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dicit ei Iesus:
ego sum via et veritas et vita
nemo venit ad Patrem, nisi per me

Control Systems for Function Restoration, Exercise, Fitness and Health in Spinal Cord Injury

A thesis submitted for the degree of Doctor of Science in Engineering (DSc (Eng)) at the University of Glasgow in January 2005

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Abstract

We describe original research contributions to the engineering development of systems which aim to restore function and enable effective exercise for people with spinal cord injury (SCI). Our work utilises functional electrical stimulation (FES) of paralysed muscle. Improving function and general health through participation in exercise is vital to the enhancement of quality of life, well-being and promotion of longevity. Crucial to the development of this research has been judicious use of advanced methods of feedback control engineering; this has been a key enabling factor in many of our original contributions.

The consequences of a spinal cord injury can be severe. The primary effects may include: paralysis and loss of sensation in the legs, arms and trunk; disruption of bladder and bowel function; and disruption of the autonomic regulation of blood pressure, heart rate and lung function. If the abdominal and chest muscles are paralysed, breathing will be compromised, and patients with a high-level cervical injury may require mechanical ventilation.

These primary effects of a spinal cord injury may, over time, lead to a range of debilitating secondary medical complications. These include reduced cardiovascular fitness, urinary tract infection and an associated risk of kidney disease, reduced bone mineral density, the possible development of pressure sores, and muscle spasticity. People with paralysed chest and abdominal muscles are at increased risk of respiratory infection.

Consideration of these factors has led us to focus our research programme in this field on novel engineering solutions which have relevance to the secondary consequences of spinal cord injury, and which may help to alleviate some of their effects. In this thesis we describe our contributions in the following areas:

1. **Control of Paraplegic Standing**: This work concerns upright stance, and aims to provide: (i) automatic feedback control of balance during stance, with the arms free for functional tasks; (ii) methods and apparatus for dynamic standing therapy, which may help to enhance the individual's retained balance skills. This area of work has successfully demonstrated the automatic control of balance during quiet standing in paraplegic subjects. Further, we have established the feasibility of ankle stiffness control in paraplegic subjects using FES, and we have shown that this can be combined with volitional upper-body inputs to achieve stable, arm-free balance.

2. **Lower-limb Cycling**: Lower-limb cycling, achieved through electrical stimulation of paralysed leg-actuating muscles, is an effective exercise intervention. We have described refinements to the engineering design of an FES-cycling system, based upon the adaptation of commercially-available recumbent tricycles (of various designs), some of which are equipped with an auxiliary electric motor. We have contributed new methods of feedback control of key variables including cycle cadence and exercise workrate. These contributions have facilitated further detailed study of the effect of the exercise on cardiopulmonary fitness, bone integrity, spasticity, muscle condition, and factors relating to the likelihood of skin breakdown (i.e. the development of pressure sores).

3. **Upper-limb Exercise in Tetraplegia**: We have developed a new exercise modality for patients with a cervical-level injury and significant loss of arm function. The system allows effective arm ergometry by combining volitional motion with electrical stimulation of the paralysed upper-arm muscles. This work has developed new apparatus and exercise testing protocols, and has examined the effect of the exercise on cardiopulmonary fitness and muscle strength in experiments with tetraplegic subjects.

4. **Modelling and Control of Stimulated Muscle:** This fundamental area of research has investigated dynamic modelling and feedback control design approaches for electrically-stimulated muscle. This work has been applied in the three areas mentioned above.

We identify promising areas for future research. These include extension of work on lower-limb cycling to patients with incomplete injuries, to those with cervical-level injuries, and to children with SCI. We wish to participate in a multi-centre clinical study of implanted nerve-root stimulation technology for restoration of bladder and bowel control, and for lower-limb exercise (including cycling). We have initiated a study of treadmill-based gait therapy for incomplete-lesion patients. The goals of this study are to develop test protocols for accurate characterisation of cardiopulmonary status, and to determine whether this form of cyclical lower-limb exercise has a positive impact on retained voluntary leg function. It is often the case that it is those people most severely affected by neurological impairment who stand to gain the most from these approaches (e.g. high-level tetraplegia, paediatric spinal cord injury, etc.). We must therefore continue to seek ways in which the work can be developed for the maximum benefit of these patients.

In conclusion, this thesis has described original research contributions to the engineering development of systems which aim to restore important function and to enable effective exercise for people with spinal cord injury. An important facet of our work has been the application of feedback control methods; this has been an enabling factor in several areas of study. We have focused on areas which promise improved fitness and general health, and which may alleviate some of the secondary consequences of spinal cord injury. This work encompasses fundamental research, clinical studies, and the pursuit of technology transfer into clinical practice.

Finally, we recognise the growing awareness of and interest in central nervous system plasticity, and in the broad field of central neural regeneration and repair. It is therefore timely to ask whether cyclical exercise interventions can lead to improvement of volitional function in patients with incomplete or discomplete lesions. Such improvements may, we speculate, result from the strengthening of muscles which retain at least partial volitional control, or from neural plasticity and re-organisation, or from regeneration effects (neurogenesis and functional connectivity). A key requirement in this line of investigation, and a major challenge, will be to develop or to utilise methods which can detect changes in a patient's volitional function and neurological status, and which can isolate the source of such changes. Should reliable methods become available, the way to the study of *recovery* of function through cyclical exercise would be opened. These considerations will remain, we propose, an indispensable complement to cell-based surgical interventions which may become available in the future.

AMDG

Acknowledgements

Professor Nick Donaldson of University College London introduced me to research in the fields of functional electrical stimulation and spinal cord injury in 1992. A period of initial collaboration led subsequently to the establishment of the Centre for Rehabilitation Engineering at the University of Glasgow in 1998. Many of the ideas and results reported in this thesis are deeply connected to the on-going collaboration with Nick and his group.

I would like to express gratitude to the clinical staff at the Queen Elizabeth National Spinal Injuries Unit at the Southern General Hospital in Glasgow. Particular thanks are due to Mr Matthew Fraser and Dr Alan McLean, Consultants in spinal injury; they have supported this work in every possible way, and are actively engaged in its further development. Likewise, Mr David Allan, Clinical Director of the Spinal Unit, has provided continuous encouragement and support for the development of clinically-relevant research.

My colleagues at the Centre for Rehabilitation Engineering, past and present, have contributed to this research programme at levels which go well above and beyond the call of duty. This commitment is acknowledged with deep gratitude. In particular, much of the research reported here has evolved through interaction with my co-worker Dr Henrik Gollee who has been with the Centre since its inception.

Most importantly, recognition must be given to all of the people with a spinal cord injury who have volunteered to participate in this work as research subjects. They have given most freely of their time and energy, and have been a constant source of education, new ideas, feedback and constructive criticism.

Figure credits: Dr Thomas Schauer and Dr Ralf-Peter Jaime, former PhD students in the Centre for Rehabilitation Engineering, kindly provided a number of schematic diagrams used in sections 1, 2 and 3.

Contents

1 Introduction

The consequences of a spinal cord injury (SCI) can be severe. For any individual the effects will depend upon the neurological level of the injury, and on the degree to which neural traffic is interrupted at the lesion site. However, the primary effects may include: paralysis and loss of sensation in the legs, arms and trunk; disruption of bladder and bowel function; and disruption of the autonomic regulation of blood pressure, heart rate and lung function [63]. If the abdominal and chest muscles are paralysed, breathing will be compromised, and patients with a high-level cervical injury may require mechanical ventilation.

These primary effects of a spinal cord injury may, over time, lead to a range of debilitating secondary medical complications. Muscle paralysis, the associated disuse atrophy, and the consequent reduction in effective muscle mass mean that patients may not be able to effectively exercise to maintain their cardiovascular fitness, which leads to a possible increased risk of heart disease. In addition to the social inconvenience of incontinence, poor bladder voiding increases the risk of urinary tract infection, and a careful bladder management programme must be initiated to guard against this and the associated risk of kidney disease. The absence of normal loading of the long leg bones results in rapid reduction in bone mineral density and consequently an increased risk of fracture. Insensate skin areas can lose vascularisation. This, together with increased skin-interface pressures (which can be exacerbated by reduced muscle bulk) and the patient's lack of sensation, can readily lead to skin breakdown and to the development of pressure sores, which may in turn be open to infection. Patients are therefore given careful instruction in the meticulous care of their skin. Absence of volitional control of the abdominal and chest muscles (particularly in cervical injuries) leads to reduced tidal volume during breathing, but also to the lack of an effective cough mechanism. It is therefore difficult to clear secretions and such patients are at risk of serious respiratory infection. Finally, muscles innervated distal to the spinal cord lesion will generally experience elevated reflex activity (spasticity) which can be very troublesome for the patient.

The dependence of functional loss on the level of the spinal cord lesion can be understood with reference to the diagrams of the somatic and autonomic nervous systems shown in figures 1 and 2. The neurological level of a spinal cord injury is defined clinically as the most distal spinal cord segment down to which normal function is evident. The bladder and bowel are regulated by a carefully balanced interaction of voluntary and autonomic control. The voluntary sphincter muscles are supplied by sacral spinal nerves, while autonomic control of bladder and bowel is via sympathetic pathways from the upper-lumbar region, and parasympathetic pathways with outflow from the sacral cord. Thus, a spinal cord lesion at any level can be disruptive of bladder and bowel function. The leg muscles are supplied from the lumbar region of the cord, while the abdomen and chest muscles are supplied from increasingly higher levels in the thoracic cord. Thus, a thoracic-level lesion will result in loss of leg function and some degree of paralysis and loss of sensation in the trunk. The hand and arms take their nerve supply mainly from the lower-cervical region. In this area, the actual loss of function is very finely tuned to the level of the injury. The hand is supplied mainly by the C8 and T1 spinal nerves, elbow extensor muscles (e.g. triceps) by C7–C8, elbow flexors (e.g. biceps) and wrist extensors by C5 and C6, and the shoulder musculature by C3–C4. Crucially, the diaphragm, which is the main muscle of inspiration during breathing, is supplied by the C3–C5 spinal nerves, but mainly by C3/C4. A patient with a neurological level above C3 will usually require the support of a mechanical ventilator for breathing.

In consideration of new exercise interventions, it is important to take account of possible disruption to the autonomic nervous system and any limitations this may impose. The

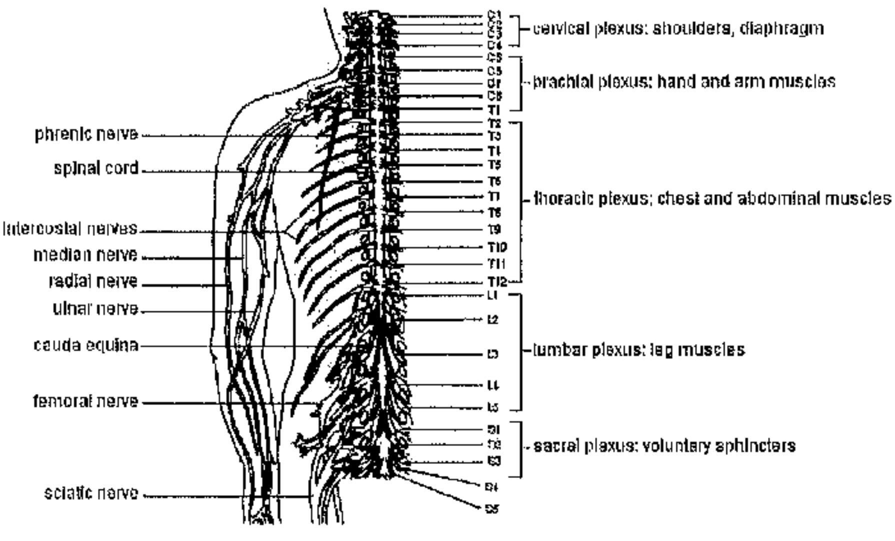


Figure 1: Somatic motor system and muscle innervation.
(Adapted from http://fourteen.apptechnc.net/~windelspecht/nervous/index.htm)

part of the sympathetic division of the autonomic nervous system which serves to produce increased activity in the heart and lungs in response to exercise has its outflow from the T1–T5 spinal nerves (see figure 2), while parasympathetic supply is from the vagus nerve (a cranial nerve). Any spinal cord lesion above T5 may therefore result in some disruption of and fundamental limitation in sympathetic exercise responses (e.g. a blunted heart rate increase), and in sympathetic-parasympathetic imbalance. This is particularly evident in complete cervical-level injuries, where supra-spinal control of sympathetic pathways is completely lost.

Spinal cord injuries have varying degrees of "completeness". The clinical classification of completeness is based upon an empirical five-point scale (A–E). A clinically-complete spinal cord injury (classification "A-complete") is defined as the absence of motor and sensory function below the level of the lesion, and is assessed by clinical inspection of motor and sensory function. A classification of "B-incomplete" means that there is some retained sensation but no motor function below the neurological level. Incomplete C and D classifications have increasing preservation of motor function, while E means that motor and sensory function are normal. Anatomically, a spinal cord injury is classified as being complete, incomplete, or "discomplete" [64]. The latter describes loss of all neurological function below the lesion (i.e. a clinical A-complete lesion), but with physiological or anatomical continuity of nervous system tracts across the lesion. Kakulas points out that true and complete anatomical transection of the cord due to trauma is not common [64].

Thus, as a result of the complexity of the peripheral nervous system and its spinal-nerve supply, together with a wide range of injury mechanisms, patients can present with complex and often unique patterns of neurological impairment following a spinal cord injury.

This analysis of the primary and secondary effects of spinal cord injury provides grounds for the focus of the research programme described in this thesis. While the lay person may

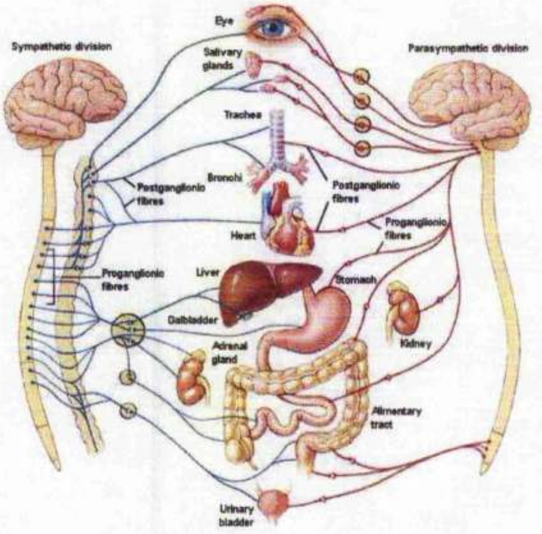


Figure 2: Autonomic nervous system: sympathetic (blue) and parasympathetic (red) divisions. (Adapted from http://www.mhhe.com/socscience/intro/ibank/set4.htm)

frequently imagine that the aim of engineering research in this field must be to restore the ability to walk, the immediate priorities of most patients and clinicians are quite different. It is true that in the time immediately following spinal trauma a patient will be acutely aware of their loss of upright mobility, and will earnestly question whether they will be able to walk again, but this viewpoint will often change during the months spent in hospital in primary rehabilitation. During this phase patients will learn about the importance and practicalities of bladder and bowel management, skin care, spasticity and perhaps a range of further secondary medical issues. These are factors which patients must integrate into their lifestyle and deal with on a day-to-day basis, and they will undergo intensive physical and occupational therapy in order to maximise their functional abilities. Patients will also be acutely aware of the physiological changes occurring in their bodies as a result of muscle disuse and restricted exercise options, including general cardiovascular de-conditioning and loss of fitness. Most spinal injuries centres offer gait training with the support of orthoses (i.e. long leg braces in conjunction with a support frame), but it is the case that very few complete-lesion patients continue to use this following discharge from hospital. Thus, the major concerns and preoccupations of most patients living with a chronic spinal cord injury, we suggest, relate not to the primary loss of the ability to walk, but to the prevention and management of the manifold secondary medical complications which may arise.

Of course, all patients would embrace any practical technological solution which restores the ability to walk. However, the results of research into restoration of walking for complete-lesion patients are currently very limited and this is not a practical and convenient option for the vast majority of patients. This situation stems from the complexity of the walking task, which requires integration of wide-ranging sensory input, control processing, and distributed actuation mechanisms to produce effective gait. Added to this is the requirement to continuously maintain upright balance and postural stability. However, the situation is different for those incomplete-lesion patients who retain the ability to stand and walk under volitional control, even if only to a restricted degree. For these patients, technological solutions have often produced effective improvement in gait.

These considerations have led us to focus our research programme in this field on novel engineering solutions which have relevance to the secondary consequences of spinal cord injury, and which may help to alleviate some of their effects. In this thesis we describe our contributions in the following areas:

1. **Control of Paraplegic Standing** (section 2): This work concerns upright stance, and aims to provide (i) automatic feedback control of balance during stance, with the arms free for functional tasks, (ii) methods and apparatus for dynamic standing therapy, which may help to enhance the individual's retained balance skills.

2. **Lower-limb Cycling** (section 3): Lower-limb cycling, achieved through electrical stimulation of paralysed muscle, is an effective exercise intervention. We have contributed new methods of feedback control of key variables including cycle cadence and exercise workrate. These contributions have facilitated further detailed study of the effect of the exercise on cardiopulmonary fitness, bone integrity, spasticity, muscle condition, and factors relating to the likelihood of skin breakdown (i.e. the development of pressure sores).

3. **Upper-limb Exercise in Tetraplegia** (section 4): We have developed a new exercise modality for patients with a cervical-level injury and significant loss of arm function. The system allows effective arm ergometry by combining volitional motion with electrical stimulation of the paralysed upper-arm muscles. This work has developed new apparatus and exercise testing protocols, and has examined the effect of the exercise on cardiopulmonary fitness and muscle strength.

4. **Modelling and Control of Stimulated Muscle** (section 5): This fundamental area of research has investigated dynamic modelling and feedback control design approaches for electrically-stimulated muscle. This work has been applied in the three areas mentioned above.

An important technique utilised in our work is *functional electrical stimulation* (FES). This is a method which facilitates the activation of paralysed muscle. FES is based upon the application of low levels of pulsed electrical current to motor nerves. Stimulation electrodes can be implanted and attached to motor nerves either centrally (in the spinal cord, or on spinal-nerve roots) or peripherally. Stimulation can also be applied percutaneously, to peripheral nerve, using needle electrodes inserted through the skin. Our work, however, has thus far utilised transcutaneous stimulation where adhesive electrodes are attached to the skin in the location of the target muscle and its nerve supply. The principle of FES is to create an electric field which serves to depolarise individual motor neurone axons beyond their resting potential and towards their firing threshold. If the threshold is exceeded then an action potential will be generated in the axon, resulting in the contraction of all muscle fibres belonging to that motor unit. Graded muscle contraction can be achieved by gradually increasing the intensity of stimulation (i.e. the charge in each stimulation pulse, which is modulated by varying the pulsewidth and current amplitude) so that an increasing number of motor units are recruited, or by increasing the stimulation frequency so that active motor units are recruited more often.

Although FES can elicit strong and effective muscle contractions, there are significant limitations. One key issue is the order in which muscle fibres of different histochemical types

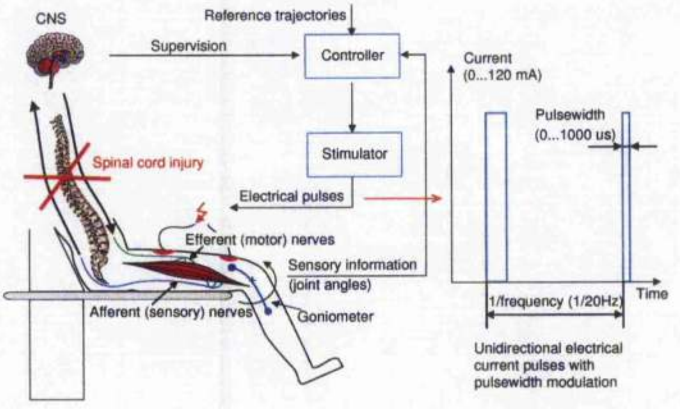


Figure 3: Musculoskeletal control system with FES.

are recruited with increasing stimulation intensity. Normal recruitment of muscle fibres, controlled by the central nervous system (CNS), follows the *size principle*. This means that at low levels of activity, where muscle forces are relatively low, muscle fibres associated with "small" motor units are recruited. These motor units have a relatively small number of muscle fibres that are characteristically highly oxidative, slow-twitch fibres with high aerobic capacity and resistance to fatigue. As activity levels are increased, so "large" motor units are increasingly recruited. These motor units each have large numbers of muscle fibres, but these are characteristically glycolytic, fast-twitch fibres which have low aerobic capacity and poor fatigue resistance. Ordinarily, such fast-twitch fibres are utilised for short periods of high-intensity, high-force activity. Muscle activation using FES, however, characteristically has an inverse pattern of muscle fibre activation where fast-twitch, fatigueable fibres are preferentially recruited at low levels of stimulation, followed by slow-twitch, fatigue-resistant fibres as stimulation intensity is increased. This is because the fast-twitch fibres of large motor units are associated with large-diameter nerve axons, which have a lower firing threshold to externally-applied stimulation. Thus, electrically stimulated muscle is generally observed to fatigue rapidly and to have limited force generation capacity. Added to this is the fact that paralysed muscle will initially be in poor condition due to disuse atrophy, and will generally display some degree of unwanted contractile activity due to elevated reflex responses (spasticity). As will be illustrated in the sequel, some of our work has focused on the search for optimised patterns of stimulation which aim to maximise force, power output and efficiency, while minimising rates of fatigue.

A further consideration in the effective application of FES is that target muscles must retain healthy innervation from the cell bodies of nerves, which reside in the anterior horn of the spinal cord, through to the neuromuscular junction. It is often the case that cell bodies or nerve roots are damaged in the region of the trauma site in the spinal cord, in which case the associated motor neurones and muscle fibres will degenerate totally and will not therefore respond to the low levels of stimulation which are approved for use in FES.

A schematic illustration of the integration of FES within a musculoskeletal control system is shown in figure 3. This illustrates the task of controlling the angle of the knee joint using activation of the quadriceps muscle group. The figure shows artificial feedback control for an individual with lower-limb paralysis; this can be contrasted with natural feedback control of movement. In both cases the problem requires the key elements of sensory input (determination of knee-joint angle), control processing, and actuation (muscle contraction). However, realisation of these elements is very different in each case: in the natural case the joint angle is determined visually and through knee-joint receptors, while in the artificial case an external sensor such as a goniometer is required; in the natural case the control processing is carried out in the brain, but in the artificial case it is implemented in an external processor; natural muscle contraction is achieved via motor control pathways under command of the brain, and computer-controlled FES is used in the artificial system.

A central theme in much of the rehabilitation-engineering work presented in the sequel (sections 2–5) is the application of advanced feedback methods for control of FES-induced musculoskeletal motion or exercise. In some areas, the feedback control method employed has itself been the enabling factor in the achievement of some functionality which had not previously been successfully achieved (e.g. control of balance during quiet standing, or sensitivity of exercise testing during lower-limb cycling). These methods stem from a body of original work on control theory, which is described in section 6.

2 Control of Paraplegic Standing

One of the possible consequences of a spinal cord injury is impaired postural control. In the case of complete lower-limb paralysis, this can encompass loss of the ability to stand, to maintain balance while standing, and of course to walk. We have conducted research on the feasibility of automatic feedback control of balance during quiet standing in complete paraplegia (section 2.1), on the feasibility of ankle stiffness control using FES while standing (section 2.2), and on the integration of the retained motor control skills of the non-paralysed parts of the upper body with artificial control of the lower limbs within the overall postural control task (section 2.3).

Upright standing therapy is routinely applied during the primary rehabilitation of patients with spinal cord injury, and patients are encouraged to continue this therapy at home upon discharge from hospital. Standing is usually achieved with the support of a static standing frame. After rising from their wheelchair, patients are stabilised using bracing at the knee and hip joints so that upright posture is possible. A typical therapy regime is for patients to stand quietly in the frame for a period of 30 min, twice per day. Standing in this way loads the long leg bones and is thought to be beneficial for maintenance of bone integrity. However, although standing therapy is clinically prescribed for maintenance of bone quality, clear scientific evidence to support its efficacy is lacking, probably due to the static nature of bone loading. Standing also helps to prevent joint contractures, improves renal function and circulation, and can relieve spasticity. A further option for standing is the use of passive long-leg braces. The patient is positioned between parallel bars, and is required to hold on to the bars to maintain postural stability.

With passive standing frames and leg braces the patient's leg muscles remain inactive. FES has previously been employed for restoration of standing in paraplegia, in conjunction with a passive frame [65, 66]. However, full postural stability is again provided by the frame, or by the subject holding on to the frame. The purpose of FES in this type of system is simply to give open-loop, bilateral stimulation of the knee extensor muscles (i.e. the quadriceps group) so that the knee joints remain extended during stance. The hip joint is held passively hyper-extended (the so-called "C-posture"), while the patient maintains postural stability by means of the arms holding on to a suitable support.

These considerations have led to the investigation of systems which might extend FES-supported standing through provision of automatic, feedback control of balance while standing. This has been termed "arm-free" or "unsupported" standing, and is functionally attractive because the patient's arms are freed from the balancing activity and become available for ordinary tasks such as the manipulation of objects and purposeful interaction with the environment. It is also possible that the use of FES while standing will be more beneficial for bone integrity than passive standing, since muscle contraction will introduce dynamic loading of the bones.

2.1 Balance control during quiet unsupported standing

The challenge of providing automatic balance control for paraplegics using FES, without the need for arm support, has been of interest for a considerable time. The key technical challenges, from a control engineering viewpoint, are that: (i) the body dynamics are inherently unstable, which imposes the fundamental limitation that any feedback system must provide a certain minimum bandwidth (there is a lower bound on the bandwidth required for closed-loop stability); and (ii) the actuators in this system are the paralysed human muscles which are generally weak, tend to fatigue rapidly, and are subject to possibly-large, random disturbances (i.e. spasticity). These two issues clearly interact, since a progressively fatigueing muscle (i.e. an actuator with continually decreasing gain) makes it more likely that the bandwidth requirement will eventually be violated and, therefore, that stability will be lost.

The first significant study of automatic (artificial) stabilisation of quiet standing in paraplegia was carried out by Jaeger [67]. In a simulation study, Jaeger assumed that all joints above the ankles were locked, and proposed simple ankle control schemes which stabilised this single-link inverted pendulum arrangement. Jaeger's idealised model consisted of second-order linear and time-invariant muscle dynamics, together with a single-link inverted pendulum model of the unstable body dynamics. This configuration was regulated by a standard PID (Proportional-Integral-Derivative) feedback controller. While it is immediately evident that a combination of proportional and derivative feedback is sufficient to stabilise this model, Jaeger did illustrate in this work that, under realistic biomechanical constraints and subject to the availability of sufficient muscle force, it might be possible to restore balance during quiet standing in this configuration, but the method was not experimentally tested.

A simulation study was also carried out by Khang and Zajac [68, 69]. This work considered a multi-link situation with nonlinear muscle actuation at the ankle, knee and hip joints. They developed an algorithm to distribute the net activation calculated by a feedback controller among all the muscles crossing each joint. A static output gain controller was derived, based upon linearised body dynamics and other approximations concerning the performance capabilities of the muscles. A limitation of the work was the inability to take into account any residual sensory and motor function in non-paralysed parts of the upper body, which might contribute to posture stabilisation. The approach was not experimentally tested with human subjects, and therefore the practicality of the proposal was not established.

Our own work in this field was initially focused on a simplified postural task, i.e. that of quiet upright stance, and the nominal feedback design was based upon a simple theoretical model of body dynamics in conjunction with empirically determined models of muscle dynamics. However, a significant advance over previous work was our intention to study the performance and limitations of our approach in experiments with human paraplegic subjects. As with Jaeger's work, we assumed a configuration in which all joints above the ankles are locked so that the body dynamics could be modelled as an ideal inverted single-link pendulum as shown in figure 4. The ankle moment m is generated by stimulation of the ankle plantarflexor muscles (the calf muscles), and by any residual mechanical stiffness in the ankle joint.

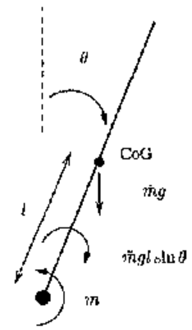


Figure 4: Idealised biomechanics of body dynamics (inverted pendulum). $\tilde{m}$ represents the above-ankle body mass, l is the distance of the centre of gravity (CoG) from the ankle joint, θ is the angle of body inclination, and m is the total ankle moment resulting from ankle-muscle activation. Body weight $\tilde{m}g$ causes a toppling moment $\tilde{m}gl\sin\theta$ around the ankle joint.

This simple biomechanical model was augmented with an empirically determined model of the relationship between muscle stimulation intensity, denoted as p (in our work, the stimulation pulsewidth), and the resulting ankle moment m. Determination of the parameters of this model consisted of an experimental procedure with the human subject, and resulted in a family of linear dynamic models each valid in some operating region (i.e. over some range of pulsewidth values), augmented with a static nonlinearity representing muscle recruitment properties. This approach was based upon original contributions to multi-level and nonlinear models for the dynamic response of electrically stimulated muscle [23, 26], as described further in section 5.

Experimentally, this simple postural configuration was realised using a custom-designed apparatus known as the "Wobbler", developed and built by Donaldson [70] and illustrated in figure 5. While standing in the apparatus the subject wears a custom-sized posterior body shell which locks the knee and hip joints and supports the head and neck. Thus, the only free motion allowed is that around the ankle joint. For safety, four light ropes are attached to the shoulders of the body brace and from there to a frame attached to the ceiling. When the ropes are tight the body cannot sway, but they can be slackened to allow motion within predefined limits in the sagittal plane and with an axis of rotation at the ankle joints. A string attached to the body brace at shoulder level is wound round a pulley attached to a potentiometer placed well behind the subject. This allows measurement of the subject's inclination angle θ. The subject's feet are positioned in footboxes connected to a rotating shaft aligned with the ankle axis which can be driven by an electric motor. Strain gauges within the shaft allow independent measurement of the left and right ankle moments and the total moment m. When the motor is switched, on a gearing mechanism induces a rocking motion in the shaft and the feet are "wobbled", whence the name of the device. The shaft angle is measured by a potentiometer.

In a first step, we designed and experimentally tested an optimal control approach for control of ankle moment during standing, but with the body mechanically constrained for stability [1]. We went on to investigate the design of stabilising controllers for the body dynamics, which incorporated the ankle moment controller. A nested-loop structure for postural control had previously been proposed by Donaldson [71] and this was adopted in our further work. The control structure is illustrated schematically in figures 6 and 7. An inner-loop controller C_m which has high bandwidth is employed for ankle moment control, while an outer-loop controller C_θ is designed to stabilise the plant consisting of the inner closed loop and the inverted pendulum dynamics. The principal advantages of the nested loop structure

Figure 5: Two views of a subject standing in the Wobbler apparatus.

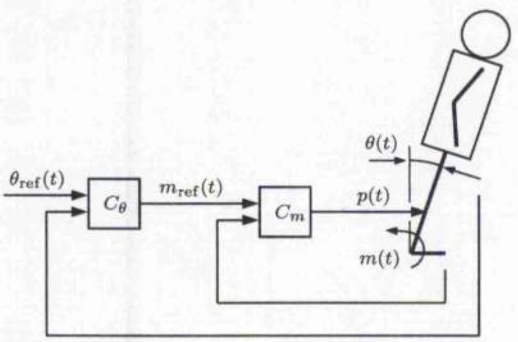


Figure 6: Nested loop control structure. θ is the inclination angle, m is the ankle moment and p the pulsewidth of the stimulation. C_m is the moment controller and C_θ is the angle controller. The desired values for ankle moment and inclination angle are m_{ref} and θ_{ref}, respectively.

are that uncertainty in the muscle dynamics is reduced, and random disturbances to ankle moment caused within the muscles (e.g. from spasticity) are regulated in the first instance by the inner-loop compensator C_m. Both of these features ease the task of control for the angle compensator C_θ. We also note that the optimal control approach employed, which is one of a family of analytical (model-based) design techniques, has the advantage that the controller order is determined by the control design specifications and the design model, in contrast to PD or PID compensators which have a fixed, low-order structure. Such flexibility has been found necessary in order to fully meet design targets and address fundamental limitations across the full frequency range of the design problem.

An optimal control approach for the design of the angle controller, implemented in the nested-loop structure in conjunction with the optimal moment controller, was fully developed and analysed [2]. We then carried out experimental evaluation with both neurologically intact

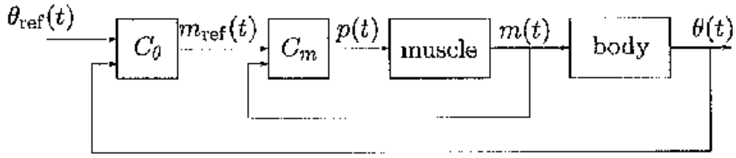


Figure 7: Nested loop structure for unsupported standing, equivalent to the structure of figure 6.

subjects (with their volitional postural control mechanisms appropriately neutralised) and with paraplegic subjects [3]. We demonstrated that intact subjects could be stabilised while standing for prolonged periods, but that paraplegic subjects were able to stand for only very short periods of time (typically for 20–30 s). Despite the modest performance outcomes, this work represents a significant advance because it illustrated for the first time in experiments with human paraplegic subjects that artificial control of balance during quiet standing is technically feasible.

We identified the principal limitations of the approach to be muscle weakness, rapid fatigue, and spasticity in the stimulated muscles. These limitations are dependent upon the physical characteristics of each subject, and may be addressed in part by a programme of muscle re-conditioning prior to standing. However, the underlying parameterisation and design approach for the artificial balance controller has a crucial effect on the length of time during which successful standing is achieved. We thus set out to develop a set of important changes to the control design approach, and to evaluate the effect of these changes. The main modifications included:

- Pole assignment design was used instead of LQG, and the nominal closed-loop poles were selected in order to achieve desired risetime and damping in the closed-loop step response. This has the advantage that the nominal closed-loop response is directly specified by the design parameters and does not depend on the plant model. This is in contrast to the previously employed LQG strategy where, for a given set of design parameters (the cost function weights), the nominal closed-loop response will be different for different plant models. This is very important because the dynamics of electrically stimulated muscle can vary widely between individuals.

- Previously, the total desired ankle moment was split equally between the left and right sides and a controller independently designed for each side. It was found, however, that paraplegic subjects often have a strong asymmetry in the left/right muscle responses, in which case it is not reasonable to demand the same moment from both sides. The solution was to apply the same stimulation pulsewidth to left and right legs, and simply to require some total ankle moment to be achieved - this has the advantage that total moment is balanced between the two legs in a natural way, depending on the ability of each leg to deliver force for a given stimulation level. In addition, the complexity of the controller is reduced since there is only a single, total moment controller, rather than a moment controller for each side.

- A number of further changes were implemented. These include the option of removing integral action from the inner control loop (in an attempt to increase the bandwidth), the option of including the closed-loop characteristics of the inner loop as part of the design plant for the outer loop (in previous work the inner loop was neglected under the assumption that it has a relatively high bandwidth – this assumption becomes less valid as the muscle fatigues), and the option of including a notch filter in the outer loop design. Each of these changes was found to be valuable.

All of these design changes, and the results of evaluation with intact subjects, are documented in the literature [4, 5]. The experimental results demonstrated that the full range of design changes improved the reliability and consistency of the control system, and that the system performed closely in accordance with the closed-loop design specifications. Subsequent evaluation with paraplegic subjects showed that, with the revised design approach, reliable and stable standing could be achieved for periods of several minutes at a time [6].

A feature of time-domain feedback design approaches such as LQG and pole assignment is that quantitative measures of plant uncertainty cannot be explicitly taken into account in controller synthesis. The effect of uncertainty can only be considered in post-design analysis, which may then lead to refinement of the design and further analysis. This iterative procedure carries on until satisfactory closed-loop characteristics are achieved. This procedure is perfectly satisfactory in many cases, and it should always be recalled that a fundamental property of feedback is a reduction in the effect of uncertainty and thus the provision of robustness. Nevertheless, we have considered design approaches which allow explicit formulation of uncertainty models and provide guarantees of robustness within the limits of the specified uncertainty. Thus, a robust design procedure specific to the control of balance during stance was developed and experimentally tested [7, 8].

The principal source of uncertainty in this formulation is considered to be the variation in muscle dynamics, which are known to depend on activation level, and which vary over time due to fatigue. The physical parameters of the body dynamics (mass etc.) are relatively straightforward to measure and vary only slowly with time. Uncertainty bounds were developed using empirical models of muscle dynamics which were obtained in open-loop identification tests at different mean levels of stimulation intensity. An H_∞ design approach was then employed for controller synthesis. The approach allows for the selection of cost function weighting elements designed to meet specific requirements across the closed-loop frequency response. This robust control approach was evaluated in experiments with a paraplegic subject and was found to perform very favourably in comparison with previous approaches: in one experiment the subject was able to stand quietly for over seven minutes [7]. During this time the stimulation intensity was observed to increase approximately linearly with time due to progressive muscle fatigue. When the upper limit on stimulation intensity was reached, the muscle could no longer provide the ankle torque required for standing and the subject fell gently forward into the safety ropes. Although the results with this type of controller were very promising, it is the case that only very limited experimental evaluation was possible so that caution must be exercised in assessment of the general strength of this approach.

In summary, this body of work has focused on the task of stabilisation of quiet standing posture in a single-link configuration where only the ankle joint is free to move. The work is important because of the rigorous feedback analysis and design approaches employed, and because for the first time prolonged periods of artificial balance control were achieved with paraplegic subjects. Having proven the feasibility of the approach, we continue efforts to refine the feedback methods and to achieve improved performance outcomes.

This fundamental work has established that artificial balance control, where ankle actuation is provided by FES of the ankle musculature, is feasible. However, we recognise the practical limitations of the simplified single-link paradigm and we have moved on to study additional degrees of freedom in the postural control task. Consequently, we have introduced the possibility of actuation at the hip joint, and we consider the reality that paraplegic subjects will retain some degree of volitional motor control in the upper body which can contribute to posture stabilisation. The requirement is therefore to develop a system which combines artificial control at the ankle joint with voluntary upper-body motor control. The results of a study in which this has been evaluated are described in section 2.3. However, it can be readily understood that the introduction of additional actuation options can ease the feedback requirement for the controllers operating in the lower limbs. For this reason, we first report the results of research into the properties of low-order control design specifications at the ankle joint, namely ankle stiffness control.

2.2 Ankle stiffness control

Several studies have investigated the mechanisms of volitional control of the ankle muscles during stance in neurologically intact subjects (e.g. [72]). Intrinsic mechanical properties, tonic muscle contraction, and reflex activity combine to produce generally complex ankle dynamics. Thus, elastic, viscous and inertial (and possibly higher-order) components of force at the ankle joint can be identified [73, 74].

Loram *et al.* [75] measured gross ankle *impedance* in intact subjects while they imitated quiet standing. They demonstrated a frequency-dependence of gross ankle dynamics, but did not attempt to identify individual physical components of ankle activity (i.e. stiffness, viscosity etc.). If the rate of change of ankle angle is low then elastic properties dominate; a number of studies (e.g. [72, 73]) have used this fact to focus on ankle stiffness properties. Fitzpatrick *et al.* [72] modelled stance as a single-link inverted pendulum. They draw the distinction between *load stiffness*, which acts merely to counteract gravity, and *reflex stiffness*, which acts to oppose perturbations around a mean, equilibrium posture (and corresponding mean load torque). The study [72] experimentally measured reflex stiffness at the ankle during imperceptibly slow perturbations while standing. The results show that reflex stiffness is significantly greater than load stiffness, and that the level of stiffness is task dependent, i.e. reflex stiffness measured while subjects were told to "stand still" is significantly greater than stiffness measured when subjects were asked to "stand at ease". EMG (electromyographic) measurements taken during the experiments also show that increased stiffness results directly from modification of reflex activity, i.e. from increased muscle activation. This is supported by Weiss *et al.* [73] who show (albeit in experiments with supine subjects) that ankle stiffness increases linearly with mean ankle torque.

The focus on stiffness in these studies results from the fact that mechanical stiffness is a simple concept which expresses some aspects of ankle activity during stance in a functionally meaningful way. It is presently unknown whether the central nervous system solely or primarily regulates ankle joint stiffness during standing. In fact, the neuromuscular mechanisms behind these observed properties remains a subject of active debate [76, 77, 78, 79, 80, 81]. Whatever the underlying mechanism, it is clear that stiffness plays an important role in posture stabilisation. Crucially, within the context of restoration of balance during standing in paraplegic subjects, the precise mechanism of neuromuscular control is relatively unimportant. What is important is the fact that stiffness properties are known to make a major contribution to balance, that typical stiffness values have been reported for intact subjects, and that the required minimal load stiffness can be easily computed from biomechanical parameters.

We have therefore proposed that stiffness may be a convenient variable to attempt to regulate in an artificial control system acting at the ankle joint. In particular, we set out to investigate whether controlled FES of the ankle muscles might be used to vary muscle contraction and ankle torque, and thereby to achieve a desired level of stiffness during standing.

As a first step, the work reported in [9] and summarised below was carried out in the

Wobbler apparatus, which allows the fundamental limitations of the approach to be studied with minimal intervention from central motor control. Thus, our initial goal has been to investigate the feasibility (and accuracy) of ankle stiffness control using FES while standing. Our use here of ankle extensor and flexor muscles extends our previous work on unsupported standing which considered only forward-leaning postures and plantarflexor stimulation.

The feedback control structure for ankle stiffness control is shown in figure 8. M represents the dynamics of the stimulated muscles and B are the body dynamics of the inverted pendulum. The strategy uses a desired stiffness k^s_{ref} (specified by the experimenter) and a measurement of ankle-joint angle θ to compute the required ankle moment as $m_{\mathrm{ref}} = k^s_{\mathrm{ref}}\theta$. This required moment is achieved by a feedback controller C_m which uses a measurement of total ankle moment m and adjusts the dorsiflexor/plantarflexor stimulation intensity variable p appropriately. Positive values of p indicate plantarflexor stimulation, while negative values of p correspond to dorsiflexor stimulation, i.e. the pulsewidth for the appropriate muscle group is always given by $|p|$. m_e is an externally-applied moment acting on the body. This external moment is applied by the experimenter to maintain postural stability and to manually move the body through a range of angles while the stiffness controller is tested. The

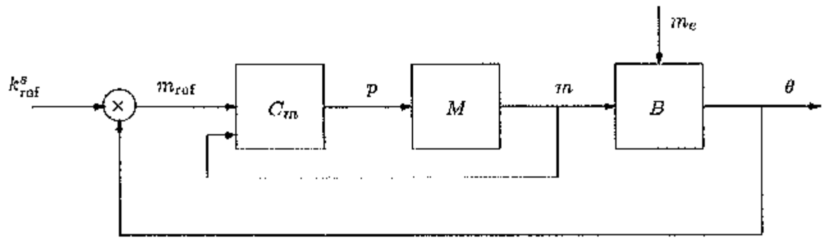


Figure 8: Ankle stiffness control structure. The block M incorporates a switching mechanism (not shown) which translates positive values of the variable p to plantarflexor stimulation and negative values ($p < 0$) to dorsiflexor stimulation.

moment controller C_m is designed using an empirically-determined linear dynamic model of the muscle response from p to m. At the start of each experimental session a series of stimulation test signals of PRBS[1] form is applied in open loop at a number of mean levels which cover the ranges required for both dorsiflexor and plantarflexor stimulation. This allows a family of linear models to be identified using a linear least-squares approach [82]. Usually the model with the highest static gain is chosen and used for controller design, i.e. a single linear controller is used over the whole operating range for the muscles. In this work we utilised a pole assignment controller for feedback design, with desired closed-loop poles determined by time-domain specifications.

A number of experimental tests were carried out using the Wobbler apparatus and this stiffness control structure. The tests involved specification of a desired stiffness level, and then variation of the ankle angle either by holding the body fixed and wobbling the feet, or by fixing the feet and manually moving the body through a range of angles. In one test, called "external posture control", a desired reference angle θ_{ref} is generated and displayed on a screen to the experimenter in real time together with the measured inclination angle (the desired angle θ_{ref} is typically a sine wave). The task of the experimenter in this test is to make the posture follow the reference angle as accurately as possible, while the stiffness control structure is active.

[1]Pseudo-Random Binary Sequence.

The results of the study, [9], determined that ankle stiffness control can be achieved with FES of the ankle plantarflexor and dorsiflexor muscles in paraplegic subjects, but that the accuracy of stiffness control is limited by muscle strength, and also depends upon the method used to switch stimulation between the two muscle groups as posture changes. The results showed in particular that stiffness control accuracy in backward-leaning postures is limited by the relatively weak dorsiflexor muscles; beyond a certain angle the stimulation pulsewidth reaches its limit and the ankle stiffness beyond this angle reverts to the intrinsic stiffness corresponding to the given maximal level of muscle activation. In forward-leaning postures, where plantarflexor stimulation was active, stiffness levels of up to 20 Nm/deg were achieved.

We also noted that the stiffness control structure of figure 8 cannot obey the properties of a pure stiffness, except in static conditions. This is because of the dynamic properties of the closed-loop moment controller which is imposed between the desired moment m_{ref} and the actual moment m. Thus, exact stiffness control can be achieved in steady state conditions up to the bandwidth of the moment control loop, but this bandwidth, which depends upon muscle force and fatigue properties, provides a fundamental limitation to achievement of stiffness control.

Of central importance in this study was an assessment of the degree to which ankle stiffness control might facilitate external control of the body. The external posture control tests showed that higher levels of ankle stiffness make it more difficult to perturb the body away from the neutral position. This was based upon estimation and analysis of the external moments m_e applied by the experimenter. The analysis showed that for nominal stiffness values which are less than the critical value $\tilde{m}gl$ (the load stiffness), external control is required to stabilise the subject at all times (i.e. these moments have to be applied in a direction *towards* the body). Conversely, when the desired stiffness is greater than $\tilde{m}gl$, external moments are required in a sense which act *away from* the body in order to actively move the subject away from the neutral position.

The conclusion, therefore, is that ankle stiffness control via FES in paraplegic subjects can generate sufficient stiffness that the task of maintaining an upright posture by application of external forces is eased. By implication, we postulated that FES might be used for ankle stiffness control within a two-link setup for paraplegic standing, where stiffness at the ankle and volitional upper-body activity combine to provide stability. We have therefore developed a system which combines artificial control at the ankle joint (stiffness control) with voluntary upper-body motor control. The results of a study in which this has been evaluated are described in the following.

2.3 Integrated voluntary control

The natural way forward from the constraints of the single-link Wobbler configuration is to allow the upper body to participate in the posture stabilisation task; in paraplegia this may be achieved in a variety of ways including voluntary trunk control, which is the mechanism used here. In the work reported here we have utilised apparatus which may be viewed approximately as a two-link inverted pendulum configuration, where the two free axes of rotation correspond biomechanically to the ankle and hip joints.

We have investigated appropriate artificial control strategies for the lower limbs (ankles) which act in concert with volitional upper-body control inputs. Having previously established the feasibility of ankle stiffness control using FES, and studied its limitations, we have first investigated the simplest possible approach to two-link standing and balancing, which is to combine controlled ankle stiffness with voluntary upper-body action [10, 11]. This work uses apparatus called the multi-purpose rehabilitation frame (MRF) which was designed by

Matjačić [83, 84] (see figure 9). It is important to note that the MRF is mechanically similar to conventional, static standing frames employed during clinical rehabilitation in that passive support is provided to lock the knee and hip joints. The primary difference is that in the MRF the ankle joint is free, which makes the standing dynamic and requires the active participation of the subject in order to achieve balance. The MRF device therefore offers not only standing therapy, but also has the potential for active balance re-training therapy.

The frame provides two degrees of freedom for rotational motion at the ankle joint, thus allowing for body movement in both the sagittal and frontal planes. However, in the work reported here the frame was constrained to motion in the sagittal plane only. Shaft encoders allow for measurement of the ankle joint angle, while the hip joint angle is measured using an ultrasonic motion detection system. Torque on the frame is recorded from hydraulic pressure transducers, while force platforms measure forces at the feet (and therefore ankle torques). The frame is equipped with hydraulic actuators which can provide support or impose perturbations in both planes of motion.

Figure 9: The multi-purpose rehabilitation frame (MRF) with paraplegic subject (level T5) while balancing.

Matjačić *et al.* [85, 86] have utilised the MRF configuration to investigate the natural ankle control strategies employed by intact subjects while standing, and to investigate artificial stiffness control at the ankle axis provided by the high-bandwidth hydraulic actuators (in this situation the intact subjects stand on a platform which is attached to and rotates with the frame in order to eliminate volitional control at the ankles). Their initial work was carried out using the MRF apparatus with motion in the sagittal plane only.

With this 2-link arrangement Matjačić *et al.* [85, 86] have shown that paraplegic and intact subjects (with immobilised ankles) are able to stand for prolonged periods provided that a certain level of stiffness (approximately 10 Nm/deg) is implemented at the ankle axis using the frame's hydraulic actuators. Mihelj *et al.* [87, 88] subsequently used the MRF device to investigate the natural mechanisms of ankle control during perturbed standing in healthy subjects for a range of initial stance postures and perturbations. They showed that for "normal" and anterior initial stance postures the angle-torque relationship was approximately linear, with a constant stiffness value in the range 9–12 Nm/deg. Matjačić has recently studied motion in both the sagittal and frontal planes [83]. Subjects stood on a platform fixed to the frame which rotated about the ankle axis, thus immobilising the ankles. This study shows that paraplegic subjects are capable of stable standing provided that a stiffness level of at least 10 Nm/deg is provided at the ankle axis using the power of the hydraulic actuators,

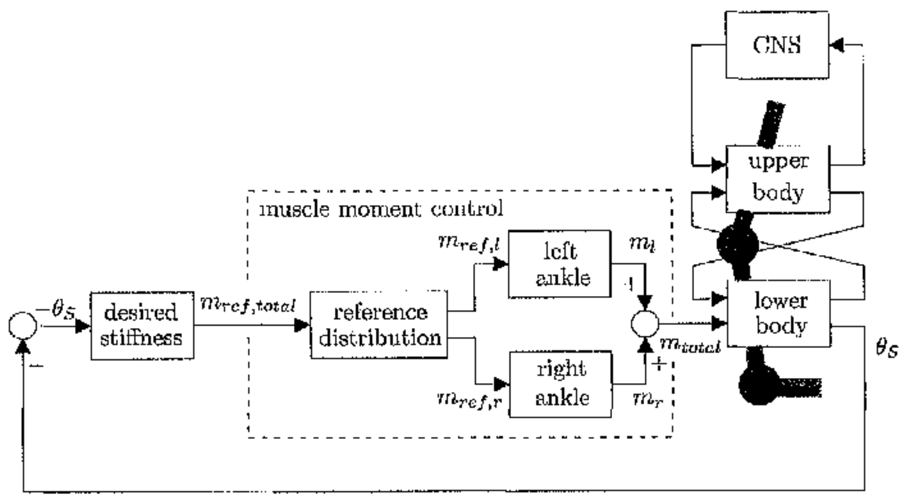


Figure 10: Ankle stiffness control and voluntary upper-body control. The blocks denoted "left ankle" and "right ankle" are closed-loop controllers for left and right ankle moments (see figure 11).

and at the hip joint from volitional control. Finally, Matjačić [89] investigated the natural mechanisms of ankle and hip control during perturbed standing in a group of neurologically intact subjects. He showed that the angle-torque relationships for the ankle and hip joints are approximately linear, with stiffness values at the ankle in the range 13–17 Nm/deg, depending on perturbation direction.

All of this evidence indicates that stiffness is an important variable for characterisation of posture regulation, and that the level of ankle stiffness achieved using FES with paraplegic subjects in our previous study ([9], where up to 20 Nm/deg was attained) might be sufficient to achieve balance in combination with the intact motor skills of the upper body. We therefore conducted a study, reported in [10], to investigate this question.

The control strategy which was implemented is illustrated in figure 10. This consists of a feedback mechanism for control of ankle stiffness which is similar to that previously employed (figure 8), together with volitional stabilising inputs from the upper body under the control of the central nervous system (CNS). In this case, however, the total required ankle moment, computed as the product of desired stiffness and measured ankle angle, is distributed to the left and right ankles according to a load balancing strategy which reflects the relative strength of each side. The blocks labelled "left ankle" and "right ankle" in figure 10 each contain closed-loop controllers for left and right ankle moments. The internal structure of the "left ankle" and "right ankle" blocks is shown in detail in figure 11. For each side, the moment controller C_m is further divided into separate controllers for plantarflexion (superscript "p") and dorsiflexion ("d"). Thus, there is a total of four ankle moment controllers. Switching between the controllers for plantarflexion and dorsiflexion is determined by the sign of the controller output u_l.

The experimental procedure consisted of an open-loop identification test for each muscle group, carried out while the frame was locked to prevent body motion. A local linear dynamic model was identified at the stimulation operating point for each muscle, and then a moment controller was designed for each model using a pole assignment design procedure.

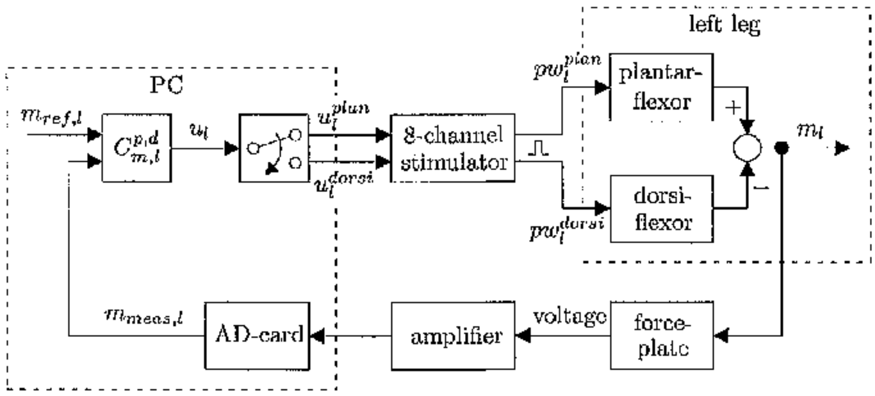


Figure 11: Closed-loop moment control structure for left ankle. The structure for the right ankle is identical, with the subscript "l" replaced by "r".

Following design and testing of the moment controllers the full control structure of figure 10 was implemented and the ability of the subject to retain balance while standing was assessed. The experiments were carried out with a desired ankle stiffness of up to 10 Nm/deg.

The experimental trials revealed that the subject was readily able to maintain upright balance using the residual sensory-motor abilities of the upper body, while supported by FES-controlled ankle stiffness. During a number of trials the stimulation was suddenly switched off at a certain time, at which point balance was immediately lost. The tests also revealed that the subject learned to balance more easily from trial to trial, and during the time each individual trial was in progress. This study confirmed the observation from previous work, [9], that the accuracy of ankle stiffness control, and the ability to achieve a specified level of stiffness, is fundamentally limited by the bandwidth of the moment controllers and, by association, the strength of the muscles. Clearly, when a moment controller's output reaches its upper limit, that particular muscle will not be able to produce the required moment and the specified stiffness will not be achieved. This was frequently observed during the trials. Despite this, the subject was able to adapt his posture in order to successfully maintain balance.

In summary, the study reported in [10] established the feasibility of balance control based on volitional upper-body motion and artificial, FES-induced stiffness at the ankle joint. The study also confirmed the potential of the device, in conjunction with the FES control strategy, to provide an effective means of balance re-training in subjects with postural impairment.

2.4 Discussion

Our work on the automatic control of balance in paraplegic subjects (section 2.1) has established a rigorous system identification and feedback design methodology, suitable for application in experimental trials. These trials represent an important advance because successful control of balance was for the first time achieved with paraplegic subjects. While the laboratory-based apparatus employed in this work may not be suitable for clinical use, it has allowed a fundamental and detailed study of the technical feasibility and limitations of artificial balance control. Future work in this area will continue refinement of feedback control approaches, with a view to maximising robustness against plant uncertainty. It may also be beneficial to consider alternative stimulation patterns (e.g. with variable inter-pulse interval) in an attempt to reduce the rate of muscle fatigue as far as possible so that standing can be prolonged. Publications: [1, 2, 3, 4, 5, 6, 7, 8].

Our study of ankle stiffness control in paraplegic subjects using FES (section 2.2) has established the feasibility of the method and identified physiological limitations. The results of the work are also important because they demonstrate the potential of FES-generated ankle stiffness to aid balance during standing. This research builds upon a large body of work studying the natural control of ankle dynamics during standing in intact subjects, which highlights the key role of stiffness. We found that the levels of ankle stiffness achieved using FES were higher than the values reported in the literature for intact subjects during perturbed standing. Thus, we introduced the possibility of balance with artificial, FES-induced, control at the ankle in combination with volitional motor control inputs from the intact parts of the upper body. Publications: [9].

This theme was pursued in our work on "integrated voluntary control", described in section 2.3. This work established that the combination of FES of the paralysed ankle-actuating muscles and volitional upper-body inputs allows stable, arm-free balance to be achieved in paraplegic subjects. In addition to the provision of dynamic standing therapy, our work highlighted the scope of the approach for retraining of balance during standing in neurologically impaired patients. This work has significant potential for clinical application since the equipment design allows straightforward access for the patient, and is broadly similar to static standing frames currently used in everyday clinical practice. Future work in this area will study higher-order dynamic control at the ankle joint, with a view to optimising the tradeoff between the quality and cost of balance. We also intend to study fully dynamic models of natural control mechanisms operating at the ankle during standing, and to determine whether there is benefit in parameterising the artificial ankle controller to mimic natural dynamics. Publications: [10, 11].

3 Lower-limb Cycling

It was established in the early 1980s that people with a spinal cord injury, including patients with a clinically complete lesion, are able to propel a cycle by means of controlled sequential stimulation of the large leg-actuating muscles [90, 91, 92]. Many subsequent studies of FES-cycling have utilised surface stimulation technology, where adhesive electrodes are attached to the surface of the skin over appropriate muscle motor points. However, FES-cycling by means of an implanted stimulator connected either to peripheral motor nerve electrodes [92], or to electrodes attached to the anterior (motor) lumbo-sacral spinal nerve roots, has also been demonstrated [93].

The feasibility of FES-cycling as an effective exercise option has led over the past twenty years to a large number of studies which have examined possible therapeutic and medical benefits which may help to reduce the general and wide-ranging effects of the secondary complications which often accompany a spinal cord injury. A recent review [94] summarises the key potential benefits: improved muscle size and strength, increased range of joint motion, and significant training effects giving improved cardiopulmonary fitness. In what appears to be the most intensive cycle training intervention reported to date, ten complete-lesion subjects (with a level from C6–T4) completed an average of 2.3 half-hour cycling sessions per week over the course of one year. The outcomes of this study included a significant (almost 20%) increase in peak oxygen uptake [95], and a significant (10%) increase in bone density measured in the proximal tibia using dual energy absorptiometry (DEXA scanning) [96]. It is also notable that despite the rapid muscle atrophy which can occur following paralysis, a programme of FES-cycling exercise may result in at least partial restoration of the depleted slow-oxidative muscle fibre population and re-capillarisation [97].

The focus of our work in this field has been to pursue the technological development of a recumbent FES-cycling system which can be used in a stationary arrangement for fitness training and exercise testing, but which also allows mobile cycling. The engineering design of the system is described below (section 3.1), and its usage within a long-term pilot study with complete-lesion paraplegic subjects is detailed in section 3.2. We have also investigated the application of feedback control methods within FES-cycling. This has been found to be beneficial for training (section 3.3) and in enabling high-sensitivity exercise testing (section 3.4). These developments have found current application in a multi-centre study of the health benefits of a high-intensity FES-cycle training programme, as described further in section 3.5.

Various combinations of muscle groups can be stimulated in order to achieve effective cycling action. Each muscle group is switched on during only part of the 360 deg crank rotation, to ensure that stimulation of each group results in a positive crank moment. This is usually achieved using a continuous measurement of crank angle during cycling. The most common approach to synchronisation of muscle stimulation has been to harness the quadriceps group for knee joint extension, the hamstrings for knee flexion, and the gluteal muscles for hip extension. However, in some studies only a subset of these groups were used, while in other studies additional muscles have been stimulated (a summary of approaches from selected surface-stimulation studies is given in table 1). For example, the addition of the shank muscles has been found to give slightly higher magnitudes of metabolic and cardiopulmonary responses (but the power output was little affected) [98, 99].

Table 1: Stimulated muscle groups or nerves using surface electrodes for FES-cycling in selected studies/systems: Q - quadriceps, H - hamstrings, G - gluteus maximus, I - iliacus, TA - tibialis anterior, GAS - gastrocnemius, P - peroneal nerve.

Source / System	Q	H	G	I	TA	GAS	P	Motor	Flywheel	Type
[91]	×			×					×	stationary
[100]	×	×						×		stationary
[90]	×		×							stationary
Ergys II[2]	×	×	×						×	stationary
Stimmaster Orion[3]	×	×	×						×	stationary
[101]	×		×			×		×		mobile
[102]	×	×	×							mobile
[103]	×	×							×	stationary
[99]	×	×	×		×	×			×	stationary
[104]	×	×							×	stationary
[105]	×	×	×					×		mobile
[106]	×	×	×				×	×		mobile
[107]	×	×	×		×	×				mobile

[2] Therapeutic Alliances Inc., United States, http://www.musclepower.com/
[3] Electrologic of America, Inc., United States, http://www.electrologic.com/

Table 1 further classifies previous FES-cycling systems as either "stationary" or "mobile". Stationary systems are those which serve only as exercise ergometers, while mobile systems can additionally be propelled either indoors or outdoors, and therefore have potential for recreational usage. As noted, several stationary systems, including two commercially-available FES-cycling ergometers,[2,3] are equipped with a flywheel. The flywheel's inertia serves to maintain smooth cycling motion, and to overcome the cycling deadspots.

The mobile systems which have been described are either modifications of commercial cycles (usually tricycles) [90, 102, 107], or they are custom-built [101, 105]. The intensity of muscle stimulation is usually manually selected by the cyclist using a throttle-like interface. Several mobile devices have also been developed which are wheelchairs with FES-cycling adaptations [106, 108, 109].

Several groups have proposed the use of an auxiliary electric motor in combination with a mobile cycle [101, 105, 106]. The primary motivation has been to overcome the limitations posed by muscle fatigue and low power output, in order to maintain reliable propulsion. In all of these approaches the motor and the muscle stimulation are manually controlled by the cyclist. In [105], for example, the total drive power (motor and stimulation) is adjusted by a throttle grip, while the ratio of muscle and motor contributions is pre-selected. Based on this information, electrical stimulation and the electric motor input are controlled in a feedforward manner. In the following, we propose feedback strategies for motor and stimulation control and we analyse their advantages.

3.1 System design and stimulation control

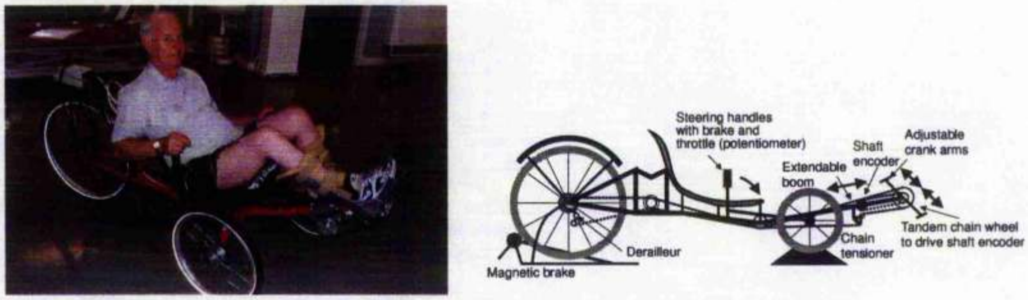
We have utilised a commercially-available recumbent tricycle design[4]. This design was initially proposed by Perkins *et al.*, with appropriate adaptations, for FES-induced paraplegic cycling [107, 93]. The system has been refined as described below and used in our work in two versions: (i) a "basic" (i.e. non-motorised) system, suitable mainly for home training and mobile usage; and (ii) a motorised system with additional instrumentation, which is valuable for exercise testing and investigational purposes.

Apparatus – basic system. The "basic" FES cycle is a self-contained system intended primarily for home training and recreational usage (see figure 12). Ankle orthoses are attached to the pedals. These stabilise the ankle joint and constrain the paralysed legs to motion in the para-sagittal plane. A basic requirement is a measurement of the crank angle. For this purpose, the tricycle is equipped with a shaft encoder. Recently, we have utilised a hollow shaft encoder mounted on the crankshaft (see figure 13). The angular speed of the crank (i.e. the cycling cadence) can be obtained if necessary by differentiation and filtering of the angle measurement. The tricycle has a throttle fitted to the left hand grip. The throttle is interfaced to the stimulation control software and allows the cyclist to manually control the stimulation intensity and thereby to adjust the cycling cadence. The throttle has a switch which when activated holds the stimulation intensity constant at the current throttle setting, thus allowing the throttle to be released. The signals from the shaft encoder and the throttle are interfaced to an electronic stimulator which controls stimulation pulses in the manner described below. Each channel of the stimulator is connected via two wires to a pair of self-adhesive electrodes attached to the surface of the skin over a target muscle group.

This self-contained system can be used for mobile cycling, and it can also be mounted on a cycle trainer for indoor exercise. The cycle trainer used in this work[5] has an electronically-

[4]Inspired Cycle Engineering Ltd., UK, http://www.ice.hpv.co.uk/

[5]Tacx, Holland, http://www.tacx.nl/



(a) Physical apparatus.

(b) Schematic diagram.

Figure 12: Recumbent tricycle and FES-cycling adaptations.

Figure 13: Shaft encoder mounting.

controlled resistance (a motor). A hand-held interface and display unit allows different levels of resistance (load) to be set, and gives an indication of cadence, drive-wheel speed and output power.

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 joint activity generated by each muscle group is illustrated in figure 14. The primary and desirable joint motion generated by stimulation of each muscle group is indicated, while the unwanted "side effects" are shown in parenthesis. Unwanted muscle actions from surface stimulation result from the bi-articular nature of the quadriceps and hamstring muscles. To achieve an effective and smooth cycling motion the muscle groups must be switched on and off at appropriate times during the 360 deg crank cycle. Typical stimulation patterns are shown schematically in figure 15. The primary effects are that the quadriceps muscles produce knee extension, the gluteal muscles give hip extension, and the hamstrings are activated for knee flexion. Each of these effects is timed in order to produce significant positive crank moments. The indicative stimulation patterns shown in figure 15(a) show the positions where significant positive torque is generated by each muscle group under static conditions, i.e. for zero angular velocity. In order to compensate for dynamic muscle latency during cycling, i.e. the time following onset

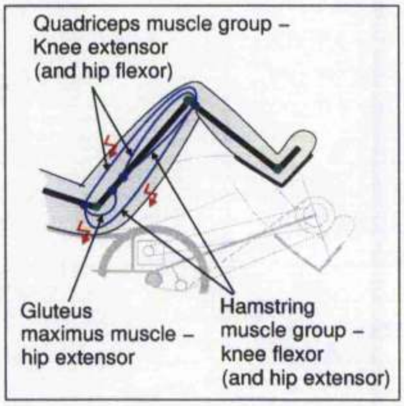
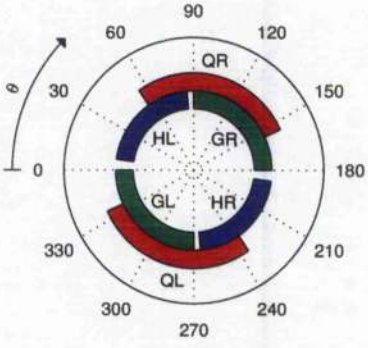
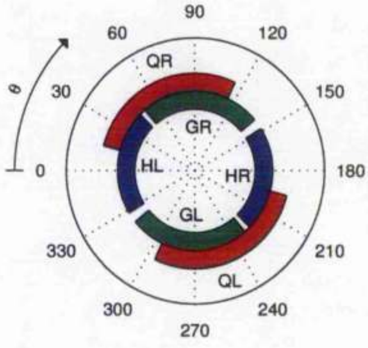


Figure 14: Stimulated muscle groups during FES-cycling using surface electrodes. Desired muscle activity generated by the stimulation is indicated, while unwanted muscle effects are noted in parenthesis.

of stimulation before significant forces are produced, the stimulation patterns have to be shifted forward as cycle cadence increases. This is illustrated in figure 15(b) for a cadence of 50 rpm; the shift angle is directly proportional to cycling cadence.



(a) Static stimulation pattern.



(b) Stimulation pattern at 50 rpm.

Figure 15: Stimulation patterns: QL - Quadriceps Left; QR - Quadriceps Right; HL - Hamstrings Left, HR - Hamstrings Right; GL - Gluteus Left; GR - Gluteus Right.

The shaft encoder and throttle signals are interfaced to a stimulation control program. Stimulation control ensures that the individual muscle groups are automatically switched on and off during cycling using a continuous measurement of crank angle. The stimulation intensity can be set by manual adjustment of the throttle. This control structure is shown schematically in figure 16.

The multi-channel, current-controlled stimulator used in this work is described in [110]. The overall stimulation intensity is determined by setting the amplitude (current), width, and frequency of the pulses. The stimulator allows adjustment of the current in 10 mA steps in the range 0–150 mA, the frequency can be set within the range 20–50 Hz, while the pulsewidth can be continuously set in the range 0–700 μs. In the work reported here, the current for each channel was individually adjusted and then fixed during cycling sessions,

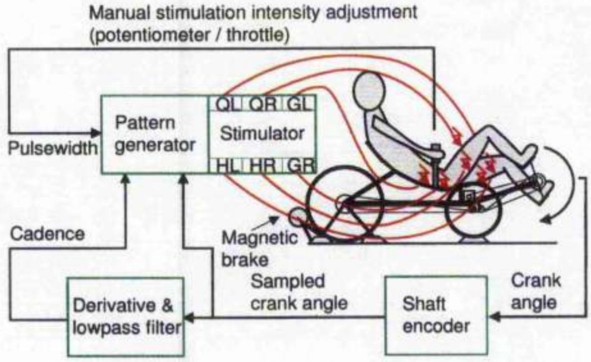
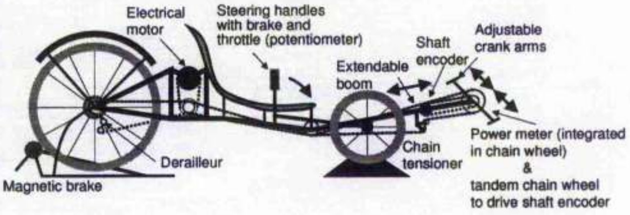


Figure 16: Manual control structure for basic cycling system.

and a constant frequency of 20 Hz was used for each channel (but see [111] for a discussion of cycling performance using higher stimulation frequencies). The stimulation intensity was then varied during cycling by adjustment of the pulsewidth across its full range. The same pulsewidth was applied to each channel.

Apparatus – motorised system. We describe a second FES-cycling system, which is based upon a similar commercially-available recumbent tricycle. In contrast to the basic system, this cycle is equipped with an electric motor and additional instrumentation, which makes the setup more suitable for exercise testing and investigational studies. The stimulation strategy is as described above. The main components of the system are shown in figure 17. An electric motor and battery packs are installed behind the seat. The motor is connected through gearing to the rear drive wheel, and is also coupled to the cranks at the front of the tricycle. Thus, even when no leg power is supplied by the cyclist, the legs are turned by the motor.

(a) Physical apparatus.



(b) Schematic diagram.

Figure 17: Motorised and instrumented recumbent tricycle for FES-cycling.

The electric motor can be used as an optional add-on to the basic FES-cycle system described above, for home and recreational usage. With a simple control strategy to balance the relative power supplied by the motor and the legs, the total output power and the range of the system can be greatly increased.

However, the principal benefit of the motorised tricycle arrangement is the ability, through motor action, to maintain smooth, constant-cadence cycling over an arbitrary range of FES-generated leg-power output levels. With the addition of a leg-power sensor and an integrated control strategy, as described in section 3.4, both cycling cadence and power output can be accurately controlled. Thus, our focus has been on using the motorised and instrumented tricycle for investigational purposes, particularly for exercise testing experiments.

The main instrumentation requirement for control of the cyclist's workrate is a sensor which can measure the leg-power output independently of the action of the electric motor. For this purpose, a torque and power measurement sensor[6] is fitted to the right crank, as shown in detail in figure 18. The sensor operates on the basis of four strain gauges which allow the torque produced by action of the leg muscles to be measured, independent of the influence of the electric motor. The instantaneous leg power is computed as the product of torque and angular velocity.

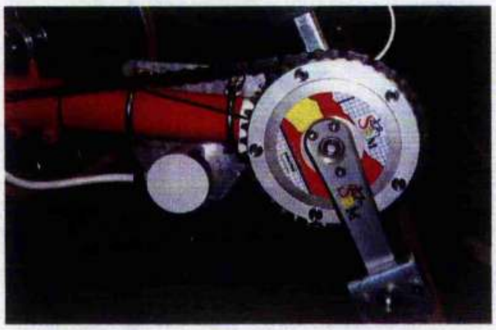


Figure 18: Right crank with torque/power measurement sensor.

For investigational purposes the motorised tricycle is mounted on the cycle trainer described above. The trainer provides a measure of the total mechanical power developed during cycling, which comprises the leg- and motor-power inputs, minus the frictional losses resulting from the tyre-trainer interface. This relationship allows the motor power input to be estimated from the instantaneous values of leg power and total power.

The signals from the shaft encoder, throttle and torque/power sensor are interfaced to analogue channels of a data acquisition card installed in a laptop computer. 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 [pulsewidth] level) and for the electric motor. The PC communicates with the stimulator through the RS232 serial interface, and with the motor through an analogue channel of the data acquisition card. The realtime control software is implemented in Matlab/Simulink,[7] in conjunction with the Real-Time Toolbox.[8]

This software/hardware setup allows a variety of control strategies to be employed, in which a number of variables can be regulated. One option, which combines feedback control of cycling cadence and leg-power output, and which is particularly useful for exercise testing, is described below (section 3.4).

[6]Schoberer Rad Messtechnik (SRM), Germany, http://www.srm.de/

[7]The MathWorks Inc., USA, http://www.mathworks.com/

[8]Humusoft Ltd., Czech Republic, http://www.humusoft.cz/

3.2 Pilot study of mobile FES cycling

Our first study of FES-cycling involving paraplegic subjects was designed as a pilot study to allow evaluation of the engineering design of the tricycles described above, and assessment of the feasibility of mobile cycling [12]. Three paraplegic subjects participated, all with a motor-complete spinal cord lesion at levels ranging from T7 to T10. The subjects initially carried out a programme of muscle strengthening using a portable electronic stimulator at home on a daily basis, for a period of 6–8 weeks. Subjects then began a programme of FES-cycle training. This involved attendance at our clinical unit once per week for a cycling session, while muscle training was carried out at home on all other days.

Two of the subjects progressed to the stage where they were able to cycle continuously and reliably with the basic trike setup on the cycle trainer for periods of up to 1 hour, while sustaining a power output of 15–20 W. This was achieved at a target cadence of 50 rpm. The performance of the third subject was limited by significant levels of spasticity, although he was eventually able to cycle effectively for up to 30 min at a time.

Approximately 3–4 months after the subjects joined the programme, mobile cycling sessions with all three subjects were initiated, on an outdoor tarmac track as depicted in figure 19. Two of the subjects were readily able to complete up to 3 km in a single session.

Figure 19: Paraplegic cyclist during outdoor cycling session.

Each outdoor "session" lasted approximately 30–40 mins and consisted of 10-minute bouts of cycling, each followed by a 5-minute rest period.

Thus, the study [12] verified the utility and reliability of the mobile FES-tricycle design, and established the feasibility of tricycle propulsion even in patients capable of a relatively low workrate, participating in a low-intensity training regime.

3.3 Feedback control of cycling cadence

As noted above, most previous engineering approaches to FES-cycling have utilised manual control of stimulation intensity in order to vary cycling cadence. Where an electric motor has additionally been used, this has typically involved a simple open-loop strategy to balance the energy input from the motor and from the stimulated muscles. With manual control, the cyclist adjusts the stimulation pulsewidth using the throttle, in order to achieve a desired cadence. When the cycle is mounted on the trainer, the total mechanical power output can then be altered by manual variation of the resistance setting and of the cadence, since for a given gear setting total power is proportional to cadence and resistive torque.

However, some authors have previously employed feedback control methods in non-motor-assisted FES-cycling for the purpose of speed control, i.e. for automatic adjustment of stimulation intensity to achieve a specified cycling cadence. With automatic feedback control the cadence (and, indirectly, power) can be regulated much more accurately. Chen *et al.* have used non-model-based PI (proportional-integral) [112] and fuzzy logic [104] methods. In the former approach, the PI parameters were determined from simulation of a feedback loop in which the position of the freely-swinging shank is controlled by quadriceps stimulation, a scenario which bears little similarity to the biomechanical constraints of FES-cycling. The latter approach used a heuristic approach for determination of controller parameters.

Our own contribution to this problem [13, 14] proposed a feedback structure as shown in figure 20. Here, when the switch is held in the position which closes the feedback loop, the cycling cadence is continuously measured and compared with a desired value (setpoint) which is usually programmed into the control software. The actual and desired cadence values are processed by a controller which automatically varies the stimulation pulsewidth to reduce the error between the cadence setpoint and the measured cadence.

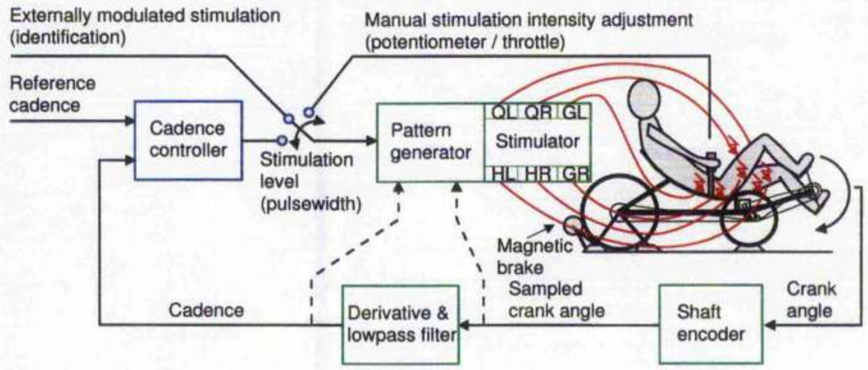


Figure 20: Control structure for the basic cycling system. Depending on the switch position, this structure can be used for manual control of cadence (using the throttle to set the pulsewidth) or for automatic feedback control of cadence. For open-loop identification of pulsewidth–cadence dynamics the switch is placed in the middle position.

Our work employed an approach based on system identification and analytical feedback design. In the identification stage, the cycle cadence response to an open-loop stimulation signal of PRBS form (i.e. a pulsewidth profile applied to all muscle groups during cycling) was measured, and the resulting input-output data were used for model parameter estimation. An analytical feedback design approach (using either pole assignment [13] or LQG methods [14]) then used the estimated model to determine the parameters of the cadence controller. Thus, the control system was directly tuned to the physical system under study. This work showed that very accurate tracking of arbitrary cadence profiles could be achieved, and that significant load disturbances could be rejected by the feedback during constant-cadence cycling [14].

With constant, feedback-controlled cycle cadence, the total output power is manually chosen by increasing or decreasing the variable resistance setting on the trainer. As the resistance is varied, the stimulation level is automatically adjusted to maintain the desired cadence. Alternatively, output power can be varied for a given resistance setting by changing the cadence setpoint.

The primary benefit of feedback control of cycling cadence is that, up to the performance limitations of the muscles, the specified cadence and, for a given load, a desired workrate are automatically obtained (i.e. the stimulation intensity is continuously and automatically adjusted, without user intervention) even when the muscle response varies as a result of fatigue or due to random disturbances. These properties are useful for the achievement of prescribed workrate profiles during training or in exercise testing. However, as described in the following section, there are considerable benefits for exercise testing purposes in using the motorised and fully instrumented tricycle setup.

3.4 Integrated control strategy for motor-assisted cycling: a testbed for physiological studies

Standard methods of exercise testing have been used to quantify the cardiopulmonary adaptations which occur in SCI persons as a result of participation in FES-cycling programmes. Such tests include an "incremental test", where workrate is progressively increased to the subject's physical limit in order to obtain peak workrate and peak oxygen uptake measures and the non-invasive (ventilatory) estimate of the lactate threshold. A further standard test is a sub-maximal, constant-load test which allows estimation of exercise response kinetics (i.e. the time constant of oxygen uptake dynamics during the transient phase of the response) and exercise economy (i.e. the oxygen cost of exercise during steady-state work).

It has been shown that FES-cycling exercise training in SCI subjects can elicit significant improvements in peak oxygen uptake, peak workrate, exercise tolerance and in gas exchange kinetics [113], even when the exercise is performed only two or three times per week.

Despite these improvements, the peak workrate and metabolic efficiency which have been achieved in FES-cycling remain low when compared to able-bodied cyclists. Petrofsky and Stacy [114] reported efficiency values below 4 %. Glaser *et al.* [115] reported net efficiency values (where the resting metabolism is used as baseline correction) of around 5 % in a paraplegic group, and values of up to 20 % in an able-bodied (AB) group performing volitional cycling at the same workrate on the same ergometer.

These poor performance figures are due to several factors. In part, they may result from some characteristics specific to the SCI, such as reduced muscle mass, muscle deconditioning and unfavourable histochemical changes, or partial disruption to sympathetic nervous system control. They will also be affected by the limitations of muscle recruitment via surface FES, including reversed fibre-type recruitment order and muscle spasms, in combination with the unfavourable biomechanics of FES-cycling (e.g. the effect of bi-articular muscles). It has been reported, for example, that during FES-cycling paraplegic subjects apply significantly larger peak forces than AB subjects cycling under voluntary muscle control [116]; AB subjects were able to achieve the same workrate by applying smaller forces over a greater percentage of each crank revolution.

Kjaer *et al.* [117] investigated the performance of eight healthy young males in both volitional and FES-induced cycling. The group first performed volitional cycling at a workrate which elicited an average oxygen uptake of $1.9\ \mathrm{l \cdot min^{-1}}$. Complete epidural anaesthesia was then administered, resulting in total leg paralysis. Cycling was subsequently achieved by means of FES at a workrate which elicited the same oxygen uptake rate of $1.9\ \mathrm{l \cdot min^{-1}}$. The average workrate for this oxygen cost was found to be approximately 120 W during volitional cycling, but less than 40 W for FES-cycling. This establishes that even in subjects where the underlying muscle condition is normal, FES-induced cycling is less efficient by a factor of approximately 3 than cycling achieved by volitional muscle control.

Thus, there is considerable scope for optimising patterns of stimulation to improve cycling performance. While exercise testing methods are crucial in establishing the efficacy of FES-cycling and in monitoring cardiopulmonary adaptations, we further recognise that they will be important in evaluating new strategies for stimulation control which aim to improve cycling performance and efficiency [16, 17].

Most previous studies of the effect of FES-cycling exercise on cardiopulmonary fitness have employed standard pulmonary gas exchange methods, and have utilised commercial, stationary FES-cycling ergometers. Unfortunately, with these devices the exercise workrate and cycling cadence variables are not well controlled. This is because, typically, the control algorithm initially attempts to maintain cadence at some target value (e.g. 50 rpm) but, as fatigue develops and the stimulation level reaches its maximum value, the cadence is allowed to drop as low as 35 rpm before the resistive load is reduced in an attempt to increase cadence. Thus, cadence is allowed to vary over a wide range and, unfortunately, it is not usually possible to record cadence during test sessions with these devices.

Moreover, technical limitations mean that the smallest workrate increment available on these devices is approximately 6 W, with a full workrate range typically of 0–42 W [118]. For many SCI subjects, the magnitude of this increment will be a substantial fraction of their maximal exercise capacity. The discriminatory power of exercise testing for cardiopulmonary and metabolic assessment is thus compromised, both for estimation of parameters of peak performance and also for submaximal kinetic descriptions. This is a consequence of (i) the breath-to-breath "noise" characteristic of gas exchange responses in awake human subjects, imposing a low signal-to-noise ratio [119], and (ii) the limited workrate range constraining the number of discrete workrate increments that can be imposed [120], thus impairing the ability to define accurately the magnitude of any change in functional status within and between subjects. In addition, the limited sensitivity of such exercise tests has made it infeasible to determine, from non-invasive ventilatory measurements, the existence of any lactate threshold in this subject group, and to study the underlying mechanisms.

As workrate in 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 does the metabolic efficiency of the exercise. For example, variations in cadence at a given workrate can influence the oxygen cost of the task, reflecting the varying energetic cost of moving the mass of the legs at different velocities [121] and the muscle fibre-type recruitment profile of the exercising muscles [122]. 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 interest that Theisen *et al.* [123] 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 leg-power output to vary.

These considerations have motivated our own work in this area. An important contribution of our work, described fully in [15], has been the development of a testbed for exercise testing in FES-cycling in which both cycling cadence and workrate are simultaneously well controlled, and in a manner that optimises the ability to undertake accurate characterisation of cardiopulmonary system responses during the exercise. This is achieved using the motorised tricycle setup with: (i) feedback control of stimulation intensity in order to vary leg-power output (exercise workrate); (ii) feedback control of the speed of an auxiliary electric motor, which maintains a desired cadence.

The motorised and instrumented tricycle setup described above (section 3.1) 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 (by adjusting the drive wheel resistance using the electronic brake). An integrated control scheme with two independent feedback loops was described previously [15], and is shown in figure 21. In the first loop, the electric motor input is automatically adjusted in such a way that the cycling cadence is controlled to a reference value by feedback. The setpoint cadence will ordinarily be preprogrammed into the control software. This feedback loop has a relatively high bandwidth as a result of the high-performance motor actuator, and is designed to compensate for other influences which affect the cadence, including load changes and variable energy contribution from the legs resulting from muscle fatigue and the effects of spasticity. 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. Effectively, the total mechanical power output of the rider-tricycle system, which is comprised of the sum of motor power and leg power, is automatically varied in order to maintain the instantaneous value of desired cadence. Independently, the muscle stimulation loop attempts to keep the muscles working at a desired power level, and thus it is the motor power which automatically varies in order to meet the overall instantaneous power requirements.

Thus, the leg-power output, which represents the cyclist's exercise workrate, can be well-controlled to arbitrary values ranging from the negative workrate measured with no stimulation (passive cycling) 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 (i.e. sub-maximal constant-load tests, and incremental tests).

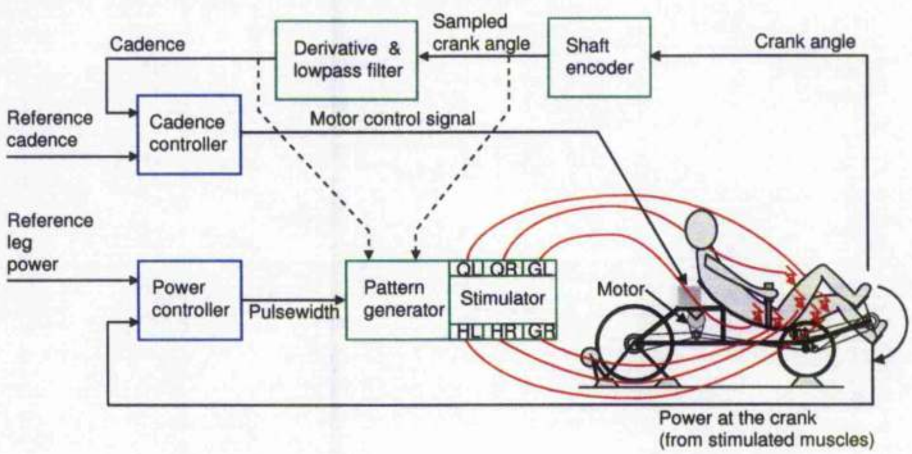


Figure 21: Integrated closed-loop control scheme. One loop automatically adjusts the motor input to keep the cycling cadence close to a setpoint value. The second loop automatically adjusts the stimulation pulsewidth to keep the leg power close to an arbitrary reference value.

An experimental evaluation of this control strategy involving paraplegic subjects demonstrated that accurate feedback control of cycling cadence and exercise workrate can be achieved simultaneously [15]. Tests of feedback control of workrate during exercise tests showed that the subject's leg power output (workrate) can be controlled to an arbitrary reference value, and that 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 reported in [15] 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 pulsewidth 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 perform 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.

Preliminary exercise tests using a portable breath-by-breath cardiopulmonary monitoring system allowed oxygen uptake response kinetics to be estimated (from constant-load tests), and peak responses to be obtained (incremental tests). The high sensitivity of the incremental test (i.e. the ability to impose arbitrarily small workrate increments) produced evidence, for the first time in this SCI subject group, that a ventilatory lactate threshold might be detectable.

Our study concluded that 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 workrate 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 peak oxygen uptake, the lactate threshold, work efficiency and the oxygen uptake time constant). This represents a substantial advance in the SCI population where the maximal exercise workrate is typically substantially compromised.

3.5 Discussion

We have described refinements to the engineering design of an FES-cycling system, based upon the adaptation of commercially-available recumbent tricycles (of various designs), some of which are equipped with an auxiliary electric motor. In general, tricycles have a low centre of gravity, resulting in good lateral stability, and they are convenient for independent transfer by paraplegic persons. The systems are suitable for indoor training at home, or in a clinical setting, and also for exercise testing sessions. Moreover, we have demonstrated mobile outdoor cycling, and mobile cycling in a leisure centre environment, thus opening the way for FES-cycling as a recreation. Publications: [12].

We have carried out work on the feedback control of cycling cadence via automatic adjustment of stimulation intensity in non-motorised cycling. Feedback has been found useful in facilitating convenient and effective training sessions, and for exercise testing where auxiliary motor support is not available (with manual adjustment of workrate at the external load). Publications: [13, 14].

An FES-cycling system has been described which utilises an auxiliary electric motor and additional instrumentation. When used with a novel integrated control strategy, this setup has been shown to be of particular value for exercise testing and for other investigations (e.g. assessment of the efficacy of new stimulation strategies). We achieved test conditions of well-controlled cadence and power output, thus facilitating high exercise-test sensitivity and therefore more reliable estimation of key parameters of aerobic function and cycling performance. Publications: [15, 16, 17].

Our current work involves a longitudinal multi-centre study of FES-cycling utilising the FES-cycling technology described above. This incorporates a health study to examine the following: the cardiopulmonary adaptations resulting from the exercise (using state of the art breath-by-breath metabolic monitoring equipment); the effects on bone density (using quantitative computed tomography [pQCT] at key sites in the tibia and femur); the influence of cycle training on muscle properties including spasticity, muscle bulk and fibre-type composition (using dynamometry and MRI scanning); and the effects on tissue integrity (using seating pressure and tissue oxygenation tests).

4 Upper-limb Exercise in Tetraplegia

This work has investigated the feasibility and therapeutic outcomes of FES-assisted arm-cranking exercise in tetraplegic patients (we focus on those with injury at cervical levels C4–C6, thus involving significant impairment of arm function). The arm-cranking exercise is assisted by FES of the paralysed upper arm muscles.

Spinal cord injury in the cervical (neck) region can result in some degree of paralysis and loss of sensation in the upper limbs. A person with a complete injury at a neurological level of C5 or C6 will generally retain control of the shoulder and the elbow flexor muscles (biceps), but will have no control of the hand or of the elbow extensor muscles (triceps). Muscles which remain under voluntary control may be weakened as a result of incomplete paralysis or due to disuse atrophy caused by limited upper-body exercise. In complete C4 injury control of the entire arm is lost.

The options for partial restoration of lost function include tendon transfer from the posterior deltoid muscle to the triceps, or the use of mechanical orthoses for passive elbow extension [124]. Alternatively, muscle contraction and limb movement may be induced by FES of the paralysed muscles.

Previous investigations of FES in C4–C6 tetraplegia have focused principally on restoration of hand function (e.g. grasp-release) [125]. Following many years of development and clinical trials an implantable hand-function neuroprosthesis has been commercially available. Effort has also been directed towards control of the elbow joint. Miller *et al.* developed a system for elbow extension by triceps stimulation which has sensors measuring elbow and shoulder position [126]. Provision of elbow extension is extremely important since C5–C6 tetraplegics cannot ordinarily perform reaching above the level of the shoulder, where gravity opposes elbow extension. This system was later refined as an incremental extension to the implantable hand-function neuroprosthesis, using a small accelerometer worn on different parts of the arm to sense orientation [127, 128]. Achievement of intermediate elbow angles is made possible by stimulating the triceps sufficiently for full extension, and using voluntary biceps action to oppose extension. It has been shown that triceps FES in C5/C6 individuals improves range of motion, extension strength, and performance of overhead reaching tasks [129]. A number of multi-channel FES systems for finger, wrist and elbow control have also been proposed [130, 131, 132].

Thus, previous FES research for C4–C6 tetraplegics has focused on open-loop systems for hand function and improved working area (i.e. overhead reach), but the provision of upper-limb exercise modalities using FES assistance has been neglected. Our own work in this area has therefore been directed towards a different goal: the development and testing of systems which allow people with C4–C6 tetraplegia to do arm cranking exercise assisted by FES of the paralysed muscles. Our devices allow arm strength and exercise workrate to be measured, and they have feedback control systems which allow precise exercise testing protocols to be administered for determination of key indices of cardiopulmonary status.

In C4 patients we have utilised FES of both the biceps and triceps muscle groups, while C5/C6 patients generally require stimulation of only the triceps, the biceps remaining under voluntary control. While C5/C6 patients are able to do arm cranking without FES, voluntary muscle weakness and the absence of elbow extension torque can limit performance. The addition of triceps via FES can, we propose, greatly improve the quality of cyclic upper-limb motion and may enable the other (voluntary) muscles to do useful work. Thus there is the potential for regular FES-assisted exercise to produce hypertrophy of the otherwise underactive voluntary muscles.

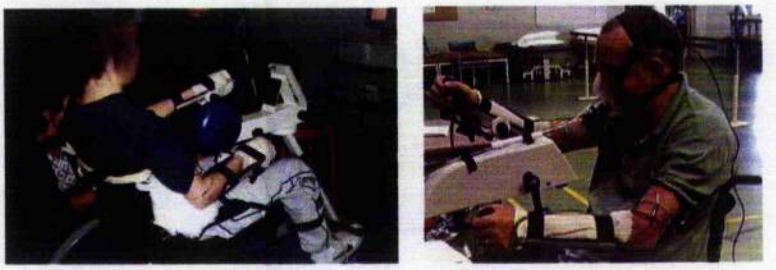
This was confirmed in a previous study [133] which considered a group of incomplete-lesion cervical-level patients having manual muscle scores of at least 1 in at least one triceps muscle group, and a score of 3–5 in the biceps. The subjects received up to eight weeks of FES-assisted arm ergometry, where only the triceps were stimulated. The study reported significant improvement in manual muscle scores after the exercise intervention. A limitation of this work is the uncertainty associated with manual muscle scoring. In addition, this work did not consider changes in cardiopulmonary status or mechanical power output.

Our own work has extended the technical scope of this previous study and suggested that such upper-limb exercise systems may provide enhanced exercise and training options leading to general improvements in the quality of the upper limbs, and possibly also better cardiopulmonary fitness. Through exercise and therapy we anticipate better range of motion, and improvements in the strength of the voluntary shoulder and upper-arm musculature. For some C5/C6 patients, this may be beneficial for daily activities such as transfers, weight shifts, and conventional wheelchair propulsion.

It is of considerable importance to determine whether FES-assisted arm cranking exercise alone can lead to significant improvements in cardiopulmonary fitness. As noted above, there is firm evidence that FES leg cycle ergometry in paraplegia leads to significant improvements in cardiopulmonary fitness, and in several other physiological variables [94]. Moreover, it has been determined that for paraplegic subjects hybrid exercise combining lower-limb FES with voluntary arm cranking provokes significantly higher physiological responses than leg or arm ergometry on their own [134, 135]. However, in tetraplegia, arm-exercise alone may be a more attractive option as this can be performed more conveniently than leg ergometry. It is therefore important in this work to examine the potential cardiopulmonary training effects of FES-assisted arm cranking exercise on its own.

Our initial work in this area developed the engineering methods and apparatus for FES-assisted arm-cranking exercise [19, 21]. This has involved instrumentation of a commercial system for upper-limb exercise and its adaptation for use with FES of the biceps and triceps muscle groups (see figure 22). Methods have been developed for system characterisation and stimulation control. Feedback control systems have been integrated with the system to facilitate accurate cardiopulmonary testing, including constant-load step tests and incremental exercise tests. These control systems and their function are similar to the systems described above for cardiopulmonary testing during lower-limb FES-cycling.

Experimental assessment with tetraplegic subjects has proven the efficacy of biceps/triceps stimulation in producing positive arm-cranking moments. We have shown that a three-month exercise training programme using this system can result in significant improvements in cardiopulmonary fitness (based on metabolic gas exchange measurements) and in arm strength and function [18, 20, 22].



(a) Exercise training session.

(b) Cardiopulmonary testing using breath-by-breath metabolic monitoring system.

Figure 22: Upper-limb exercise system with adaptations for FES of the biceps/triceps muscle groups.

This approach therefore provides a new modality for exercise and therapy offering potential improvements in the quality of the upper limbs, in the execution of several important daily activities, and in general fitness and health.

5 Modelling and Control of Stimulated Muscle

We have carried out basic research into dynamic modelling approaches for electrically stimulated muscle [23, 24, 25, 26], and into robust and nonlinear feedback design methods for control of muscle "outputs" such as force or joint position [26, 27, 28].

In general, the dynamic response of stimulated muscle, including the magnitude and speed of the response, displays nonlinear and time-varying characteristics, and depends upon activation level and the regularity of inter-pulse intervals (or stimulation frequency). The magnitude of the response broadly obeys a saturation-type nonlinearity with respect to activation level (i.e. stimulation intensity). This is because initially, as simulation intensity is increased from zero, the activation threshold of very few motor units will be reached. This is followed by a range of stimulation intensities where increasing numbers of motor units are recruited. Finally, at high levels of stimulation, it is possible that all motor units associated with the target nerve are activated so that a further increase in stimulation level does not elicit increased muscle response.

This broad recruitment characteristic has frequently been represented by a static saturation-type nonlinearity known as the "recruitment curve". A common approach for isometric contraction is to place the recruitment curve in series with a linear dynamic block to obtain the overall muscle response model. This has the structure of a classical Hammerstein model [23]. But the Hammerstein model has the significant limitation that the dynamic part of the muscle response is fixed, i.e. it remains the same for the whole range of activation levels. Physiological considerations indicate, however, that this cannot be the case. With the inverse pattern of recruitment characteristic of FES, it is to be expected that the dynamics of muscle contraction at low stimulation levels will be relatively fast because of the preferential recruitment of fast-twitch muscle fibres. Conversely, one would expect local contraction dynamics to slow considerably at high average levels of stimulation, since in these operating regions slow-twitch fibres will predominate.

These considerations have been investigated in [23], where we found that the dynamic response speed of human muscle decreased dramatically (by a factor of 5) as stimulation level was increased across its full range. We have also demonstrated a large inter-subject variability of both the recruitment nonlinearity and the response dynamics in the ankle plantarflexor muscles in paraplegic subjects [24]. We have subsequently established that model fidelity can be greatly increased by the use of nonlinear model structures [25, 26], but at the price of increased model complexity.

These findings are important and must be taken into consideration in the design of feedback control systems for stimulated muscle responses, such as those used in the approaches described above for control of unsupported standing or for control of cadence and workrate during FES-cycling. Thus, we have developed and experimentally evaluated a robust linear design approach which uses explicit bounds on the expected range of dynamic variation in muscle response [27]. The uncertainty bounds are obtained by an empirical, multi-level identification procedure which establishes dynamic transfer functions at a range of operating points. While only one of these models is utilised as the nominal model for compensator design, the full range of dynamics is incorporated into the dynamic uncertainty bounds, and this allows robust stability to be established. We have also investigated a nonlinear feedback design approach in which a linear compensator is designed for each nominal model in the family of identified response models. The overall nonlinear compensator is then obtained by interpolation of the individual compensator outputs [26]. This work showed that nominal closed-loop performance specifications could be maintained across a wide range of muscle activation levels, albeit at the cost of a greater complexity in the feedback system.

It is important to emphasise the general finding, at least for isometric muscle contraction, that the dynamics of the muscle response are *benign*. This conclusion should be understood from the systems-theoretic viewpoint, based on the observation that the dynamics are generally low-order, stable, and minimum phase. Thus, there are no fundamental limitations (upper or lower bounds) on the achievable closed-loop bandwidth in feedback design for such nominal models, assuming the availability of sufficient actuator power. However, significant limitations to feedback design properties do arise when the contracting muscle is considered as part of a larger biomechanical system where joint motion results from muscle contraction. Examples of this include the problem of knee-joint angle regulation via quadriceps stimulation [28], or the problem considered in detail in section 2.1 of balance control during quiet standing. These problems do involve non-minimum-phase (knee joint) or unstable open-loop (inverted pendulum) dynamics, and therefore display significant feedback design challenges. The challenge is only made more difficult by the nonlinear and time-varying nature of the muscle contraction dynamics, and the presence of reflex-generated disturbances.

The results of this area of research have provided important insights which have been useful in the design of protocols for characterisation of muscle response dynamics (i.e. model structure selection and parameter estimation) and for feedback design, as applied to the specific problems described in sections 2–4.

6 Feedback Control Theory

6.1 Polynomial equation approach

The above work builds upon earlier fundamental contributions to control theory which have utilised the *polynomial equation approach* to controller synthesis. Our feedback design applications within rehabilitation engineering have utilised the results of theoretical work in this area. The polynomial approach stands in contrast to the alternative state-space and Wiener-Hopf methods. The polynomial approach derives from an algebraic framework, thus leading to distinct numerical implementation, and to a number of theoretical advantages.

Initial work extended existing theory to include optimal rejection of measurable disturbances [29, 30, 31] and a completely general formulation of dynamic cost-function weights, but was limited to single-input single-output systems. A simplified formulation was presented with particular reference to self-tuning control implementation, and the method was applied to the practical problem of temperature control in a major control loop of a power station. This work was published as a monograph [32].

The feedforward control solution for measurable disturbances was extended to the multivariable case [33], and a number of original contributions to the algebraic structure of the polynomial matrix solution to the multivariable control problems were derived [34, 35, 36, 37].

Finally, a polynomial-equation solution to the *standard problem* of control theory was obtained, first for the scalar case [38, 39, 40] and then for multivariable systems [41]. The overall status of developments in the field of polynomial methods in control and filtering theory was summarised in an edited volume [42].

6.2 Other contributions in control engineering

Further original contributions have been made in two areas: the application of neural networks to problems in control engineering, and the application of advanced methods of feedback design to a range of practical applications within the automotive industry.

Neural networks were described in a number of novel configurations for the implementation of nonlinear control systems [43, 44, 45, 46]. A survey paper published during this era attracted considerable attention [47]. Further work established the theoretical equivalence of certain classes of neural networks and other nonlinear function approximation schemes [48, 49]. Finally, a link was then established between certain types of neural-network-like structures and traditional gain scheduling approaches to the control of nonlinear systems [50, 51, 52, 53]. With this new approach, a connection was established to feedback design approaches based upon polynomial methods, since the proposed control structure used the network simply to smoothly interpolate between a family of feedback compensators designed using standard methods.

A period of work in the automotive industry provided an opportunity for the practical implementation of several of the novel control approaches described above, and this led to a number of original publications [54, 55, 56, 57, 58, 59, 60].

7 Discussion of Current and Future Research Areas

In this section we discuss the focus of current research in the areas described above, and we outline activities aimed at transferring research outcomes into clinical practice. We also describe new areas of research which we aim to develop in future.

7.1 Current research

Control of Paraplegic Standing. Research on balance control during standing in paraplegia will proceed with two parallel activities: feedback control of balance, and therapeutic balance retraining. In the former area, we continue fundamental research into robust feedback approaches which are able to deal with the specific structure of plant uncertainty arising in this problem. We have also started to investigate practical implementation of the approach using body-worn sensors and control computers, thus freeing the subject from the constraints of laboratory apparatus.

Our research on therapeutic balance retraining for neurologically-impaired patients is based upon the combination of FES of the paralysed ankle-actuating muscles and volitional upper-body inputs. As noted in section 2.4, current fundamental research in this area is focused on dynamic models of postural control activity at the ankle joint. Further evaluation of the approach will be undertaken with a group of spinal-cord-injured subjects and a group of stroke subjects.

It is also of interest to compare the effects on bone density of passive standing and FES-assisted standing. When FES is used on the muscles actuating the ankle and knee joints, significant dynamic forces are likely to be generated in the tibia and femur, and this may improve bone integrity.

Lower-limb Cycling. Current work in this area is focused on a longitudinal study of the health benefits of a high-intensity programme of cycling exercise. We are studying cardiopulmonary adaptations, possible changes in leg-bone density, changes in muscle properties, and the effects on tissue integrity. Establishing these outcome measures is an important step towards clinical acceptance of FES-cycling.

Importantly, we have also demonstrated that patients can accommodate and reliably use the FES-cycling apparatus in their homes on a long-term basis (subjects on the current study have a tricycle at home), and that public recreational facilities are able to accommodate mobile cycling sessions both in parks and within leisure centres. We have an arrangement with Glasgow City Council which provides regular access to Bellahouston Park and Leisure Centre. This arrangement of clinical uptake, and home and leisure usage provides a model which we believe can be replicated more widely to the benefit of patients who wish to use FES-cycling to maintain fitness and achieve health benefits.

The final step which is required to make this model a success, however, is that the equipment for FES-cycling must obtain proper approval as a medical device, and it must be manufactured by a company with the appropriate certification. We have an agreement with a company which is able to provide this, and we will continue to work with them towards the commercial production of a certified FES-cycling system.

Research on the technical aspects of FES-cycling is now focusing on the development of new stimulation patterns which might improve the efficiency of cycling, reduce fatigue rates and achieve better endurance. The term "pattern" is understood here in a general sense to include the active angle ranges for muscle groups, and the actual impulse sequences applied to the muscles. In the first instance we are developing and testing appropriate outcome measures for assessment of alternative patterns [16, 17]. We subsequently intend to study the efficacy of different angle ranges and irregular stimulation sequences using these tools.

Our research studies on FES-cycling have until now included only complete-lesion, adult paraplegic subjects (T3–T12). This has been done in an attempt to achieve homogeneity within the subject groups. However, we wish to expand the availability of FES-cycling to other patients including those with incomplete lesions and those with a cervical-level injury, both within future research studies and for general usage. With incomplete-lesion patients, the possibility arises of utilising EMG signals from muscles which remain under volitional control to better schedule the stimulation of the other muscle groups. We also believe it is important to develop reliable ergometer-based systems for cyclical FES-assisted lower-limb exercise, as a complement to the mobile systems. This option will be much more convenient for clinical and home usage for many patients, particularly those with tetraplegia or poor trunk stability who may have difficulty in transferring to the recumbent device.

Our previous work has utilised surface stimulation of target muscle groups. However, as noted below, a key aim of our proposed work on implanted stimulation technology will be to achieve lower-limb cycling exercise. It will then be important to compare the efficiency of cycling exercise using implanted stimulation with previous results obtained with superficial systems.

We have recently completed the development of a child-sized recumbent FES tricycle, and we have established a new collaboration with Shriners Hospital for Children in Philadelphia, USA. Our immediate goal is to test the feasibility of FES-cycling in the paediatric SCI population, both for stationary ergometry and for mobile cycling. If successful, this may lead to a controlled study of the health benefits of cycling exercise in children with a spinal cord injury.

Arm-cranking Exercise. Our initial work in this area investigated the feasibility and physiological outcomes of arm-cranking exercise in tetraplegic patients, assisted by functional electrical stimulation of the paralysed upper arm muscles. The work encompassed development of arm-crank ergometer devices with integrated stimulator, implementation of feedback systems for accurate control of workrate during exercise testing, development of physiological test protocols for determination of the key parameters of cardiopulmonary status, and an experimental evaluation of the physiological outcomes of a three-month exercise intervention with six volunteer tetraplegic subjects. Experimental assessment with these subjects has proven the efficacy of biceps/triceps stimulation in producing increased arm-cranking moments and power output. We have shown that a three-month exercise training programme using this system can result in significant improvements in cardiopulmonary fitness (based on metabolic gas exchange measurements) and in arm strength and function.

Future work will involve a longitudinal exercise study where training is based in the homes of chronic tetraplegic patients, and periodic assessment of physiological outcomes is carried out in the clinic. We will recruit a subject group of sufficient size to enable analysis of the statistical significance of changes resulting from the exercise in the key parameters of cardiopulmonary fitness and upper-limb function.

In parallel, we will pursue transfer of expertise, know-how and methods arising from the research into the clinical setting. We are working with clinical colleagues to introduce the apparatus and methods developed within research studies into the routine annual medical review of tetraplegic patients. This would involve a short exercise test, administered during the patient's review visit to the clinic, to establish and record the key indices of cardiopulmonary fitness.

We are collaborating with two companies with the aim of progressing to the certification (CE-marking) and commercial production of an FES-assisted arm-crank ergometer device. The availability of a certified device for upper-limb exercise will facilitate the wider clinical uptake of the methods developed and evaluated during our research.

7.2 Future work

Implanted Stimulation Technology. An implanted device for control of bladder and bowel function has been widely used and has met with considerable clinical success [136]. Stimulation electrodes are attached intradurally to the anterior (motor) sacral nerve roots, and hence the device is known as SARSI (sacral anterior root stimulator implant). To be used successfully, a posterior rhizotomy of the S2–S4 sacral nerve roots is usually also performed (i.e. the sensory nerve roots are cut). This eliminates the possibility of reflex incontinence, while coordinated sphincter-detrusor activity during bladder emptying is obtained by stimulation of the anterior S3–S4 roots.

The implanted stimulator used in SARSI has been modified for lower-limb function by Donaldson [137, 138] and evaluated in research studies. Here, the stimulation electrodes are attached to lumbar and sacral motor nerve roots which supply the muscles of the lower limbs (L3–S1). The device is therefore known as LARSI (lumbo-sacral anterior root stimulator implant). Three subjects received a LARSI device, and, despite the poor specificity of muscle responses, it has been found effective in achievement of lower-limb cycling function using recumbent tricycles similar to those described earlier in this thesis.

Donaldson has recently developed a new device based on SARSI/LARSI technology which combines the functionality of both. The new device, known as SLARSI, will therefore offer the following key features: prevention of reflex incontinence (S2–S4 dorsal rhizotomy); voluntary bladder voiding and bowel control (stimulation of S3/S4); and lower-limb exercise (stimulation of L3–S1), which is most likely to be cycling.

A clinical trial of the SLARSI stimulator is planned, and will involve several centres. We aim to be involved in this study as a trial site, providing medical facilities for surgical implantation and clinical follow-up, and engineering expertise required for outcome evaluation.

The need to perform a dorsal rhizotomy appears to be a growing concern in spinal-cord-injured patients for whom the SARSI/SLARSI devices may be clinically indicated. This reluctance probably arises from awareness of the possibility of some late-stage recovery of sensory function, together with the hope for future introduction of surgical interventions as a possible treatment for SCI. While one may reasonably speculate that such treatments will need to be applied in the immediate post-traumatic phase of the SCI, and that improvements in functional outcome may be limited, the fact remains that even patients living today with a long-term SCI will very carefully consider whether they wish to undergo further neurological damage, even for the sake of a clinically-proven device which can offer large improvements in their quality of life today.

There is however the possibility that new research into "neuromodulation" of unwanted reflex activity in the bladder might provide an alternative to the rhizotomy. In this approach, stimulation is applied to the sensory nerve roots in order to block reflex signals which lead to incontinence [139]. Neuroprosthetic devices for bladder function might therefore in future combine neuromodulation of sensory nerves with stimulation of S3/S4 motor pathways for voiding.

As an alternative technological approach, peripheral nerves can be stimulated through local intra-muscular injection of stimulation "pellets" (known as "Bions"), which can be controlled by an external antenna [140]. The most recent versions of the Bion include the capability for sensory recording. This is a rapidly developing technology which we may investigate for application in abdominal muscle stimulation, or for lower-limb exercise.

Treadmill Gait Training. The options available to patients with a spinal cord injury for gait training and walking were discussed in section 1. We are of the view that this type of exercise is most relevant to those incomplete-lesion patients who retain the ability to stand and to take steps under volitional control, perhaps with orthotic support. It is possible that in these patients gait can be improved by judicious use of FES, and if necessary by partial body-weight support. We have initiated such a research study for this patient group. The initial goal is to develop new treadmill-based test protocols for accurate characterisation of cardiopulmonary status, involving precise manipulation of treadmill speed and slope. This work will extend a previous study with normal subjects [141], and tailor the approach to patients with a spinal cord injury. Traditional outcome measures for this type of exercise include walking speed, and distance achieved within a fixed time. Methods for precise characterisation of physiological responses are however lacking. It is also of considerable interest to determine whether this form of cyclical lower-limb exercise intervention has a positive impact on retained voluntary function in these incomplete-lesion patients. Thus, our study will also develop test protocols for characterisation of volitional lower-limb function, using a dynamometer system to obtain muscle responses. In summary, this work aims to contribute rigorous testing protocols for cardiopulmonary status and volitional function, within the context of treadmill gait therapy.

Abdominal Stimulation. We have recently carried out a pilot study of abdominal muscle stimulation for improved tidal volume and cough peak flowrate. One complete-lesion tetraplegic patient (level C4) participated in the study [61, 62]. Patients with paralysed abdominal and chest muscles generally lack an effective cough mechanism and are at increased risk of respiratory infection. Current work in this area is developing triggering mechanisms for automatic control of stimulation during quiet breathing and coughing. Future work will consider the possible application of feedback methods to automatically control stimulation in order to achieve satisfactory levels of physiological variables related to breathing rate and tidal volume, such as blood-oxygen saturation levels. It is possible that this approach might be applied in high-level tetraplegic patients who ordinarily require the support of a mechanical ventilator.

8 Conclusions

This thesis has described original research contributions to the engineering development of systems which aim to restore important function and to enable effective exercise for people with spinal cord injury. An important facet of our work has been the application of feedback control methods; this has been an enabling factor in several areas of study. We have focused on areas which promise improved fitness and general health, and which may alleviate some of the secondary consequences of spinal cord injury. This work encompasses fundamental research, clinical studies, and the pursuit of technology transfer into clinical practice.

We have identified promising areas for future research. It is often the case that it is those people most severely affected by neurological impairment who stand to gain the most from these approaches (e.g. high-level tetraplegia, paediatric spinal cord injury, etc.). We must therefore continue to seek ways in which the work can be developed for the maximum benefit of these patients.

Finally, we recognise the growing awareness of and interest in central nervous system plasticity, and in the broad field of central neural regeneration and repair. It is therefore timely to ask whether cyclical exercise interventions can lead to improvement of volitional function in patients with incomplete or discomplete lesions. Such improvements may, we speculate, result from the strengthening of muscles which retain at least partial volitional control, or from neural plasticity and re-organisation, or from regeneration effects (neurogenesis and functional connectivity). A key requirement in this line of investigation, and a major challenge, will be to develop or to utilise methods which can detect changes in a patient's volitional function and neurological status, and which can isolate the source of such changes [142]. Should reliable methods become available, the way to the study of *recovery* of function through cyclical exercise would be opened. These considerations will remain, we propose, an indispensable complement to cell-based surgical interventions which may become available in the future.

Note: original papers highlighted with the * symbol are included in full within this thesis.

List of Original Contributions

[1] K. J. Hunt, M. Munih, N. Donaldson, and F. M. D. Barr, "Optimal control of ankle joint moment: toward unsupported standing in paraplegia," *IEEE Trans. Automatic Control*, vol. 43, pp. 819–832, June 1998. **This thesis: pp. 51–64.** *

[2] K. J. Hunt, M. Munih, and N. Donaldson, "Feedback control of unsupported standing in paraplegia. Part I: optimal control approach," *IEEE Trans. Rehab. Eng.*, vol. 5, pp. 331–340, December 1997. **This thesis: pp. 65–74.** *

[3] M. Munih, N. Donaldson, K. J. Hunt, and F. M. D. Barr, "Feedback control of unsupported standing in paraplegia. Part II: experimental results," *IEEE Trans. Rehab. Eng.*, vol. 5, pp. 341–352, December 1997. **This thesis: pp. 75–86.** *

[4] K. J. Hunt, H. Gollee, R.-P. Jaime, and N. Donaldson, "Feedback control of unsupported standing," *Technology and Health Care*, vol. 7, no. 6, pp. 443–447, 1999.

[5] K. J. Hunt, H. Gollee, R. Jaime, and N. Donaldson, "Design of feedback controllers for paraplegic standing," *Proc. IEE on Control Theory and Applications*, vol. 148, pp. 97–108, March 2001. **This thesis: pp. 87–98.** *

[6] H. Gollee, K. J. Hunt, and D. Wood, "New results in feedback control of unsupported standing in paraplegia," *IEEE Trans. Neural Sys. Rehab. Eng.*, vol. 12, pp. 73–80, March 2004. **This thesis: pp. 99–106.** *

[7] W. Holderbaum, K. J. Hunt, and H. Gollee, "H-infinity robust control design for paraplegic standing," *Control Engineering Practice*, vol. 10, no. 11, pp. 1211–1222, 2002.

[8] W. Holderbaum, K. J. Hunt, and H. Gollee, "Robust discrete time control design for unsupported paraplegic standing: experimental results," *Eur. J. Control*, vol. 10, no. 3, 2004.

[9] K. J. Hunt, H. Gollee, and R.-P. Jaime, "Control of paraplegic ankle joint stiffness using FES while standing," *Med. Eng. Phys.*, vol. 23, pp. 541–555, 2001. **This thesis: pp. 107–121.** *

[10] R.-P. Jaime, Z. Matjačić, and K. J. Hunt, "Paraplegic standing supported by FES-controlled ankle stiffness," *IEEE Trans. Neural Sys. Rehab. Eng.*, vol. 10, pp. 239–248, December 2002. **This thesis: pp. 122–131.** *

[11] Z. Matjačić, K. J. Hunt, H. Gollee, and T. Sinkjaer, "Control of posture with FES systems," *Med. Eng. Phys.*, vol. 25, pp. 51–62, January 2003. **This thesis: pp. 132–143.** *

[12] K. J. Hunt, T. Schauer, N.-O. Negård, W. Stewart, and M. H. Fraser, "A pilot study of lower-limb FES cycling in paraplegia," in *Proc. 7th Ann. Conf. Int. Functional Electrical Stimulation Soc.*, (Ljubljana, Slovenia), 2002.

[13] K. J. Hunt, M. Rothe, T. Schauer, A. Ronchi, and N.-O. Negård, "Automatic speed control in FES cycling," in *Proc. 6th Ann. Conf. Int. Functional Electrical Stimulation Soc.*, (Cleveland, USA), pp. 300–302, 2001.

[14] T. Schauer, K. J. Hunt, N.-O. Negård, M. H. Fraser, and W. Stewart, "Regelung der Trittgeschwindigkeit beim Liegedreiradfahren von Querschnittsgelähmten (Cadence control for recumbent cycling of paraplegics)," *Automatisierungstechnik*, vol. 50, no. 6, pp. 271–278, 2002. **This thesis: pp. 144–151.** *

[15] K. J. Hunt, B. Stone, N.-O. Negård, T. Schauer, M. H. Fraser, A. J. Cathcart, C. Ferrario, S. A. Ward, and S. Grant, "Control strategies for integration of electric motor assist and functional electrical stimulation in paraplegic cycling: utility for exercise testing and mobile cycling," *IEEE Trans. Neural Sys. Rehab. Eng.*, vol. 12, pp. 89–101, March 2004. **This thesis: pp. 152–164.** *

[16] C. Ferrario, B. Stone, K. J. Hunt, S. A. Ward, A. N. McLean, and M. H. Fraser, "Oxygen cost of different stimulation patterns for FES cycling," in *Proc. 9th Ann. Conf. Int. Functional Electrical Stimulation Soc.*, (Bournemouth, England), pp. 171–173, 2004.

[17] E. S. Idsø, T. A. Johansen, and K. J. Hunt, "Finding the metabolically optimal stimulation pattern for FES-cycling," in *Proc. 9th Ann. Conf. Int. Functional Electrical Stimulation Soc.*, (Bournemouth, England), pp. 239–241, 2004.

[18] K. J. Hunt, A. N. McLean, S. Coupaud, and H. Gollee, "Upper-limb exercise in tetraplegia using functional electrical stimulation," *Advances Clin. Neurosci. Rehab.*, vol. 3, pp. 24–25, Nov/Dec 2003. **This thesis: pp. 165–166.** *

[19] H. Gollee, K. J. Hunt, S. Coupaud, A. N. McLean, and M. H. Fraser, "An apparatus for FES-assisted arm-cranking exercise in tetraplegia," in *Proc. 7th Ann. Conf. Int. Functional Electrical Stimulation Soc.*, (Ljubljana, Slovenia), 2002.

[20] S. Coupaud, H. Gollee, K. J. Hunt, A. N. McLean, and M. H. Fraser, "Physiological assessment of FES-assisted arm cranking exercise," in *Proc. 1st FESnet Conference* (K. J. Hunt and M. Granat, eds.), (Glasgow, UK), September 2002.

[21] S. Coupaud, H. Gollee, K. J. Hunt, S. A. Ward, A. N. McLean, and M. H. Fraser, "Development of methods for arm-cranking exercise using functional electrical stimulation (FES) in tetraplegia," in *Proc. 2nd IEEE EMBSS UKRI PG Conf. Biomed. Eng. Med. Phys.*, (Birmingham, U.K.), pp. 41–42, 2003.

[22] S. Coupaud, H. Gollee, K. J. Hunt, M. H. Fraser, D. B. Allan, and A. N. McLean, "Potential cardiopulmonary benefits of functional electrical stimulation (FES) assisted arm-cranking exercise in tetraplegia," in *Proc. 43rd Annual Scientific Meeting, Int. Spinal Cord Soc.*, (Athens, Greece), 2004.

[23] K. J. Hunt, M. Munih, N. Donaldson, and F. M. D. Barr, "Investigation of the Hammerstein hypothesis in the modelling of electrically stimulated muscle," *IEEE Trans. Biomed. Eng.*, vol. 45, pp. 998–1009, August 1998. **This thesis: pp. 167–178.** *

[24] M. Munih, K. J. Hunt, and N. Donaldson, "Variation of recruitment nonlinearity and dynamic response of ankle plantarflexors," *Med. Eng. Phys.*, vol. 22, no. 2, pp. 97–107, 2000.

[25] N. Donaldson, H. Gollee, K. J. Hunt, J. C. Jarvis, and M. K. N. Kwende, "A radial basis function model of muscle stimulated with irregular inter-pulse intervals," *Med. Eng. Phys.*, vol. 17, pp. 431–441, September 1995.

[26] H. Gollee and K. J. Hunt, "Nonlinear modelling and control of electrically stimulated muscle: a local model network approach," *Int J. Control*, vol. 68, no. 6, pp. 1259–1288, 1997.

[27] K. J. Hunt, R. Jaime, and H. Gollee, "Robust control of electrically-stimulated muscle using polynomial H_∞ design," *Control Engineering Practice*, vol. 9, no. 3, pp. 313–328, 2001.

[28] F. Previdi, T. Schauer, S. M. Savaresi, and K. J. Hunt, "Data-Driven Control Design for Neuroprostheses: A Virtual Reference Feedback Tuning (VRFT) Approach," *IEEE Trans. Control Sys. Tech.*, vol. 12, pp. 176–182, January 2004.

[29] M. Šebek, K. J. Hunt, and M. J. Grimble, "LQG regulation with disturbance measurement feedforward," *Int. J. Control*, vol. 47, pp. 1497–1505, 1988.

[30] K. J. Hunt, "General polynomial solution to the optimal feedback/ feedforward stochastic tracking problem," *Int. J. Control*, vol. 48, pp. 1057–1073, 1988. **This thesis: pp. 179–195.** *

[31] K. J. Hunt, "A single degree of freedom polynomial solution to the optimal feedback/feedforward stochastic tracking problem," *Kybernetika*, vol. 24, pp. 81–97, 1988.

[32] K. J. Hunt, *Stochastic Optimal Control Theory with Application in Self-tuning Control*, vol. 117 of *Lecture Notes in Control and Information Sciences*. Berlin: Springer-Verlag, 1989.

[33] K. J. Hunt and M. Šebek, "Optimal multivariable regulation with disturbance measurement feedforward," *Int. J. Control*, vol. 49, pp. 373–378, 1989. **This thesis: pp. 196–201.** *

[34] K. J. Hunt, M. Šebek, and M. J. Grimble, "Optimal multivariable LQG control using a single diophantine equation," *Int. J. Control*, vol. 46, pp. 1445–1453, 1987.

[35] K. J. Hunt and M. Šebek, "Implied polynomial matrix equations in multivariable stochastic optimal control," *Automatica*, vol. 27, pp. 395–398, 1991. **This thesis: pp. 202–205.** *

[36] K. J. Hunt, "Polynomial optimisation of discrete-time multivariable regulators," *Proc. IEE Pt. D*, vol. 139, pp. 190–196, March 1992.

[37] J. Ježek and K. J. Hunt, "Coupled polynomial equations for LQ control synthesis and an algorithm for solution," *Int. J. Control*, vol. 58, no. 5, pp. 1155–1167, 1993.

[38] K. J. Hunt, V. Kučera, and M. Šebek, "Optimal regulation using measurement feedback: A polynomial approach," *IEEE Trans. Automatic Control*, vol. 37, pp. 682–685, May 1992.

[39] K. J. Hunt and M. Šebek, "Polynomial LQ optimisation for the standard control structure: scalar solution," *Automatica*, vol. 29, pp. 507–509, March 1993.

[40] K. J. Hunt and V. Kučera, "The standard H_2-optimal control problem: a polynomial solution," *Int. J. Control*, vol. 56, no. 1, pp. 245–251, 1992. **This thesis: pp. 206–212.** *

[41] K. J. Hunt, M. Šebek, and V. Kučera, "Polynomial solution of the standard multivariable H_2-optimal control problem," *IEEE Trans. Automatic Control*, vol. 39, pp. 1502–1507, July 1994. **This thesis: pp. 213–218.** *

[42] K. J. Hunt, ed., *Polynomial Methods in Optimal Control and Filtering*, vol. 49 of *IEE Control Engineering Series*. Peter Peregrinus, 1993.

[43] K. J. Hunt and D. Sbarbaro, "Neural networks in process control," *Process Engineering*, vol. May, pp. 59–63, 1991.

[44] K. J. Hunt and D. Sbarbaro, "Neural networks for non-linear Internal Model Control," *Proc. IEE Pt. D*, vol. 138, pp. 431–438, September 1991.

[45] K. J. Hunt and D. Sbarbaro, "Adaptive filtering and neural networks for realisation of Internal Model Control," *Intelligent Systems Engineering Journal*, vol. 2, pp. 67–76, Summer 1993.

[46] D. Sbarbaro, K. J. Hunt, and P. J. Gawthrop, "Designing nonlinear controllers using connectionist networks," *Mathematics and Computers in Simulation*, vol. 40, pp. 657–663, 1996.

[47] K. J. Hunt, D. Sbarbaro, R. Żbikowski, and P. J. Gawthrop, "Neural networks for control systems: a survey," *Automatica*, vol. 28, pp. 1083–1112, November 1992.

[48] K. J. Hunt, R. Haas, and M. Brown, "On the functional equivalence of fuzzy inference systems and spline-based networks," *Int. J. Neural Systems*, vol. 6, pp. 171–184, June 1995.

[49] K. J. Hunt, R. Haas, and R. Murray-Smith, "Extending the functional equivalence of radial basis function networks and fuzzy inference systems," *IEEE Trans. Neural Networks*, vol. 7, pp. 776–781, May 1996.

[50] K. J. Hunt and T. A. Johansen, "Design and analysis of gain-scheduled control using local controller networks," *Int. J. Control*, vol. 66, no. 5, pp. 619–651, 1997.

[51] T. A. Johansen, K. J. Hunt, and H. Fritz, "A software environment for gain scheduled controller design," *IEEE Control Systems Magazine*, vol. 18, pp. 48–60, April 1998.

[52] T. Göttsche, K. J. Hunt, and T. A. Johansen, "Nonlinear dynamics modelling via operating regime decomposition," *IMACS Journal on Mathematics and Computers in Simulation*, vol. 46, pp. 543–550, 1998.

[53] T. A. Johansen, K. J. Hunt, and I. Petersen, "Gain scheduled control of a solar power plant," *Control Engineering Practice*, vol. 8, no. 9, pp. 1011–1022, 2000.

[54] K. J. Hunt, J. C. Kalkkuhl, H. Fritz, and T. A. Johansen, "Constructive empirical modelling of longitudinal vehicle dynamics using local model networks," *Control Engineering Practice*, vol. 4, pp. 167–178, February 1996.

[55] K. J. Hunt, R. Haas, and J. C. Kalkkuhl, "Local controller network for autonomous vehicle steering," *Control Engineering Practice*, vol. 4, pp. 1045–1051, August 1996.

[56] T. A. Johansen, K. J. Hunt, P. J. Gawthrop, and H. Fritz, "Off-equilibrium linearisation and design of gain scheduled control with application to vehicle speed control," *Control Engineering Practice*, vol. 6, pp. 167–180, May 1998.

[57] J. Kalkkuhl, K. J. Hunt, and H. Fritz, "FEM-based neural network approach to nonlinear modeling with application to longitudinal vehicle dynamics control," *IEEE Trans. Neural Networks*, vol. 10, pp. 885–897, July 1999.

[58] K. J. Hunt, T. A. Johansen, J. C. Kalkkuhl, H. Fritz, and T. Göttsche, "Nonlinear speed control design for an experimental vehicle using a generalised gain scheduling approach," *IEEE Trans. Control Sys. Tech.*, vol. 8, pp. 381–395, May 2000.

[59] M. Stämpfle, K. J. Hunt, and J. Kalkkuhl, "Efficient simulation of parameter dependent vehicle dynamics," *International Journal for Numerical Methods in Engineering*, vol. 52, no. 11, pp. 1273–1299, 2001.

[60] C. Preusse, H. Keller, and K. J. Hunt, "Fahrzeugführung durch ein Fahrermodell," *Automatisierungstechnik*, vol. 12, pp. 540–546, Dezember 2001.

[61] H. Gollee, S. Coupaud, K. J. Hunt, D. B. Allan, M. H. Fraser, and A. N. McLean, "A feedback system for automatic electrical stimulation of abdominal muscles to assist respiratory function in tetraplegia," in *Proc. 8th Vienna International Workshop on Functional Electrical Stimulation*, (Vienna, Austria), September 2004.

[62] H. Gollee, S. Coupaud, K. J. Hunt, D. B. Allan, M. H. Fraser, and A. N. McLean, "Automatic electrical stimulation of abdominal muscles to increase tidal volume and cough peak flow rates in tetraplegia," in *Proc. 43rd Annual Scientific Meeting, Int. Spinal Cord Soc.*, (Athens, Greece), September 2004.

General Bibliography

[63] D. Grundy and A. Swain, *ABC of Spinal Cord Injury.* BMJ Publishing, 1996.

[64] B. A. Kakulas, "Neuropathology: the foundation for new treatments in spinal cord injury," *Spinal cord*, vol. 42, no. 10, pp. 549–563, 2004.

[65] A. R. Kralj and T. Bajd, *Functional Electrical Stimulation: Standing and Walking after Spinal Cord Injury.* CRC Press, 1989.

[66] D. J. Ewins, P. N. Taylor, S. E. Crook, R. T. Liczynski, and I. D. Swain, "Practical low cost sit/stand system for mid-thoracic paraplegics," *J. Biomed. Eng.*, vol. 10, pp. 184–188, 1988.

[67] R. J. Jaeger, "Design and stimulation of closed-loop electrical stimulation orthoses for restoration of quiet standing in paraplegia," *J. Biomechanics*, vol. 19, pp. 825–835, 1986.

[68] G. Khang and F. E. Zajac, "Paraplegic standing controlled by functional neuromuscular stimulation: Part I - computer model and control system design," *IEEE Trans. Biomed. Eng.*, vol. 36, pp. 873–884, September 1989.

[69] G. Khang and F. E. Zajac, "Paraplegic standing controlled by functional neuromuscular stimulation: Part II - computer simulation studies," *IEEE Trans. Biomed. Eng.*, vol. 36, pp. 885–894, September 1989.

[70] N. Donaldson, M. Munih, G. F. Phillips, and T. A. Perkins, "Apparatus and methods for studying artificial feedback-control of the plantarflexors in paraplegics without interference from the brain," *Med. Eng. Phys.*, vol. 19, no. 6, pp. 525–535, 1997.

[71] N. Donaldson, F. M. D. Barr, G. F. Phillips, and T. A. Perkins, "Unsupported standing of paraplegics by stimulation of the plantarflexors: some results from the wobbler apparatus," in *Neuroprosthetics: from basic research to clinical application* (A. Pedotti, M. Ferrarin, J. Quintern, and R. Riener, eds.), Springer-Verlag, 1996.

[72] R. C. Fitzpatrick, J. L. Taylor, and D. I. McCloskey, "Ankle stiffness of standing humans in response to imperceptible perturbations: reflex & task-dependent components," *J. Physiol.*, vol. 454, pp. 533–547, 1992.

[73] P. L. Weiss, I. W. Hunter, and R. E. Kearney, "Human ankle joint stiffness over the full range of muscle activation levels," *J. Biomechanics*, vol. 21, no. 7, pp. 539–544, 1988.

[74] R. E. Kearney, R. B. Stein, and L. Parameswaran, "Identification of intrinsic and reflex contributions to human ankle stiffness dynamics," *IEEE Trans. Biomed. Eng.*, vol. 44, pp. 493–504, June 1997.

[75] I. D. Loram, S. M. Kelly, and M. Lakie, "Human balancing of an inverted pendulum: is sway size controlled by ankle impedance?," *J. Physiol.*, vol. 523, no. 3, pp. 879–891, 2001.

[76] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-Perczak, "Stiffness control of balance in quiet standing," *J. Neurophysiol.*, vol. 80, pp. 1211–1221, 1998.

[77] P. G. Morasso and M. Schieppati, "Can muscle stiffness alone stabilize upright standing?," *J. Neurophysiol.*, vol. 83, pp. 1622–1626, 1999.

[78] D. A. Winter, A. E. Patla, S. Rietdyk, and M. G. Ishac, "Stiffness control of balance in quiet standing," *J. Neurophysiol.*, vol. 85, pp. 2630–2633, 2001.

[79] I. D. Loram and M. Lakie, "Human balancing of an inverted pendulum: position control by small, ballistic-like, throw and catch movements," *J. Physiol.*, vol. 540, pp. 1111–1124, 2002.

[80] I. D. Loram and M. Lakie, "Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for stability," *J. Physiol.*, vol. 545, pp. 1041–1053, 2002.

[81] I. D. Loram, N. Maganaris, and M. Lakie, "Paradoxical muscle movement in human standing," *J. Physiol.*, vol. 556, pp. 683–689, 2004.

[82] L. Ljung, *System Identification: Theory for the user*. Prentice Hall, 2nd ed., 1999.

[83] Z. Matjačić, I. L. Johannesen, and T. Sinkjaer, "A multi-purpose rehabilitation frame: a novel apparatus for balance training during standing of neurologically impaired individuals," *J. Rehab. Res. Dev.*, vol. 37, no. 6, pp. 681–691, 2000.

[84] Z. Matjačić, "A multi-purpose rehabilitation frame: an apparatus for experimental investigations of human balance and postural control," *J. Med. Eng. Tech.*, vol. 24, pp. 250–254, 2000.

[85] Z. Matjačić and T. Bajd, "Arm-free paraplegic standing - part I: Control model synthesis and simulation," *IEEE Trans. Rehab. Eng.*, vol. 6, pp. 125–138, June 1998.

[86] Z. Matjačić and T. Bajd, "Arm-free paraplegic standing - part II: Experimental results," *IEEE Trans. Rehab. Eng.*, vol. 6, pp. 139–150, June 1998.

[87] M. Mihelj, Z. Matjačić, and T. Bajd, "Postural activity of constrained subject in response to disturbance in sagittal plane," *Technology and Health Care*, vol. 7, pp. 437–442, 1999.

[88] M. Mihelj, Z. Matjačić, and T. Bajd, "Postural activity of constrained subject in response to disturbance in sagittal plane," *Gait and Posture*, vol. 12, pp. 94–104, 2000.

[89] Z. Matjačić, "Control of ankle and hip joint stiffness for arm-free standing in paraplegia," *Neuromodulation*, vol. 4, no. 1, pp. 37–46, 2001.

[90] J. Petrofsky, H. Heaten, and C. Phillips, "Outdoor bicycle for exercise in paraplegics and quadriplegics," *J. Biomed. Eng.*, vol. 5, pp. 292–296, Oct. 1983.

[91] J. Petrofsky, C. Phillips, H. Heaton, and R. Glaser, "Bicycle ergometer for paralyzed muscle," *J. Clin. Eng.*, vol. 9, pp. 13–19, 1984.

[92] H. Kern, M. Frey, J. Holle, W. Mayr, G. Schwanda, H. Stohr, and H. Thomas, "Functional electrostimulation of paraplegic patients - 1 year's practical application. Results in patients and experiences," *Z.-Orthop.*, vol. 123, pp. 1–12, 1985.

[93] T. A. Perkins, N. Donaldson, N. A. C. Hatcher, I. D. Swain, and D. E. Wood, "Control of leg-powered paraplegic cycling using stimulation of the lumbo-sacral anterior spinal nerve roots," *IEEE Trans. Neural Sys. Rehab. Eng.*, vol. 10, pp. 158–164, September 2002.

[94] T. Janssen, R. Glaser, and D. Shuster, "Clinical efficacy of electrical stimulation exercise training: Effects on health, fitness, and function," *Topics in Spinal Cord Injury Rehabilitation*, vol. 3, no. 3, pp. 33–49, 1998.

[95] T. Mohr, J. L. Andersen, F. Biering-Sorensen, H. Galbo, J. Bangsbo, A. Wagner, and M. Kjaer, "Long term adaptation to electrically induced cycle training in severe spinal cord individuals," *Spinal Cord*, vol. 35, pp. 1–16, 1997.

[96] T. Mohr, J. Podenphant, F. Biering-Sorensen, H. Galbo, G. Thamsborg, and M. Kjaer, "Increased bone mineral density after prolonged electrically induced cycle training of paralyzed limbs in spinal cord injured man," *Calcif. Tissue Int.*, vol. 61, no. 19, pp. 22–25, 1997.

[97] T. Martin, R. Stein, P. Hoeppner, and D. Reid, "Influence of electrical stimulation on the morphological and metabolic properties of paralyzed muscle," *J. Appl. Physiol.*, vol. 72, pp. 1401–1406, 1992.

[98] S. Figoni, M. Rodgers, and R. Glaser, "Effects of electrical stimulation of shank musculature during ES-leg cycle ergometry," *Med. Sci. Sports Exerc.*, vol. 26, p. 77, 1994.

[99] R. Glaser, W. Couch, T. Janssen, J. Almeyda, D. Pringle, S. Collins, and T. Mathews, "A development system to enhance FES leg cycle ergometer technology," *RESNA*, pp. 279–281, June 1996.

[100] K. Eichhorn, W. Schubert, and E. David, "Maintenance, training and functional use of denervated muscles," *J. Biomed. Eng.*, vol. 6, pp. 205–211, July 1984.

[101] D. Pons, C. Vaughan, and G. Jaros, "Cycling device powered by the electrically stimulated muscles of paraplegics," *Med. Biol. Eng. Comp.*, no. 27, pp. 1–7, 1989.

[102] J. S. Petrofsky and J. Smith, "Three-wheel cycle ergometer for use by men and women with paralysis," *Med. Biol. Eng. Comp.*, vol. 30, pp. 364–369, May 1992.

[103] G. C. Jang, J. J. Chen, C. T. Shih, and T. C. Huseh, "Design of FES-cycling system and its stimulation patterns," *Chinese J. Med. Bio. Eng.*, vol. 13, pp. 305–316, December 1993.

[104] J.-J. Chen, N.-Y. Yu, D.-G. Huang, B.-T. Ann, and G.-C. Chang, "Applying fuzzy logic to control cycling movement induced by functional electrical stimulation," *IEEE Trans. Rehab. Eng.*, vol. 5, no. 2, pp. 158–169, 1997.

[105] M. Gföhler, M. Loicht, and P. Lugner, "Exercise tricycle for paraplegics," *Med. Biol. Eng. Comp.*, vol. 36, pp. 118–121, January 1998.

[106] T. Angeli, M. Gföhler, and T. E. L. Rinder, "Tricycle for paraplegics using functional electrostimulation," *Med. Biol. Eng. Comp.*, vol. 37, pp. 326–327, 1999.

[107] T. Perkins, N. Donaldson, R. Fitzwater, G. Phillips, and D. E. Wood, "Leg powered paraplegic cycling system using surface functional electrical stimulation," in *Proc. 7th Int. Workshop on FES*, (Vienna, Austria), 2001.

[108] R. Glaser, J. Bruner, S. Feinberg, and S. Collins, "Locomotion via paralyzed leg muscles, feasibility study for a leg propelled vehicle," *J. Rehab. Res. & Dev.*, vol. 20, pp. 53–61, 1983.

[109] R. B. Stein, D. Roetenberg, S. L. Chong, and K. B. James, "A wheelchair modified for leg propulsion using voluntary activity or electrical stimulation," *Med. Eng. Phys.*, vol. 25, pp. 11–19, January 2003.

[110] G. F. Phillips, J. R. Adler, and S. J. G. Taylor, "A portable programmable eight-channel surface stimulator," in *Proc. Ljubljana FES Conference*, pp. 166–168, 1993.

[111] P. Eser, N. Donaldson, H. Knecht, and E. Stüssi, "Influence of different stimulation frequencies on power output and fatigue during FES-cycling in recently injured SCI people," *IEEE Trans. Neur. Sys. Rehab. Eng.*, vol. 11, no. 3, pp. 236–240, 2003.

[112] J.-J. Chen, C.-T. Shih, D.-G. Huang, N.-Y. Yu, M.-S. Ju, and T.-C. Huseh, "Development of FES-cycling system with closed-loop control," *Chinese J. Med. Biol. Eng.*, vol. 14, no. 3, pp. 195–208, 1994.

[113] T. J. Barstow, A. M. E. Scremin, D. L. Mutton, C. F. Kunkel, T. C. Cagle, and B. J. Whipp, "Changes in gas-exchange kinetics with training in patients with spinal cord injury," *Med. Sci. Sports Exer.*, vol. 28, pp. 1221–1228, 1996.

[114] J. S. Petrofsky and R. Stacy, "The effect of training on endurance and the cardiovascular responses of individuals with paraplegia during dynamic exercise induced by functional electrical stimulation," *Eur. J. Appl. Physiol.*, vol. 64, pp. 487–92, 1992.

[115] R. M. Glaser, S. F. Figoni, and S. P. Hooker, "Efficiency of FNS leg cycle ergometry," in *Proc. 11th Ann. Conf. IEEE Eng. Med. Biol. Soc.*, pp. 961–963, 1989.

[116] P. Sinclair, G. Davis, R. Smith, B. Cheam, and J. Sutton, "Pedal forces produced during neuromuscular electrical stimulation cycling in paraplegics," *Clin. Biomech.*, vol. 11, no. 1, pp. 51–57, 1996.

[117] M. Kjaer, G. Perko, N. H. Secher, R. Boushel, N. Beyer, S. Pollack, A. Horn, A. Fernandes, T. Mohr, S. F. Lewis, and H. Galbo, "Cardiovascular and ventilatory responses to electrically induced cycling with complete epidural anaesthesia in humans," *Acta Physiol. Scand.*, vol. 151, pp. 199–207, 1994.

[118] R. M. Glaser, S. F. Figoni, S. P. Hooker, M. M. Rodgers, B. N. Ezenwa, A. G. Suryaprasad, S. C. Gupta, and T. Mathews, "Efficiency of FNS leg cycle ergometry," in *Proc. 11th Ann. Conf. IEEE Eng. Med. Biol. Soc.*, pp. 961–963, 1989.

[119] N. Lamarra, B. J. Whipp, S. A. Ward, and K. Wasserman, "Effect of interbreath fluctuations on characterizing exercise gas-exchange kinetics," *J. Appl. Physiol.*, vol. 62, pp. 2003–2012, 1987.

[120] J. Roca and B. J. Whipp, eds., *Clinical Exercise Testing*, vol. 2 of *European Respiratory Monographs*. Sheffield: European Respiratory Journals Ltd, 1997.

[121] K. Wasserman, J. E. Hansen, D. Y. Sue, R. Casaburi, and B. J. Whipp, *Principles of exercise testing and interpretation*. Philadelphia: Lippincott, Williams and Wilkins, 1999.

[122] E. F. Coyle, L. S. Sidosis, J. F. Horowitz, and J. D. Beltz, "Cycling efficiency is related to the percentage of type I muscle fibers," *Med. Sci. Sports Exerc.*, vol. 24, pp. 782–788, 1992.

[123] D. Theisen, C. Fornusek, J. Raymond, and G. M. Davis, "External power output changes during prolonged cycling with electrical stimulation," *J. Rehab. Med.*, vol. 34, pp. 171–175, 2002.

[124] V. R. Hentz and C. Leclercq, *Surgical rehabilitation of the upper limb in tetraplegia.* Edinburgh: W. B. Saunders, 2002.

[125] P. H. Peckham and M. W. Keith, "Motor prostheses for restoration of upper extremity function," in *Neural Prostheses* (R. B. Stein, P. H. Peckham, and D. P. Popović, eds.), pp. 162–187, Oxford University Press, 1992.

[126] L. J. Miller, P. H. Peckham, and M. W. Keith, "Elbow extension in the C5 quadriplegic using functional neuromuscular stimulation," *IEEE Trans. Biomed. Eng.*, vol. 36, no. 7, pp. 771–780, 1989.

[127] J. H. Grill and P. H. Peckham, "Functional neuromuscular stimulation for combined control of elbow extension and hand grasp in C5 and C6 quadriplegics," *IEEE Trans. Rehab. Eng.*, vol. 6, pp. 190–199, June 1998.

[128] P. Crago, W. Memberg, M. Usey, M. Keith, R. Kirsch, G. Chapman, M. Katorgi, and E. Perreault, "An elbow extension neuroprosthesis for individuals with tetraplegia," *IEEE Trans. Rehab. Eng.*, vol. 6, pp. 1–6, March 1998.

[129] A. M. Bryden, W. D. Memberg, and P. E. Crago, "Electrically stimulated elbow extension in persons with C5/C6 tetraplegia: A functional and physiological evaluation," *Arch. Phys. Med. Rehabil.*, vol. 81, no. 38, pp. 80—88, 2000.

[130] N. Hoshiyama, A. Naito, M. Yajima, and Y. Handa, "A multichannel FES system for the restoration of motor functions in high spinal cord injury patients: a respiration-controlled system for multijoint upper extremity," *IEEE Trans. Biomed. Eng.*, vol. 36, pp. 754–760, July 1989.

[131] A. Naito, M. Yajima, H. Fukamachi, K. Ushikoshi, Y. Handa, N. Hoshimiya, and Y. Shimizu, "Functional electrical-stimulation (FES) to the biceps brachii for controlling forearm supination in the paralyzed upper extremity," *Tohoku J. Exp. Med.*, vol. 173, pp. 269–273, 1994.

[132] R. H. Nathan and A. Ohry, "Upper limb functions regained in quadriplegia: a hybrid computerised neuromuscular stimulation system," *Arch. Phys. Med. Rehabil.*, vol. 71, no. 6, pp. 415–421, 1990.

[133] B. M. Needham-Shropshire, J. G. Broton, T. L. Cameron, and K. J. Klose, "Improved motor function in tetraplegics following neuromuscular stimulation-assisted arm ergometry," *J. Spinal Cord Med.*, vol. 20, no. 1, pp. 49–55, 1997.

[134] D. L. Mutton, A. M. E. Scremin, T. J. Barstow, M. D. Scott, C. F. Kunkel, and T. G. Cagle, "Physiologic responses during functional electrical stimulation leg cycling and hybrid exercise in spinal cord injured subjects," *Arch. Phys. Med. Rehabil.*, vol. 78, pp. 712–718, July 1997.

[135] T. J. Barstow, A. M. E. Scremin, D. L. Mutton, C. F. Kunkel, T. G. Cagle, and B. J. Whipp, "Peak and kinetic cardiorespiratory responses during arm and leg exercise in patients with spinal cord injury," *Spinal Cord*, vol. 38, pp. 340–345, 2000.

[136] G. S. Brindley, "The first 500 patients with sacral anterior root stimulator implants: general description," *Paraplegia*, vol. 32, pp. 795–805, December 1994.

[137] N. Donaldson, T. A. Perkins, and A. C. M. Worley, "Lumbar root stimulation for restoring leg function: stimulator and measurement of muscle actions," *Artif. Organs*, vol. 21, pp. 247–249, March 1997.

[138] D. N. Rushton, N. Donaldson, F. M. Barr, V. J. Harper, T. A. Perkins, P. N. Taylor, and A. M. Tromans, "Lumbar root stimulation for restoring leg function: results in paraplegia," *Artif. Organs*, vol. 21, pp. 180–182, March 1997.

[139] A. P. Kirkham, S. L. Knight, M. D. Craggs, A. T. Casey, and P. J. Shah, "Neuromodulation through sacral nerve roots 2 to 4 with a Finetech-Brindley sacral posterior and anterior root stimulator," *Spinal Cord*, vol. 40, no. 6, pp. 272–281, 2002.

[140] G. E. Loeb, R. A. Peck, W. H. Moore, and K. Hood, "BION(TM) system for distributed neural prosthetic interfaces," *Med. Eng. Phys.*, vol. 23, pp. 9–18, January 2001.

[141] J. Porszasz, R. Casaburi, A. Somfay, L. J. Woodhouse, and B. J. Whipp, "A treadmill protocol using simultaneous changes in speed and grade," *Med. Sci. Sports Exerc.*, vol. 35, no. 9, pp. 1596–1603, 2003.

[142] P. H. Ellaway, P. Anand, E. M. Bergstrom, M. Catley, N. J. Davey, H. L. Frankel, A. Jamous, C. Mathias, A. Nicotra, G. Savic, D. Short, and S. Theodorou, "Towards improved clinical and physiological assessments of recovery in spinal cord injury: a clinical initiative," *Spinal Cord*, vol. 42, no. 6, pp. 325–337, 2004.