by the main body of the program. The main body of the program also performed tasks such as updating the video output and processing sensor signals; thus minimising the extraneous code that had to be included in each procedure.

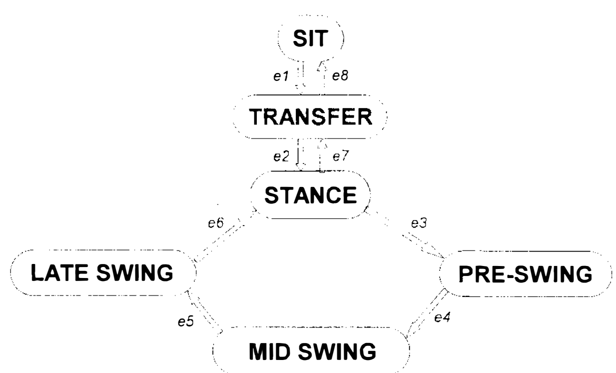
The finite state control strategy was initially written intuitively, based on the insights obtained from a review of the literature on the biomechanics of swing-through gait. It was then adapted over the course of the training period. Adaptation was an iterative process: the controller's performance was assessed from both the video recording and the subjects' comments, modifications were then made, and the controllers performance was re-assessed following the next session.

The final state transition diagram is shown in Figure 2.

Evaluation of the gait
The synthesised gaits were evaluated as follows:

Distance trails These tests were performed at Phillipshill Hospital, Glasgow in 1991, (the then location of the West of Scotland Spinal Injuries Unit).

The gait trials took place along a long, straight corridor with a level floor. Each subject walked along the corridor until s/he became sufficiently fatigued to need to stop. The subjects' heart rates were monitored during the tests using an ECG telemetry unit (sportsTester PE 3000, Polar-Electro Fitness Technology, Finland), to ensure that the test could be stopped immediately if the heart rate reached potentially dangerous levels. Distance travelled, time and number of strides were measured.



STATES	EVENTS
SIT: no stimulation	*e1*: investigator presses key
TRANSFER: **quads** ramp up or down	*e2*: time-out
STANCE: **gluts** and **quads**	*e3*: subject presses switch
PRE-SWING: **gluts** and **quads** and **reflex**	*e4*: both crutches loaded
MID-SWING: **reflex** and **hams**	*e5*: time-out
LATE-SWING: **quads** and **reflex**	*e6*: time-out
	e7: investigator presses key
	e8: time-out

quads = quadriceps stimulation
gluts = gluteal stimulation
reflex = flexion withdrawal reflex (usually peroneal site)
hams = hamstrings stimulation

Figure 2 State-transition diagram for swing-through gait

Single-stride trails These tests were performed in the gait laboratory of the Bioengineering Unit, University of Strathclyde. The intention was to study the kinetic and kinematic parameters of the gait in detail. The walkway was approximately 10 m long and consisted of level, smooth, linoleum tiles. All previous training had taken place along this walkway. For each subject, the kinematic parameters of the gait cycle were recorded by means of a TV based three-dimensional motion analysis apparatus (VICON VX system, Oxford Metrics Ltd, Oxford, UK).

Lightweight polystyrene spherical (25 mm diameter), retro-reflective markers were attached over the following landmarks.*

- 'Toe' (fifth metatarsal head)
- 'Heel' (inferior posterior aspect of shoe)
- 'Ankle' (lateral malleolus)
- 'Knee' (lateral epicondyle of femur)
- 'Hip' (greater trochanter)
- 'Shoulder' (acromion)
- 'Ear' (temporomandibular joint)
- 'Elbow' (lateral epicondyle of humerus)
- 'Hand' (lateral aspect of crutch hand-grip)
- Additionally, a marker was placed at the crutch tip.

The markers were attachment by double-sided adhesive tape. All markers were placed on the right side of the subject's body (the symmetrical nature of the gait permitted a unilateral analysis to be performed). The position of each segment was defined by the markers on its proximal and distal joints. This did not permit segment rotations to be measured, but was justified due to the sagittal planar nature of swing-through gait for all segments except the crutches and arms during crutch swing.[16]

Four cameras were positioned so as to allow any marker to be seen by at least two cameras in every TV frame (a necessary condition for the reconstruction of the 3-D co-ordinates). The marker positions were sampled at 50 Hz (the fastest rate allowed by the motion analysis system).

Each subject wore shorts and a 'T'-shirt. The markers were attached either directly to the subject's

*These markers positions are similar to those used by Wells,[10] with modifications to account for the use of elbow rather than axillary crutches.

skin (wherever possible, above bony prominences so as to minimise movement) or on to an orthosis (AFO, crutch) or shoe. Clothing was secured with adhesive tape to prevent it obscuring markers.

The subjects wore lightweight running shoes which were slightly over-sized to allow AFOs to be worn. They were securely laced.

Once all the markers were attached, an initial calibration test was performed. The subject stood (using FES stimulation and a rollator) on the force platform, which was in the centre of the measurement volume. An investigator standing on the motion analysis system's 'blind side' (ie the opposite side of the subject to the cameras) assisted the subject in maintaining a neutral stance (fully extended knees, neutral hip angle, upright trunk). The angles measured by the motion analysis system were used as the zero reference angles in subsequent analyses. The subject's weight was determined from the force platform output.

The subjects were then asked to proceed along the walkway at a self-selected ('comfortable') speed, using a swing-through gait. They typically achieved two complete strides before reaching the measurement area, the third, fourth and fifth strides being recorded. They continued for a further two strides before stopping.

Each subject performed a number of trials, separated by 5 min (seated) restbreaks, until fatigue precluded any further trials.

Only two of the subjects (A and B) were able to complete these tests, the third subject (C) having previously withdrawn from the programme. A trained, unimpaired subject (subject D) performed swing-through gait with braced ankles to serve as a comparison to the FES gait.

Results and discussion

The distance walked, average speed, average stride time and average stride length for each subject for the distance trials are given in Table 1.

The kinematic parameters derived from the stride by stride trials are given in Table 2.

No forces were detected in the strain-gauged karabiner of the safety harness during any of the gait trials, indicating that it was not used as a support during gait.

Table 1 Parameters obtained from distance trials for three spinal-cord injured subjects with complete thoracic lesions performing FES assisted swing-through gait

Subject (lesion level)	*Distance walked (m)*	*Time (s)*	*Mean speed (m/s)*	*Mean stride length (m)*	*Mean stride time (s)*
A (T11)	55.5	138	0.40	1.26	3.15
B (T6)	43.3	144	0.30	1.08	3.60
C (T11)	50.6	133	0.38	1.13	2.97

Table 2 Parameters obtained from single-stride trials for two spinal-cord injured subjects with complete thoracic lesions performing FES assisted swing-through gait and one trained, unimpaired subject performing swing-through gait

Subject and lesion	*Total strides*	*Overall speed (m/s)*	*Mean stride length (m) ±1 SD*	*Mean stride time (s) ±1 SD*	*Mean body-swing time (s) ±1 SD*	*Mean crutch swing time ±1 SD*	*Mean 1st double support time ±1 SD*	*Mean 2nd double support time ±1 SD*	*Double support ratio (%)*	*Mean maximum knee flexion (degree)*	*Mean maximum hip flexion (degree)*	*Mean distance beyond crutches (m)*
A (T11)	7	0.43	1.19	2.75	0.74	0.64	0.93	0.43	49.7	10.0	43.4	0.51
			±0.07	±0.33	±0.07	±0.06	±0.19	±0.17		±5.3	±2.3	±0.05
		(0.54)	(1.24)	(2.28)	(0.72)	(0.58)	(0.66)	(0.32)	(43.0)	(4.6)	(43.0)	(0.57)
B (T6)	8	0.35	1.13	3.24	0.71	0.49	1.22	0.82	63	50.9	42.8	0.25
			±0.05	±0.69	±0.15	±0.12	±0.59	±0.36		±6.9	±3.3	±0.07
		(0.45)	(1.11)	(2.48)	(0.68)	(0.64)	(0.76)	(0.4)	(46.8)	(49.7)	(45.3)	(0.24)
D (no lesion)	4	1.08	1.76	1.64	0.70	0.67	0.18	0.09	16.5	65.5	46.4	0.87
			±0.05	±0.06	±0.06	±0.02	±0.03	±0.01		±5.0	±2.8	±0.03
		(1.12)	(1.82)	(1.62)	(0.64)	(0.66)	(0.22)	(0.1)	(19.8)	(67.7)	(49.7)	(0.92)

Figures in brackets are for the fastest single stride

Distance trails
The speeds and distances attained during the distance walking tests are lower than the values of 1.0–1.5 m/s for 150 m and 1.0 m/s for 2000 m suggested by Marsolais and Kobetic[17] as being necessary for an acceptable gait, or the minimum speed of 0.5 m/s required for community walking.[18] However, they are probably adequate for a paraplegic person who uses a wheelchair as her/his main form of locomotion, but occasionally needs to cover short distances for exercise or when use of a wheelchair is inappropriate.

The maximum distances were obtained for continuous walking. If the subject was to use 'hybrid' system,[19] incorporating floor reaction orthoses (FROs), they would be able to take regular rest-stops without needing stimulation (the FROs would passively extend the subject's knees, resting the quadriceps muscle group; the subject's hips could be maintained in extension by the adoption of a 'C' posture, resting gluteus maximus). In this way, the onset of both local muscular fatigue and systemic fatigue could be deferred.[20] This would increase the range of the gait but decrease the average speed.

The mid-thoracic (T6) paraplegic walked for a shorter distance and at a lower speed than the low-thoracic subjects (although he walked for a slightly longer time). This may be because the subject's reduced control of his torso necessitated more use of his upper limbs for stabilisation, leading to a less confident and more tiring gait.

All three subjects achieved average stride lengths in excess of one metre.

Stride by stride analysis

Temporal parameters More insight can be gained into the results by examining the inter-stride variabilities of the temporal and kinematic parameters.

The speeds of subjects A and B performing FES swing through gait (Table 2) were similar to, but slightly higher than, the speeds measured over the longer distance trials (Table 1). Subject B (T6 lesion) had a similar stride length for FES swing-through gait to that of subject A (T11 lesion), but a longer stride time, which led to a lower speed. This longer stride time was a result of the subject spending more time in both double support periods. This can be explained by the lower trunk stability of this subject compared to the T11 subject. What is particularly noticeable is that there was a long second double support time, followed by a short crutch swing time. The subject did not seem to 'trust' his body to move as a rigid single link that pivoted about his feet. Instead, he paused in the second double support phase (losing forward momentum) to ensure that he was stable, then quickly threw himself forwards, bringing his crutches through and planting them as quickly as possible.

Part of the reason for the long second double support period may be that the subject's hip extensors were activated at a fixed delay after the initiation of swing, and this delay might have been too long. A better control strategy might have activated them at a more appropriate time, providing hip stability immediately on heel-strike; this would have allowed a shorter second double support time (although it may have been at the price of reducing hip flexion during swing, and thus reducing stride length).

The first double stance period might also have been shortened by a better control strategy. The subject was required to explicitly press a switch to initiate the body-swing phase; if instead, the subject's intention to initiate swing was detected from his/her posture and preparatory movements, then the timing and execution of the swing phase might have been improved.

The non-impaired subject demonstrated higher speeds for the two gait modes than both of the paraplegic subjects. This was due both to his longer stride length and to his shorter stride time. The shorter stride time resulted from a shorter period spent in both double support phases. What is noticeable about the non-impaired gait is the smaller inter-stride variability, indicating a higher degree of skill in crutch and foot positioning. This illustrates the great advantages of proprioception, sensation, and full muscular control in producing a consistent gait.

These results concur with those of Wells[10] who also reported (for non- and artificially impaired subjects) a decrease in double support time with increasing speed, and an increase with increasing disablement.

The body-swing times were similar for all subjects; the crutch-swing (bodystance) times were also similar (with the exception of the T6 paraplegic subject discussed above). This reflects the pendular nature of both swing phases, and suggests that a decrease in stride time (and hence an increase in speed) should be achieved by minimising the double support times.

The double support phases do not contribute to forward progression, and extra time spent in them leads to loss of forward momentum and kinetic energy. Their duration may be reduced by training, providing extra-stability, providing 'artificial proprioception'† and improving control strategies. This may be the best way to increase the speed and reduce the energy cost of FES swing-through gait.

The fastest single strides give an indication of the potential speeds of the gait. As expected from the previous discussion, the increase in speed for each gait mode for both paraplegics corresponded to a reduction in double support ratio. In the non-impaired subject, the (smaller) increase in speed was mainly due to longer strides, and corresponded to a slight increase in double support ratio.

†This can be achieved by 'sensory feedback' – feeding back joint positions or contact forces to areas of the body with preserved sensation, by means of sound, vibration, vision or electrical stimulation.[20]

Distance parameters The mean stride length of the non-impaired subject was greater than that for the paraplegic ambulators. In particular, the distance that this subject's feet landed in front of the crutches was greater than that for the T11 paraplegic subject which itself was greater than that for the T6 paraplegic subject. Sufficient kinetic energy is required at heel-strike to enable the body to pivot about the feet and pass through a vertical (maximum potential energy) position. The further the feet land in front of the crutches, the more energy, and hence the more speed, is required to reach this position. Good trunk and hip extension are also required to ensure that the body acts as a rigid inverted pendulum and does not buckle at heel strike.

Angular parameters At low angles of knee flexion, the downward inclination of the toes actually reduces ground clearance. Calculations show that the minimum knee flexion angle for increased ground clearance is approximately 45°. The mean, maximum angle of knee flexion occurring during swing for the T11 subject was only 10°, despite simultaneous stimulation of hamstrings and flexor reflex. This angle would have slightly decreased the ground clearance by inclining the toe further downwards. Further observation showed that this subject displayed a strong reflex resistance to flexion of her leg when this immediately followed the termination of a period of quadriceps stimulation. The subject had discontinued taking *Baclofen* approximately one year previously, and this may explain her heightened reflex activity. Solutions to this problem may involve re-administration of the drug, or preferably, using an alternative site of stimulation for the flexion reflex that will inhibit the unwanted knee-extensor activity‡

The T6 paraplegic subject (who was taking an anti-spasmodic drug) demonstrated a larger knee flexion angle of 51°. However, this is only slightly greater than the minimum value necessary to provide additional ground clearance and so the extra clearance gained will be small (approximately 1 cm) and must be weighed against the reduced security of a gait without permanently extended knees. However, if the subject's AFOs were set in dorsi-flexion, more ground clearance would result from knee flexion. Selection of an alternative site for application of the flexion-reflex stimulation may also have improved this subject's angle of knee flexion.

The knee flexion angle produced by the non-impaired subject was higher than that of two paraplegic subjects. The mean of 65° would produce ground clearance of approximately 4 cm based on the subject's height.

‡There is some evidence that inappropriate quadriceps contraction may be moderated by choosing different flexion-reflex stimulation sites (D Rudel, unpublished work at Ljubljana University and Rehabilitation Institute, Slovenia, and The Bioengineering Unit, Strathclyde University).

A further benefit of active knee flexion during the swing phase of gait is that it will reduce the moment of inertia of the swinging leg, and thus reduce the required hip flexion moment (or more importantly for FES gait, where producing hip flexion is difficult, increase the hip flexion angle resulting from a given hip flexing moment).

The maximum hip flexion angles of both paraplegic subjects were close to those of the non-impaired subject. Good hip flexion allows a longer stride to be taken, and can thus improve the speed of the gait.

Other aspects of the gait

It was apparent from early trials that the production of good body-stance-phase hip extension was vital for effective swing-through gait. This hip extension was successfully produced by bilateral stimulation of gluteus maximus; the placement of the electrodes also probably recruited some of the hamstrings group, which further helped hip extension. However, there were some problems associated with this stimulation site: firstly, electrode positioning was critical: a slightly misplaced electrode was either ineffective, or worse still, (in subject A) sometimes elicited a flexion response. Secondly, it was difficult for a paraplegic subject to apply gluteal electrodes independently, so she could not train these muscles at home. As a consequence, the muscle groups responsible for hip extension fatigued rapidly, and their endurance was one of the limiting factors (together with systemic fatigue) in the duration of an experimental session.

The T6 paraplegic subject would probably also have benefited from some trunk stabilisation, which could have been provided by stimulation of the erector spinae or quadratus lumborum groups (however, this was precluded as all eight available channels of stimulation were already being utilised) or by the fitting of a lumbar brace.

The overhead support and harness were very important in raising the confidence of the subjects sufficiently for them to attempt this (initially precarious) gait. During the early training sessions they often stumbled, but falls were always arrested by the support. The system did not seem to impede the gait in any way.

Finally, the opinions of the paraplegic subjects are pertinent. They accepted that the gait was faster than either KAFO or 4-point FES ambulation, but expressed the view that they would not use it for community walking, due to the unnatural gait style. This contrasts with the opinions of many commentators on gait in spinal cord injury, who report that (KAFO) swing-through gait is often the gait of choice.[5-7] The explanation for this disparity is probably that those paraplegics who can successfully perform KAFO swing-through gait have lower lumbar lesions, which allow them to produce a much faster gait (Rovick and Childress[21] report a swing-through speed of 0.9 m/s in one paraplegic ambulator). These faster speeds make the gait a practical alternative to a wheelchair for community ambulation, and thus compensate for any lack of cosmesis. At the much lower speeds (and ranges) achieved in this study, any advantages over the use of a wheelchair are not sufficient to offset the unnaturalness of the gait and the time required to apply the system. It is hoped that use of a permanently implanted system would eliminate preparation time, and if this were combined with more advanced control strategies to increase speed, the cost/benefit balance for this form of gait would be improved sufficiently to make it practical.

Conclusion

To the authors' knowledge, this is the first demonstration of FES assisted swing-through gait with free knees. We have shown that it is possible to produce a relatively fast, dynamic swing-through gait in the laboratory using surface FES.

Future work

In order to make this gait practical outside the laboratory the following points must be addressed:

1 *Knee flexion*: the angle of knee flexion during mid swing must be increased, investigation of alternative sites and stimulation strategies for the flexion withdrawal response, possibly augmented by direct stimulation of hamstrings may solve this problem.
2 *Hip and trunk stability*: the stability of hip and trunk seems crucial to the confidence and speed of the paraplegic ambulator. This may be addressed by use of a stretch electrode garment which allows gluteal electrodes to be easily stimulated and hence trained for greater fatigue resistance. Orthotic solutions such as lumbar/sacral corsets should also be considered, especially for the mid-thoracic injuries.
3 *Co-ordination with subject's intention to move*: in a gait as dynamic as this it is essential to co-ordinate FES induced movements with those of the subject's voluntarily controlled musculature. We are investigating machine learning techniques that detect the subject's intention to move by his preparatory posture changes, we will use these signals to automatically trigger the appropriate stimulation state.
4 *Better sensors and control strategies*: the present system defines many state transitions by fixed time delays, this is not appropriate for non-laboratory gait in which the environment is not well defined. It will be necessary to use a closed loop system incorporating robust sensors for an orthosis that is suitable for community ambulation.
5 *Finer motor control*: it may be necessary to move towards a percutaneous or fully implanted system in order to achieve the precise motor control required for a safe, dynamic and efficient gait. This would also have the advantage of reducing the donning and doffing time and avoid the use of the flexion reflex.

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