Development of an Isokinetic Functional Electrical Stimulation Cycle Ergometer

Ché Fornusek, MBioMedE*, ▪ Glen M Davis, PhD*, ▪ Peter J Sinclair, PhD†, ▪ Bruce Milthorpe, PhD‡

**Rehabilitation Research Center and †School of Exercise and Sports Science, Faculty of Health Sciences, University of Sydney, and ‡Graduate School of Biomedical Engineering, University of New South Wales, Sydney, Australia.*

▪ Abstract

An isokinetic functional electrical stimulation leg cycle ergometer (iFES-LCE) was developed for individuals with spinal cord injury (SCI). The iFES-LCE was designed to allow cycle training over a broad range of pedalling cadences (5–60 rev/min) to promote both muscular strength and cardiorespiratory fitness. A commercially available motorized cycle ergometer was integrated with a custom built FES system, a laptop computer, and a specialized chair that restricted lateral leg movements. Sample biomechanical data were collected from an SCI subject performing FES cycling to demonstrate the iFES-LCE's performance characteristics. Calibration of the iFES-LCE system revealed a linear relationship between torque applied to the axle of the motorized ergometer and the braking motor current generated to maintain velocity. Performance data derived from iFES-LCE motor torque agreed closely with similar data collected using strain-gauge instrumented pedals (cross-correlations = 0.93–0.98). The iFES-LCE was shown to work well across a range of pedaling cadences. We conclude that the new iFES-LCE system may offer improved training potential by allowing cycling over a broad range of pedaling cadences, especially low cadence. This device also improves upon the accuracy of other ergometers by adjusting for the passive load of the legs. ▪

Address correspondence and reprint requests to: Ché Fornusek, MBioMedE, Rehabilitation Research Center, Faculty of Health Sciences, University of Sydney, PO Box 170, Lidcombe, NSW 2141, Australia. Email: cfornuse@mail.usyd.edu.au.

Neuromodulation, Volume 7, Number 1, 2004 56–64

Key Words: FES, isokinetic exercise, paraplegia.

INTRODUCTION

Spinal cord injury (SCI) frequently induces secondary degenerative changes including loss of cardiovascular fitness, muscle atrophy, osteoporosis, and poor circulation in the affected limbs (1–3). Functional Electrical Stimulation (FES) cycling, whereby computer-controlled electrical stimulation elicits leg muscle contractions in an appropriate sequence to produce a cycling motion, has been successfully used with SCI to minimize some of these degenerative changes. Increases in muscle mass and FES-evoked strength or endurance have been commonly found following training (4–7).

Despite these positive physiologic benefits, FES cycling exercise is not widely accepted as a necessary part of the ongoing health care and rehabilitation of the SCI individual. Broad acceptance is often impeded by the cost of equipment and the fact that, for many, the benefits do not outweigh the training time required with such exercise systems. In addition, muscle spasms, denervation, and the presence of residual sensation often prevent some individuals with SCI from undertaking FES exercise.

In the past, FES cycling ergometers (eg, ERGYS, Therapeutic Alliances, Inc., Fairborn, OH; StimMaster Electrologic of America, Dayton, OH) have been dominated by devices that rely upon an electrically braked flywheel to provide a resistive load. The flywheel's momentum smoothes out FES-evoked muscle contractions and reduces the deceleration during "dead spots", where muscle contractions are absent. In these systems, a computer controls stimulation amplitude to keep pedaling cadence near to 50 rev/min. When maximum stimulation output is reached (usually 140 mA) but is insufficient to maintain the target cadence, pedaling cadence slowly decreases. When pedaling cadence falls below a minimum threshold (~35 rev/min), then FES is terminated and cycling exercise stops. Controlling cycling cadence in this fashion becomes increasingly difficult when the desired target cadence is below 35 rev/min. Chen and colleagues (8) developed a model-free fuzzy-logic control system to improve the "smoothness" of the cycling motion, but this was only tested down to ~35 rev/min. Over the years, advances have been made on the electronic braking and the stimulation systems, but pedaling cadence is still controlled by an electrically braked flywheel.

FES cycle ergometers that use a motor to assist pedaling have been previously described (9–11), but little research has been undertaken into the physiologic responses to such exercise. When using motorized FES ergometers, care must be taken because the motor has the potential to injure the subject by forcing a limb movement in the presence of a strong muscle spasm. Recent technologic advances have lead to the development of specialized low-power motorized cycle ergometers with safety features that prevent muscle injury. Motorization also allows a larger proportion of SCI persons to undertake FES cycling, ie, those unable to generate and maintain sufficient muscle forces to rotate a large flywheel or in those with low tolerance to FES due to residual sensation.

PURPOSE AND DESIGN OBJECTIVES

A new type of FES cycle was conceived to overcome some of the limitations inherent in earlier systems. An isokinetic functional electrical stimulation leg cycling ergometer (iFES-LCE) was designed to allow isokinetic exercise over a wide range of pedaling cadences (5–60 rev/min) while providing accurate real-time feedback of muscle performance. The new iFES-LCE can be used to investigate the potential benefits of training at low cadence compared to the traditional training cadences of 35–50 rev/min. This paper describes the design and construction of the new isokinetic FES cycle ergometer as well as some potential applications for client benefits. Sample biomechanical performance data from a SCI individual has been included to illustrate some of the key features and potential advantages of the iFES-LCE system.

MATERIALS AND METHODS

IFES-LCE System Overview

The iFES-LCE comprised a motorized cycle ergometer, a chair with passive leg restraints (Fig. 1), a portable computer, and a muscle stimulator. Figure 2 portrays a block diagram of the iFES-LCE and its components. In brief, the cycle ergometer sends crank position, crank angular velocity, and motor current information to the portable computer. Using instantaneous crank position data, the computer program directs the muscle stimulator

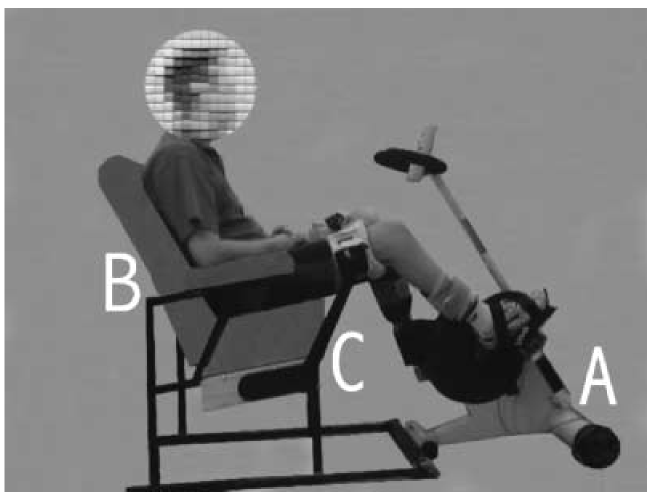

Fig. 1. Components of the iFES-LCE: A) motorized ergometer and B) chair with C) leg restraints. Muscle stimulator and laptop not shown.

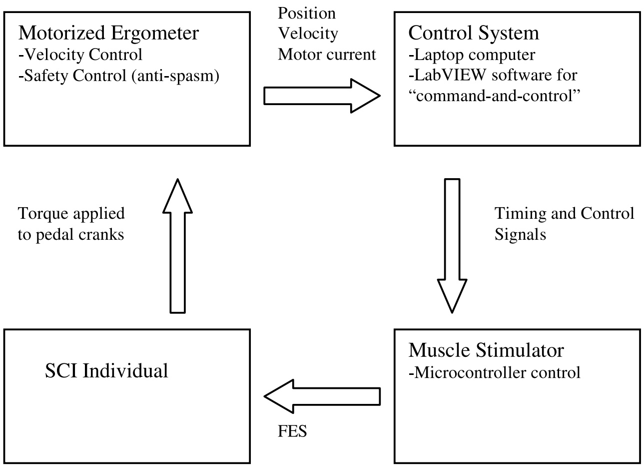

Fig. 2. Block diagram of components of the iFES-LCE with each component's basic function.

to provide stimulation to six muscle groups (left and right gluteals, hamstrings, and quadriceps muscles) at the appropriate times to elicit a smooth cycling pattern. By using the instantaneous motor current and angular velocity measurements collected over a single crank revolution, the computer calculates average power output for each revolution. Muscle stimulation amplitude can either be controlled manually, via feedback, or according to a preprogrammed time course. When the computer program is in "Feedback mode" it will attempt to alter FES-evoked muscle forces to meet a requisite power output by increasing or decreasing the stimulation amplitude. A preprogrammed stimulation amplitude is very useful when stimulation needs to be kept identical between testing sessions. Each of the components in the system is discussed subsequently in more detail.

Motorized Cycle Ergometer

A commercially available motorized cycle (Motomed Viva 1, Reck Medizintechnik GmBH, Betzenweiller, Germany), especially designed for low-intensity cycling exercise at cadences of up to 60 rev/min, was selected as a component of the iFES-LCE. In practice, any motorized cycle ergometer will suit this system as long as speed control circuitry within the motorized cycle can maintain a selected pedaling cadence by driving or braking the motor as required. The current from the motor reflects the torque required to oppose FES-evoked muscle contractions exerted by the subject or to assist pedaling during the "dead spots". Accurate measurements of exercise power output and instantaneous torque were calculated from the motor current of the cycle by subtracting the passive motor torque (ie, when the limbs were being moved solely by the motor) from the active torque.

The Motomed Viva (or similar) cycle possessed certain design features that made it a good candidate component of the iFES-LCE ergometer:

i) Torque limitation (spasm protection) circuitry. This circuitry restricts the amount of torque generated in the motor from going above preset safe levels.
ii) Motor and crank position data are easily available via RS-232 data transfer.
iii) It is compact and portable.
iv) It is relatively inexpensive.

The motorized cycle was calibrated by lowering and raising weights of known mass on a reel attached to the crank's axle, while a computer recorded motor current and axle velocity. The calibration procedure was performed across a range of cadences (10–60 rev/min) and torques up to 14 Nm. From the recorded data, a calibration equation was derived relating the torque applied to the axle to the recorded motor current.

Command-and-Control Computer Software

A PC laptop computer was used to control the stimulator hardware and record and display muscle performance data. The parameters of muscle stimulation (eg, stimulation amplitude and timing) were controlled by computer software written in G-language (LabVIEW 4.1, National Instruments Corporation, Austin, TX). The program received data from the cycle (motor current, crank position, and crank velocity) through the RS-232 serial port. Instantaneous crank position was used to control the initiation and duration of FES-evoked muscle contractions while motor current and velocity were used to calculate power output and torque being generated by the subject. The laptop computer controlled the muscle stimulator (DS2000) via its parallel port. The control program contained safety features to stop muscle stimulation if the crank velocity dropped to zero or the serial data transfer was interrupted for any reason.

Table 1. Characteristics of the iFES-LCE and available commercial FES cycle ergometers[a].

	ERGYS 2[1]	Stim Master Galaxy[2]	iFES-LCE
Load Control	Braked Flywheel	Braked Flywheel	Motor
Operating Velocity	35–50 rev/min	35–50 rev/min	5–60 rev/min
Exercise Mode	Isotonic	Isotonic	Isokinetic
Power Calculation	Offset Zero	Offset Zero	True Zero
No Channels	6	6	6
Amplitude	0–140 mA*	0–140 mA*	0–140 mA*
Output Waveforms	Various biphasic	Various biphasic	Various biphasic
Stimulation Frequency	30–60 Hz	30–60 Hz	10–100 Hz
Pulse Width	200–500 μs	200–500 μs	50–500 μs

[a]Comparison of the characteristics shows that the stimulation parameters are similar but that the iFES-LCE possesses the ability to operate over a wider range of pedaling cadences due to instantaneous angular velocity being controlled by a motor.
[b]Therapeutic Alliances ERGYS·
[c]Electrologic of America StimMaster.
[d]Constant current.

Stimulator Hardware

A constant current (maximum 140 mA), battery powered, optically isolated, six-channel FES system (DS2000, University of Sydney) was used to deliver muscle stimulation. The stimulator was designed with flexibility so that it could be used for experimental investigations into optimizing FES stimulation parameters for exercise with SCI individuals.

The stimulator was controlled by a C program embedded on a PIC microcontroller (Microchip Technology, Mountainview, CA). The stimulator's software allowed the computer to control stimulation amplitude, stimulation on/off timing for each channel, pulse width, frequency, biphasic or monophasic pulses, and stimulus ramp time (Table 1). In the case of iFES-LCE, the control computer directed the stimulation timing based upon crank position and the stimulation amplitude to produce FES cycling at the desired intensity. The stimulator's software also performed safety checking that the correct stimulation amplitude was being delivered and that there were no open circuits in the stimulation path.

Instrumented Pedals

Although not a component of the iFES-LCE as deployed for SCI individuals wanting to exercise, instrumented pedals (12) were used to validate the accuracy of the system's calibration. Additionally, our instrumented pedals were not originally designed to measure the small forces elicited during FES-evoked cycling but were the best equipment available from our laboratory. The instrumented pedals, which were designed to measure able-bodied cycling torques, based upon the designer's specifications had absolute error bound of ±0.7 Nm on the current system. The pedals measured forces in the sagittal plane ("shear" and "normal" forces) and the pedal angle with respect to the crank. Calibration was performed by applying known weights to the pedals and producing a range of forces that exceeded those expected during FES-evoked cycling. Instrumented pedal data was filtered by a low pass analog filter then sampled with a data acquisition card (DAC1200, National Instruments) at 60 Hz.

The horizontal and vertical components of the forces acting on each pedal were calculated from the measured shear and normal pedal forces and the angle of the pedal. The torque applied to each crank was then calculated by using these global forces, the crank angle, and pedal angle. Zero degrees for the crank angle was defined as the position when the right crank was at top dead center (TDC). Total torque was calculated as the sum of torque from each crank.

Pilot Cycling Trials

The iFES-LCE has been used successfully to train SCI subjects in our laboratory. Data from a paraplegic SCI subject performing two different types of cycling trials were selected to demonstrate some facets of the performance of the iFES-LCE under "real world" conditions. For these trials, oval (8 cm × 13 cm), gel-backed, self-adhesive, skin-surface electrodes were used. FES parameters comprised monophasic rectangular pulses of 250 μs duration applied at a stimulation frequency of 35 Hz. The crank angles for muscle stimulation were 300°–30°, 60°–160°, and 6°–73°, respectively, for the Quadriceps, Hamstrings, and Gluteals.

The first trial involved FES cycling at 20 rev/min and compared torques measured from the instrumented pedals and the ergometer motor. Velocity performance was also examined using data measured from the ergometer. During the measurement period stimulation amplitude was held constant at 100% (140 mA). Measurements were made while only the quadriceps muscle group of one leg was stimulated, and also while all involved muscle groups of both legs were stimulated. Torque vs. angle (Fig. 3, top panel) and crank velocity vs. angle curves (Fig. 3, bottom panel) portrayed torque measurements and velocity regulation performance, respectively. Torque curves at 20 rev/min are shown for FES-LCE with only one muscle group (left quadriceps) and with bilateral cycling of both legs (Fig. 4). Data were ensemble-averaged over 12 s of ergometer and instrumented pedal data.

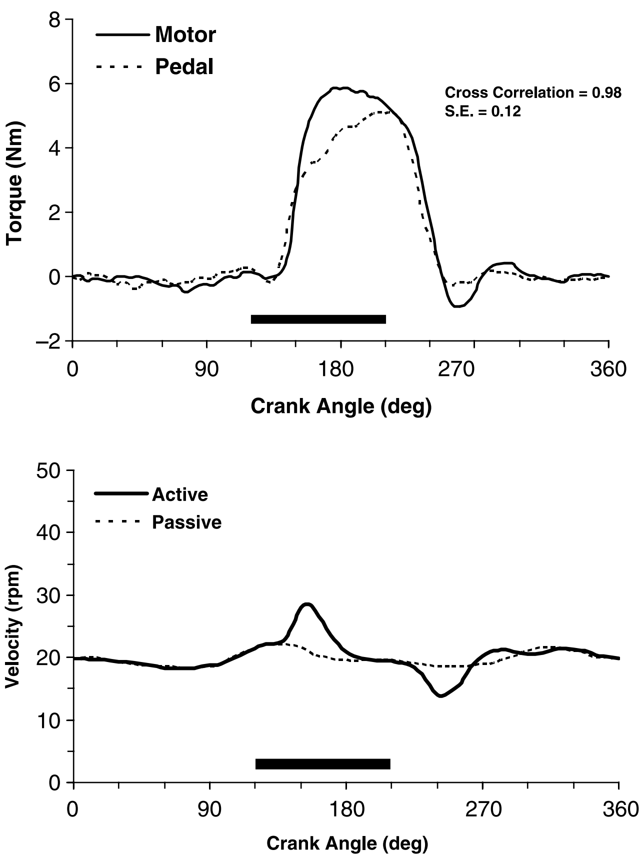

Fig. 3. Upper Panel: Comparison of torque-angle curves from the instrumented pedals and the iFES-LCE for one subject. Cycle cadence was 20 rev/min. Torques generated from stimulation of L. Quadriceps. Black bar indicates FES duration (duty cycle = 25%). Cross-correlation function calculated at lag = 0 with standard error. Lower Panel: iFES-LCE crank velocity regulation during torques generated by L. Quadriceps in upper panel. Cycle cadence was 20 rev/min. Black bar indicates FES duration (duty cycle = 25%).

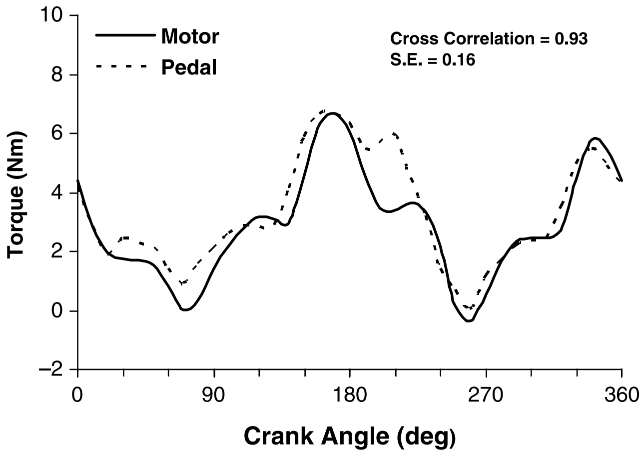

Fig. 4. Comparison of torque-angle curves from the instrumented pedals and the iFES-LCE for one subject. Cycle cadence was 20 rev/min. Torques generated from stimulation of R and L Quadriceps, Hamstrings, and Gluteal muscle groups. Cross-correlation function calculated at lag = 0 with standard error.

Linear regression was employed to derive the relationship between axle torque and motor current. Cross-correlations (lag 0) and descriptive statistics were used to assess the "goodness of fit" between torque data derived from strain-gauge instrumented pedals and torques calculated from the motor for a single muscle group and for all muscle groups during FES-evoked cycling. All statistical analyses (SPSS version 11.0, Chicago, IL) were considered significant at 5% confidence interval.

The second trial involved an SCI subject performing an incremental exercise test of increasing power output at a constant cadence of 40 rev/min. This trial was to demonstrate how effectively the ergometer controlled power output in "Feedback mode", as well as to demonstrate one application of the new ergometer. Four different power outputs were tested. The software was set to keep the power outputs between ranges of 3–4 W, 6–7 W, 9–10 W, and 12–13 W, respectively. Based upon whether the power output matched the "target workload" for the preceding revolution, stimulation was either increased/decreased 0.8% (~1.1mA) or left unchanged. For example, at the 3 W "target workload", if the power output fell below 3 W during a revolution the stimulation amplitude was increased, but if it was above 4 W the stimulation amplitude was decreased.

RESULTS

iFES-LCE Calibration

Calibration of the iFES-LCE revealed a consistent linear relationship ($R^2 > 0.999$, SEE (standardized error of estimate of regression) = 0.128 Nm) between the torque applied to the axle and the counteracting driving or braking motor current generated to maintain a constant angular velocity. The relationship is expressed in Equation 1, where current is a digital representation of motor current from the MOTOmed.

(1) $\text{Torque (Nm)} = -0.041 \cdot \text{Current} + 1.15$

Both parameters in the prediction equation were statistically significant ($p < 0.05$). The calibration equation was consistent across all cadences assessed over the range 10–60 rev/min. Further investigation of the data revealed that any small errors from the torque predictions appeared to be independent of actual torque or axle velocity.

iFES-LCE Crank Torque vs. Pedal Torque

The pilot cycling trials demonstrated a high cross-correlation between the crank torque calculated from motor current and that from the instrumented pedals. Figure 3 (top panel) displays a sample torque curve from the stimulation of the left quadriceps muscle group (RMSE 0.43 Nm, cross-correlation 0.98, $p < 0.05$). The start and stop angles of torque generation generally agreed well. However, the calculated motor torque was observed to overshoot the pedal torque data during the onset of torque generation and undershoot the pedal data after the end of the FES-evoked muscle contraction.

The Motomed Viva ergometer did not produce "true" isokinetic cycling, but the average velocity (cadence) maintained over one crank revolution was very stable (Fig. 3, bottom panel, passive cycling). The initial effect of an FES-evoked muscle contraction applied to the crank was a sharp increase in angular velocity. This increase was likely due to a delayed response from the ergometer's internal speed-control feedback circuitry. An increase in pedaling resistance quickly re-established the target velocity (Fig. 3 bottom panel, active cycling). When the FES-evoked contraction was terminated, the crank velocity dropped sharply and was eventually stabilized around the target velocity (20 rev/min) by a decrease in pedaling resistance that included an undershoot component and a damped oscillatory response. At higher cycling cadences (> 40 rev/min) the magnitude of the velocity perturbations due to a given torque disturbance were proportionally less, but the velocity settling time translated into a larger percentage of a crank revolution.

Torque curves from FES-evoked cycling with bilateral gluteal, quadriceps, and hamstring muscle groups being stimulated were more complicated (Fig. 4). The motor and instrumented pedal torque-angle curves followed approximately the same path (RMSE 0.81 Nm, cross-correlation significant 0.93, $p < 0.05$), but with some noticeable deviations. The speed-control circuitry of the ergometer attempted to keep the crank velocity constant (20 rev/min), but some oscillation of crank velocity was apparent.

Incremental Power Output Test

During the incremental test the feedback system maintained the power output well and the produced power output was close to the desired output. The power output and stimulation amplitude from the incremental test can be seen in Fig. 5. Power output (± RMS error) averaged 3.4 ± 0.2 W, 6.5 ± 0.3 W, 9.4 ± 0.4 W, and 12.4 ± 0.5 W, respectively, for the four different power levels tested.

DISCUSSION

This paper describes the design and validation of a new FES cycle ergometer of "isokinetic" type, which could be used as a tool for basic research into FES-evoked exercise, could be utilized for home use by SCI clients, and would be an improvement upon existing designs. The iFES-LCE fulfilled these design criteria well; the ergometer was useful

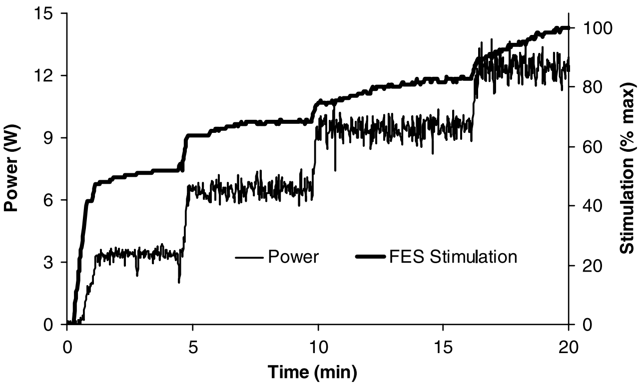

Fig. 5. Power outputs and FES stimulation amplitudes during the incremental exercise test performed at a constant cadence of 40 rev/min. Power output (W; left axis) and stimulation amplitude (% maximum; right axis) is displayed for each revolution. Note that at each of the "target workloads", a slight upward drift of stimulation amplitude implied the onset of muscle fatigue.

for research purposes due to its accuracy and FES parameter flexibility, while its size and cost (~$US8,000) also made it appropriate for home use. A comparison of the features of the iFES-LCE and existing commercial FES ergometers is shown in Table 1. The major difference between the iFES-LCE and the other two ergometers in the table is that it uses a motor. The motor's isokinetic speed control circuitry allows the iFES-LCE to operate accurately over a wide range of cadences and to calculate power output against a true zero. The stimulation parameters available on each system are quite similar.

The iFES-LCE provided good estimation of the torque levels generated during FES cycling. The linear regression established during the calibration of the system showed "goodness of fit" ($R^2 > 0.999$) and a small SEE leading to an accurate estimate of average and instantaneous torques produced over a wide range of pedaling cadences (10–60 rev/min). There were some discrepancies between the torque data from the ergometer and the instrumented pedals. These may have been due to a delayed response of the internal velocity-control circuitry, which responded with an apparent overshoot or undershoot in motor current (and therefore calculated instantaneous torque) during acceleration and deceleration of the crank. Considering the pedals' measurement tolerance and the fact that the pedals and ergometer were measuring slightly different aspects of torque generation (from motor vs. at foot), then the differences between the two were not unreasonably large. The iFES-LCE torque data represented a sampled, slightly delayed, and smoothed feedback response from the motor, whereas the pedals represented the instantaneous torques applied at any given moment.

For constant workloads (during calibration with weights of a constant mass) the motorized cycle regulated velocity well, but for momentary, dynamic perturbations, the velocity varied to a greater degree (Fig. 3, bottom panel). The variability of velocity around a constant average value was due to the delayed velocity-control feedback response noted previously. So, although not absolutely isokinetic, the ergometer approximated constant cadence FES cycling very well across a wide range of cadences.

The iFES-LCE accurately calculated instantaneous and average exercise power outputs from the torque and crank velocity real-time data. Existing FES flywheel ergometers calculate power output against an offset zero. In these systems, "zero" power output is defined by the amount of power required to passively turn the legs and the unbraked flywheel. This offset zero can vary quite markedly between subjects of disparate muscle tone and is also dependent upon their seating position (13). In contrast, the iFES-LCE directly calculated the passive power (the power required to turn the legs) and then subtracted this from the current gross exercise power to give an accurate measure of the "true" power output generated by FES-evoked muscle contractions on each revolution of the crank. The passive power can be checked easily and readjusted in case of changes due to muscle tone or seating position.

During the incremental power output test, the feedback system performed fairly well, keeping the average power outputs within 0.1 W of the mean of the each "target workload" range (Fig. 5). If improvements in power output stability were desired, this could be performed by reducing the "target workload" window to 0.5 W instead of 1 W or by increasing the stimulus amplitude resolution of the muscle stimulator. However, we found the settings used in this test to be sufficient for experiments in our laboratory.

The capability to perform FES cycling at a wide range of cadences, especially low cadences, may lead to improved training outcomes for SCI individuals who might use it. Pedaling cadence is an important characteristic for FES-induced exercise that has not received widespread attention. First, crank angular velocity determines the speed of the muscle contractions. For concentric muscle contractions, the force developed by a muscle decreases as the speed increases. This force-velocity relationship is a fundamental property of skeletal muscle and has been expressed during electrically evoked muscle contraction experiments using surface electrodes as well as for voluntary contractions (14,15). Notwithstanding that muscle fiber characteristics have been shown to be different between denervated and healthy muscles (16), the muscles' force-velocity relationship suggests that slower contraction velocities should produce higher intramuscular forces.

Second, pedaling cadence determines the period of the muscles' contraction-relaxation phases with a lower cadence having a longer period. Intermittent stimulation of healthy muscle has demonstrated that stimulation protocols with equivalent duty cycles, but different contraction-relaxation periods, have dissimilar fatigue rates. In this case, fatigue has been defined as the gradual decrease in muscle force output while the FES-evoked stimulus to contract remains constant. Bergström and Hultman (17) used isometric FES-evoked muscle contractions to demonstrate that fatigue rate was greater in protocols with shorter muscle contraction and relaxation periods. Applying these results to FES cycling suggested that a lower rate of fatigue should be prevalent at low cycling cadences. Furthermore, the muscle's force-velocity relationship would advocate that FES-evoked contractions at slow contraction velocities should produce more force and a reduced fatigue rate compared to traditional cycling at 35–50 rev/min.

If lower pedaling cadences do result in more force being generated throughout a training session then this has important implications for FES-induced muscle training where a desire for increased muscle bulk and augmented leg strength might be important prerequisites for FES-elicited standing or walking programs. Generating higher muscle forces during FES training should lead to an improved potency of strength training stimulus, due to the "Overload Principle" (18). Increased load on the lower limb joints through low cadence training may provide other benefits, for example, augmenting bone density and muscle girth to reduce possible complications due to osteoporosis and pressure sores, respectively.

In a parallel investigation to the development of the iFES-LCE system, Theisen and coworkers (19) observed that cycling with the iFES-LCE at 50 rev/min invoked a significant cardiorespiratory stimulus, including exercise-induced increases of heart rate and oxygen uptake. Further experimental research is required to compare the magnitude of the cardiorespiratory response induced by the iFES-LCE with that produced by presently available commercial FES-LCE systems. The iFES-LCE system described herein can provide a useful research tool due to its ability to control stimulation parameters to a single or multiple muscle groups. Muscle stimulation parameters can be altered easily and in real-time through the computer software interface or via programming changes in the control software. Muscle stimulation can be restricted to a single muscle group to simplify experimental trials (Fig. 3) or can permit optimization of numerous stimulation parameters to compare the variable effects of altering FES frequency, pulse width, amplitude, and waveform type on instantaneous torque and power output.

The system also has the potential to be an effective, portable, and relatively low-cost FES cycle ergometer appropriate for home use. Further development for a home-based system would involve eliminating the laptop computer while integrating all of the control software onto the microcontroller of the muscle stimulator. The iFES-LCE in its current deployment paves the way for further research and development into the effects of pedaling cadence, stimulation parameters, and training protocols for FES cycling in individuals with SCI.

ACKNOWLEDGMENTS

The authors express their gratitude to the volunteer subjects who participated in the many trials necessary for the development of this new FES cycle ergometer. Additionally, the authors are grateful to Mr. Peter Bennet (Multi-TEMS Pty Ltd, Mona Vale, NSW Australia) and Rech Medizintechnik GmBH (Betzenweiler, Germany) who provided technical assistance in the development of the system.

REFERENCES

1. Le CT, Price M. Survival from spinal cord injury. *J Chron Dis* 1982;35:487–492.

2. Garland DE, Stewart CA, Adkins RH et al. Osteoporosis after spinal cord injury. *J Orthop Res* 1992;10: 371–378.

3. Phillips WT, Kiratli BJ, Sarkarati M et al. Effect of spinal cord injury on the heart and cardiovascular fitness. *Curr Probl Cardiol* 1998;23:641–716.

4. Pacy PJ, Hesp R, Halliday DA et al. Muscle and bone in paraplegic patients, and the effect of functional electrical stimulation. *Clin Sci* 1988;75:481–487.

5. Faghri P, Glaser R, Figoni S, Miles D, Gupta S. Feasibility of using two FNS exercise modes for spinal cord injured patients. *Clin Kinesiol* 1989;43 (3):62–68.

6. Faghri PD, Glaser RM, Figoni SF. Functional electrical stimulation leg cycle ergometer exercise: training effects on cardiorespiratory responses of spinal cord injured subjects at rest and during submaximal exercise. *Arch Phys Med Rehabil* 1992;73:1085–1093.

7. Mohr T, Andersen JL, Biering-Sorensen F et al. Increased bone mineral density after prolonged electrically induced cycle training of paralyzed limbs in spinal cord injured man. *Calcif Tissue Int* 1997;61:22–25.

8. Chen JJ, Yu NY, Huang DG, Ann BT, Chang GC. Applying fuzzy logic to control cycling movement induced by functional electrical stimulation. *IEEE Trans Rehab Eng* 1997;5:158–169.

9. Popp MH. Design and construction of a laboratory system for neuro-muscular stimulation of the lower extremities during cycling 1983. M.SC. dissertation, The University of Cape Town.

10. Eichhorn KF, Schubert W, David E. Maintenance, training and functional use of denervated muscles. *J Biomed Eng* 1984;6:205–11.

11. Gfohler M, Angeli T, Eberharter T, Lugner P, Mayr W, Hofer C. Test bed with force-measuring crank for static and dynamic investigations on cycling by means of functional electrical stimulation. *IEEE Trans Neural Syst Rehabil Eng* 2001;9:169–80.

12. Newmiller J, Hull ML, Zajac FE. A mechanically decoupled two force component bicycle pedal dynamometer. *J Biomech* 1988;21:375–86.

13. Schutte LM, Rodgers MM, Zajac FE, Glaser R. Improving the efficacy of electrical stimulation-induced leg cycle ergometry: An analysis based on a dynamic musculoskeletal model. *IEEE Trans Rehabil Eng* 1993;1:109–25.

14. Dudley GA, Harris RT, Duvoisin MR, Hather BM, Buchanan P. Effect of voluntary vs. artificial activation on the relationship of muscle torque to speed. *J Appl Physiol* 1990;69:2215–21.

15. Westing SH, Seger JY, Thorstensson A. Effects of electrical stimulation on eccentric and concentric torque-velocity relationships during knee extension in man. *Acta Physiol Scand* 1990;140:17–22.

16. Gerrits HL, De Haan A, Hopman MT, can der Woude LH, Jones DA, Sargent AJ. Contractile properties of the quadriceps muscle in individuals with spinal cord injury. *Muscle Nerve* 1999;22:1249–56.

17. Bergstrom M, Hultman E. Energy cost and fatigue during intermittent electrical stimulation of human skeletal muscle. *J Appl Physiol* 1988;65:1500–5.

18. Miller C, Thepaut-Mathieu C. Strength training by electrostimulation conditions for efficacy. *Int J Sports Med* 1993;14:20–8.

19. Theisen D, Fornusek C, Raymond J, Davis GM. External power output changes during prolonged cycling with electrical stimulation. *J Rehabil Med* 2002;34:171–5.