Workrate and Cadence Control for Exercise Testing in FES Cycling

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Introduction

Lower-limb FES cycling for paraplegic and tetraplegic subjects has been previously described, with systems for both stationary ergometry and for mobile outdoor cycling being available. The potential therapeutic and medical benefits of FES cycling exercise have recently been surveyed [1]. A key finding is that regular FES cycling exercise leads to improved cardiopulmonary fitness, with a corresponding reduction in the likelihood of cardiovascular disease and improvements in general health.

Most previous exercise studies have utilised commercial FES cycling ergometers. However, with such devices, the exercise workrate and cycling cadence variables are not always well controlled. This is because the control algorithm initially attempts to control cadence to 50 rpm but, as fatigue develops, the cadence is allowed to drop to as low as 35 rpm before the resistive load is reduced in an attempt to increase cadence. Moreover, technical limitations mean that the smallest workrate increment available on these devices is limited to 6 W (with a full workrate range typically of 0-42W) which, for many spinal-cord injured (SCI) subjects, will be a substantial fraction of their maximal exercise capacity. These large, relative increments in workrate limit the sensitivity of the exercise test [2], thus compromising the ability to differentiate between subjects and to accurately define the magnitude of any change in functional status.

As workrate for cycle ergometry is given by the product of angular velocity and resistive torque, the operating point of the exercise depends on both of these variables, as can the efficiency of the exercise. Thus, for a given workrate, variations in cadence can affect the magnitude of the gas exchange responses independently of workrate, consequent to factors such as the oxygen cost of moving the mass of the legs [e.g. 2] and also the recruitment profile of muscle fibre types [4]. Thus, it is crucial that both cadence and load are well controlled. Failure to do so would represent a serious methodological weakness in FES exercise testing studies.

It is of considerable interest that Theisen et al [3] recently proposed an FES-cycling system in which cadence is regulated by feedback control of an electric motor, but stimulation intensity is kept constant, thus allowing power output to vary. The aim of the present investigation is therefore to develop a testbed for exercise testing in FES cycling, in which both cycling cadence and workrate are simultaneously well controlled. A recumbent tricycle with auxiliary electric motor is used, which is adapted for paraplegic users and instrumented for stimulation control. We propose a novel integrated control strategy which simultaneously provides feedback control of leg power output (via automatic adjustment of stimulation intensity) and cycling cadence (via electric motor control). Cardiopulmonary monitoring is carried out using real-time, breath-by-breath gas exchange measurements. We provide indicative results from one paraplegic subject who undertook a maximal incremental exercise test.
Methods

**FES apparatus:** The system is a commercially available recumbent tricycle (Inspired Cycle Engineering Ltd., UK) which is adapted for FES-induced paraplegic cycling. Ankle orthoses stabilise the ankle joint and leg motion. A shaft encoder provides continuous measurement of crank angle; the angular speed (i.e. the cycling cadence) is then obtained by differentiation and filtering.

The tricycle is equipped with an electric motor, connected through gearing to the rear drive wheel, and coupled to the cranks at the front of the tricycle. Thus, even when no power is supplied by the subject, the legs are turned by the motor. A sensor integrated in the crank measures the subject’s leg power independently of the action of the electric motor (Schoberer Rad Messtechnik (SRM), Germany).

This self-contained FES system can be used for mobile cycling, but can also be mounted on a cycle trainer for indoor exercise and training sessions. The cycle trainer (Tacx, Holland) allows different levels of resistance (load) to be set.

The signals from the shaft encoder, throttle and torque/power sensor are interfaced to a PC. These signals are processed by control software running in the PC in order to produce control signals for the stimulator (individual channel control and intensity level) and for the electric motor.

**Stimulation strategy:** Pairs of surface electrodes are attached to each of six muscle groups, i.e. the left and right quadriceps, hamstring and gluteal muscles. The muscle groups are automatically activated at appropriate times during the 360 deg crank cycle, using the continuous crank-angle measurement. The current for each channel was individually adjusted (up to a maximum of 120 mA) and subsequently fixed during cycling sessions; a constant frequency of 20 Hz was used for each channel. The stimulation intensity can be varied during cycling by adjustment of the pulsewidth across a range of 0-800 µs. The same pulsewidth was applied to each channel.

**Feedback control of cycling cadence and power output:** The motorised and instrumented tricycle can be used for simultaneous feedback control of cycling cadence and of leg power output, combined with manual control of total power output at the drive wheel. We propose an integrated control scheme with two independent feedback loops, as shown in figure 1. In the first loop, the electric motor input is automatically adjusted in such a way that the cycling cadence (or, equivalently, depending on the gear engaged, the cycle’s forward speed) is controlled to a desired value by feedback. The setpoint cadence will ordinarily be pre-programmed into the control software. This feedback loop has a relatively high bandwidth, and is designed to compensate for other influences which affect the cadence, including load changes and variable power input from the legs. While cycling with constant cadence, the total workrate at the drive wheel can be manually adjusted by varying the resistive load setting and by changing gear.

The second loop provides feedback control of leg power, as measured at the cranks. Stimulation intensity (here, pulsewidth) is automatically adjusted to keep the measured power close to a reference value, which can be set arbitrarily in the control software.

The net effect of this control scheme is that smooth cycling motion at constant cadence can always be achieved by the motor control loop, even if the leg power contribution varies or becomes low as a result of fatigue, or if the total load changes. Moreover, the leg power output, which represents the subject’s workrate, can be well-controlled to arbitrary values ranging from zero-stimulation workrate (which is negative - see later) up to the level obtained with maximal stimulation intensity. The level of the desired leg power can thus be chosen to keep the subject’s legs working at an “optimal” operating condition, or to achieve a pre-specified workrate profile for exercise testing (e.g. step or incremental).
Subjects. We present indicative data for one male paraplegic subject, aged 29, with a motor-complete spinal cord lesion at level T8/9. He had been participating in a pilot study of FES cycling for approximately 18 months, with one 1-hour cycling session per week. On different days, he completed a series of three incremental exercise tests. All procedures were approved by the Southern General Hospital Ethics Committee; the subject provided informed consent prior to participation.

Gas exchange monitoring. Pulmonary gas exchange measurements were recorded using a portable breath-by-breath system (MetaMax 3B, Cortex Biophysik GmbH, Germany). The breath-by-breath oxygen uptake ($\dot{V}_O_2$) responses for each exercise test were edited to remove outliers and then averaged.

Results

Incremental exercise test results are shown in figure 2. Simultaneous feedback control of cycling cadence and leg power was utilised, at a controlled cadence of 50 rpm (figure 2, upper graph). During the first four minutes the legs were turned at this cadence, but without stimulation. This was achieved by setting the power reference to a value lower than the “retarding torque” observed without stimulation. Subsequently, the power reference was increased in steps of 2 W each minute. Note that the “power” plot in figure 2 shows both the reference power and the measured leg power – the power controller is sufficiently accurate that these two signals are indistinguishable, for $t > 240$ s. Power increments were increased until the stimulation pulsewidth reached a pre-specified limit of 600 $\mu$s (figure 2, middle graph, truncated at $t = 960$ s), at which point no further increase in power output is possible, and the test was therefore discontinued.

Following the imposition of the stimulation at 240 s, $\dot{V}_O_2$ (in l.min$^{-1}$) (figure 2, lower graph) showed an essentially lagged-linear increase with time and therefore workrate, after an initial “kinetic” phase that reflects both the limb-to-lung vascular transit and the system time constant [2]. This response profile is similar to earlier reports for volitional cycling [2], however with a ramp slope ($\Delta\dot{V}_O_2/\Delta$workrate) value of 23.8 ml.min$^{-1}.W^{-1}$ and a mean response time ($\tau'$) of 71 s.
Discussion

The results of this investigation demonstrate that accurate feedback control of cycling cadence and exercise workrate can be achieved simultaneously. The subject’s leg power output (workrate) can be controlled to an arbitrary reference value, and workrate increments can be arbitrarily small. Thus, our control approach significantly extends the functional workrate range for FES-cycling exercise, because the exercise baseline is the workrate corresponding to zero stimulation input (in the tests shown, the workrate baseline is approximately -9 W), rather than a 0 W baseline (which requires a stimulation input sufficient to fully rotate the mass of the legs). In our approach, the stimulation can be gradually increased from 0 µs such that the exercise workrate begins at around -9 W and then increases gradually towards 0 W and beyond. During the “negative workrate” phase, the subject’s legs are not contributing to the total mechanical work done against the load but, as stimulation increases the workrate towards 0 W, the muscles do an increasing amount of work to move the legs, thus decreasing the level of work done by the electric motor to move the legs. When the workrate rises above 0 W, the legs begin to contribute to the total work done against the load.

The corresponding \( \dot{V}_{O_2} \) response kinetics to the incremental FES forcing appear to be linear and first-order. This is consistent with earlier descriptions for volitional cycling in normal subjects [2]. However, the ramp slope value (23.8 ml.min\(^{-1}\).W\(^{-1}\)) was some two-fold higher than reported for volitional cycle ergometry [2], which suggests a degree of inefficiency that might reflect in part the predominant recruitment of fast-twitch muscle fibres with low oxidative efficiency [4]. The long mean response time is consistent with a poor degree of conditioning [2].

In conclusion, the integrated control strategy is effective in facilitating FES exercise testing under conditions of well-controlled cadence and power output. Our control approach significantly extends both the range and the exercise-test sensitivity for FES-cycling and should thus allow more stringent characterisation of physiological
response profiles and, therefore, estimation of key parameters of aerobic function, such as the lactate threshold and the $\dot{V}O_2$ time constant [2]. This represents a substantial advance in the SCI population where the maximal exercise workrate is typically substantially compromised.

Acknowledgements
This work was supported by the UK Engineering and Physical Sciences Research Council (Grant GR/M94717), and by the INSPIRE Foundation.

References