In the Name of God

Assessment of Energy Expenditure During Upper Body Exercise In Paraplegics

A Thesis Submitted for
the Degree of Doctor of Philosophy
in the Faculty of Medicine

by

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AUTHOR'S DECLARATION

All works described in this thesis is my own and has not been presented as a thesis or a part of a thesis for a degree in this or any other university.

All the experiments were performed by me, except where, for testing the paraplegic subjects, two people were required: in these instances data collection was jointly accomplished by myself and Mr J. Wilson. Mr J. Sinclair in the electronic unit helped me diagnose and repair a faulty face mask.

I receive statistical advice from Mr. T. Aitchison (Department of Statistics), but all data analyses were performed by me.

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This project would not have been possible without 20 able-bodied volunteers from students and staff of the university and also, 12 paraplegics (10 of them from the basketball team), who generously gave their time on several occasions. I am grateful to all of them.

It is also a great pleasure to acknowledge the assistance of the following persons:

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Finally, and above all, I should like to mention the invaluable help of my family during the whole period of my study life in universities from 1983; to inspired, assisted and motivated me to wards the complete of my study.

DEDICATION:

to paraplegic patients

SUMMARY

Assessment of energy expenditure is helping the design of wheelchairs and walking aids for paraplegic locomotion. During exercise in able-bodied subjects a redistribution of blood takes place to increase cardiac output and to supply the blood for exercising muscles. But in paraplegics, the impaired sympathetic innervation and lack of muscular pump disturb the redistribution of blood and diminish venous return (Hopman et al., 1992b). In upright arm activity particularly, venous return may be much poorer in paraplegic, than in able-bodied subjects. However, up to now the same indicators as in able-bodied subjects have been used for assessment of energy expenditure in paraplegics. The aim of this project was to assess the energy expenditure during arm crank ergometry and crutch walking in paraplegic and able-bodied subjects.

The experiments can be divided into 4 phases i.e.:

1) assessment of energy expenditure during seated arm crank ergometry in 10 sports-active paraplegics and 20 able-bodied subjects

2) comparison of the effect of posture on the oxygen consumption, heart rate and perceived exertion in 7 sports-active paraplegics and 20 able-bodied subjects

3) A comparison of assessment of energy expenditure between sports and non-sports persons with spinal cord injury during arm crank ergometry

4) assessment of energy expenditure during crutch walking in 5 sports-active paraplegics and 10 able-bodied subjects

Firstly, an arm crank ergometer, modified from a Monark cycle ergometer, was precisely calibrated. Secondly, a face mask was checked and calibrated against the Douglas bag technique for measurement of oxygen consumption.

PHASE 1: Both seated and upright arm crank ergometry were carried out with an incremental protocol at three work rates lasting 5 minutes each. Oxygen consumption was measured by the Douglas bag technique during the last two minutes of each exercise period, heart rate was continuously monitored by Sport Tester and perceived exertion was evaluated at the end of each experiment. Six paraplegics and 6 able-bodied subjects repeated the experiment within 2 weeks for determining the reproducibility of the measured variables.

The results showed that oxygen consumption, heart rate and perceived exertion were each reproducible with less than 5% differences between test and retest trials. Significant correlations of both heart rate and perceived exertion with oxygen consumption were found in both subject groups. In spite of some previous work (e.g., Hopman et al., 1992b), in this study, heart rate responses to seated arm crank ergometry up to 40 watts did not show significant differences between sports-trained paraplegic and able-bodied subjects. This can be explained by the low intensities of exercise, the training level of the subjects and the low level of lesions of the paraplegics.

PHASE 2: The results of upright arm crank ergometry have shown that the posture (seated and upright) and type of subjects (paraplegic and able-bodied) do not have a significant effect on the oxygen consumption and perceived exertion. However, in spite of a non significant difference in the effect of posture on the oxygen consumption, heart rate response to postural change was significantly different between paraplegic and able-bodied subjects. The greater increase of heart rate in paraplegics, in response to upright posture, can be explained as a compensatory response to reduced stroke volume due to lower venous return. In addition, an unstable heart rate during upright arm crank ergometry in the thoracic paraplegic (who had stable heart rate during seated arm crank ergometry at the same power output), made the assessment of energy expenditure by heart rate based methods difficult.

PHASE 3: Sports-active paraplegic subjects were selected in the hope that they would be free of physiological characteristics associated with lack of fitness. The results of tests on 10 sports-active paraplegics were therefore compared with the data of the only 2 patients who were not members of the basketball team but who volunteered to perform seated and upright arm crank ergometry. Because of the small number of non-sports paraplegics no statistical analysis has been carried out. Actually, at the same power output, the means of heart rate and perceived exertion (i.e., means of all 3 work rates for both seated and upright arm crank ergometry) were respectively 6.5% and 20% lower in non-sports than in sports-active paraplegics, whereas the training effect of sports participation would have lead one to expect that they would be lower in the sports group. Since both the non-sports paraplegics had incomplete lesions, it may be suggested that completeness of injury in paraplegics has a more important effect on their physiological responses to exercise than the regularity of their sports participation. However, it must be remembered that the sample size (n=2) is small, and a definite conclusion cannot be drawn.

PHASE 4: Finally, assessment of energy expenditure during walking with axillary crutches and knee-ankle-foot orthoses in paraplegics was compared with that in able-bodied subjects. All subjects walked at their self-selected preferred speed on a figure of eight track. Able-bodied subjects also walked at slow and fast speeds. Oxygen consumption was measured in the last two minutes of exercise. Heart rate was continuously monitored and perceived exertion was evaluated at the end of each exercise. The duration of each exercise was 5 minutes and by determining the walking distance, the velocity was calculated. Two methods were used to express energy efficiency of walking (energy cost, which is expressed as ml O_2 . kg^{-1} . m^{-1} and physiological cost index i.e., increased heart rate per metre).

In able-bodied subjects, stable heart rates were rapidly reached at all 3 speeds. However, the T_6 paraplegic did not show a stable heart rate during 5 minutes crutch walking and in the T_{11} patient, the heart rate rose rapidly to 180 beats . min^{-1}, at which value it plateaued. The significant correlations between all 3 variables (oxygen consumption, heart rate and perceived exertion) with walking speed obtained during crutch walking and the correlation coefficient between heart rate and oxygen consumption during crutch walking was also significant in able-bodied subjects. Preferred speed of crutch walking was lower in thoracic paraplegics than lumbar paraplegics and able-bodied subjects. In thoracic paraplegics, at preferred speed, both energy cost and physiological cost index were higher than those in lumbar paraplegics and able-bodied subjects. Therefore, crutch walking with knee-ankle-foot orthoses is a high energy cost activity in thoracic paraplegics relative to able-bodied subjects. This high energy cost may partially explain the widely-acknowledgea reluctance of such patients to use orthosis. In addition, the unstable heart rate during crutch walking with long leg brace conflicted with the use of physiological cost index for assessment of energy expenditure in thoracic paraplegics.

To conclude, the non significant difference of all 3 measured variables between paraplegic and able-bodied subjects, the reproducibility of measured variables and also, steady condition of heart rate during seated arm crank ergometry have shownthe consistency of measured variables for assessment of energy expenditure during seated arm activities (e.g., wheelchair propulsion) in paraplegics.

On the other hand, limitation of heart rate, unstable heart rate or long times to achieve steady state heart rate during upright activities in thoracic paraplegics, made the assessment of energy expenditure by heart rate based methods e.g., physiological cost index, doubtful and unreliable.

The different responses to these 2 types of arm exercise in thoracic paraplegics could be partly or wholly explained by impaired venous return mechanisms in paraplegics - problems which would be particularly severe in the upright posture.

FIGURES

TABLES

BOX

CHAPTER 1

INTRODUCTION, LITERATURE REVIEWS AND OBJECTIVES

1.1 INTRODUCTION

1.1.1 NATURE OF PROBLEM

All instances of muscle paralysis are tragic to victim, family and friends (Glaser, 1985). "Of the many forms of disability which can beset mankind, a severe injury or disease of the spinal cord undoubtedly constitutes one of the most devastating calamities in human life" (Guttman, 1976).

Spinal cord injury has also been found important from very earliest periods of civilisation. The Edwin Smit Surgical Papyrus was written about 5000 years ago by an Egyptian physician and contains a prescription for a cervical spinal cord lesion due to dislocation or fracture of the spine (Guttman, 1976). However, prior to world war II, 80% of spinal cord injured victims had died within 3 years of injury (Glaser, 1992).

With improved medical management and better understanding of the pathophysiology of spinal cord injury, their life expectancy has now increased (Davis, 1993). Glaser (1992) pointed out that there are more than 200,000 individuals with spinal cord injury in United States and this population is also expected to increase at a rate of approximately 8,000 individuals annually. In addition, spinal cord injury most commonly afflicts persons between the ages of 15 and 28 years (Nash, 1994). Therefore, in recent years, there has been an increased awareness of the problems and needs of spinal cord injured individuals (Glaser, 1992).

1.1.2 ORGANISATION OF NERVOUS SYSTEM

The nervous system can be divided into central and peripheral nervous systems. The central nervous system includes the brain and spinal cord. The peripheral nerves system contains both afferent neurones which travel towards the brain and efferent neurones which travel away from the brain. The efferent neurones can be divided into two main types i.e., somatic (innervate skeletal muscles) and autonomic (innervate smooth muscles). Firing of autonomic neurones is mediated by sympathetic and parasympathetic fibres. The figure 1.1 illustrates the central nervous system and the outflow levels for the somatic and autonomic nervous system.

1.1.3 ANATOMY AND STRUCTURE OF SPINAL CORD

The vertebral column represents a flexible bony structure consisting of 24 independent vertebrae (7 cervical, 12 thoracic and 5 lumbar). They are connected together by articulations and ligaments. There are 5 sacral and 4 coccygeal vertebrae which are fused.

The spinal cord begins above the foramen magnum, where it is continuous with the medulla oblongata of the brain. It terminates below, in the adult, at the level of the lower border of the first lumbar vertebra.

There are 31 segments in the spinal cord (8 cervical, 12 thoracic, 5 lumbar, 5 sacral and 1 coccygeal). Roots C_1 to C_7 leave above the appropriate vertebral body, where as root C_8 exits below the 7th cervical vertebra and the remainder exit below the appropriate vertebral body. The 12 thoracic segments lie within the area covered by the upper nine thoracic vertebrae; five lumbars and the remaining lower

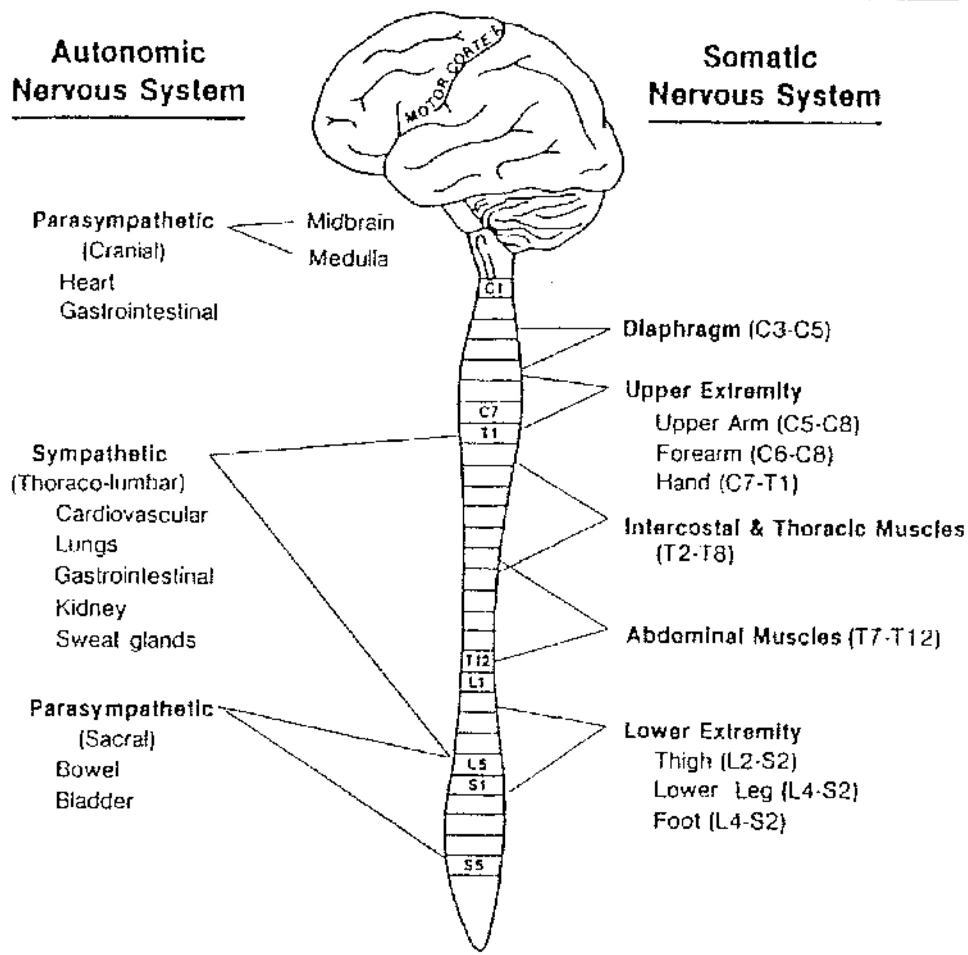

FIGURE 1.1

Diagrammatic illustration of the central nervous systems and outflow levels for the somatic and autonomic nervous system neurons.

(Source: from Glaser, 1989)

segments lie between T_{10} and L_1 vertebrae. These 31 pairs of spinal roots are divided into anterior or motor roots and posterior or sensory roots, and leave the spinal canal through the inter vertebral foramina of the corresponding vertebrae.

The length of the roots increases progressively in a rostrocaudal direction, because of the increasing distance between cord segments and the corresponding vertebral segments. The lumbosacral roots are therefore the longest, and constitute the cauda equina in the lower part of the subarachnoid space.

1.1.4 SPINAL CORD INJURY

When the spinal cord is damaged, there is loss of motor and sensory function below the level of spinal cord lesion. Depending on the location and extent of the injury, paralysis may occur. The various types of paralysis may be classified as follows.

i) Monoplegia (mono = one), i.e., paralysis of one extremity only.

ii) Paraplegia (para = beyond), i.e., paralysis of both lower extremities.

iii) Hemiplegia (hemi = half), i.e., paralysis of one side of the body.

iv) Quadriplegia (quad = four), i.e., paralysis of the two upper and two lower extremities.

Complete transection of the spinal cord (which means that the cord is cut through, transversely) results in a loss of all sensations and voluntary movements below the level of the transection. In addition, the loss of innervation from sympathetic nervous system results in diminished reflex control of blood flow below the level of spinal cord lesion (more detail in section 1.2.2.2).

If the upper cervical cord is transected, quadriplegia results and if the transection is between the cervical and lumbar regions, paraplegia results. It should be taken into consideration that not all injuries to the spinal cord lead to paraplegia and not all paraplegia is caused by trauma. For example, compression of the spinal cord from one side by a tumour may cause monoplegia or hemiplegia. Paraplegia is defined as complete paralysis of both lower limbs and paraparesis is incomplete paralysis of both lower limbs. Instead of paraplegia and paraparesis some authors use the terms complete paraplegia and incomplete paraplegia.

1.1.5 CAUSE OF SPINAL CORD INJURY

The spinal cord may be damaged by traumatic or non traumatic causes. Non-traumatic causes included tumours, multiple sclerosis, motor neurone disease and congenital malformation. Poliomyelitis is also an important cause.

About half of all spinal cord injuries are, in modern societies, attributed to motor vehicle accidents. For example, in the UK between 1984-1988 the causes of spinal cord injury, as categorised by Silver (1993), were: road traffic accidents (52.7%), falls (25.1%), criminal injury (4.7%) and total sports (17.5%). The sporting injuries consisted of swimming (7%), riding (4.7%), rugby (4.1%) and other (1.7%). The survey of Noguchi (1994) in the spinal cord injury resulting from sport in Japan (between 1975 to 1991) have also shown that the swimming (51.4%) is the most common sport causing spinal cord injury. In times of war, of course, the effect of the armed conflict must be added to the statistics. The results from study of 432 wheelchair users (Okhovatian, 1990) in Iran showed that 30%

of these patients had spinal cord injuries from gun shot or mortar-shell fragments during the war.

1.2 LITERATURE SURVEY

1.2.1 INTRODUCTION

Gordon & Vanderwalde (1956) were the first researchers who reported energy expenditure during ambulation in paraplegics. They suggested that patients with spinal cord lesions at T_{12} or above (thoracic paraplegics) would not be able to use a walking aid. Much more recently it was again reported that the currently available orthoses did not offer a realistic means of mobility for paraplegic individuals with injury levels higher than T_{12} (Marsolais et al., 1988). Nevertheless, over recent years, considerable effort has been invested in the development and supply of supportive devices, which enable paraplegics to achieve some mobility in the upright position (Bowker et al., 1992). Energy expenditure therefore is important in assessment of these devices.

In able-bodied subjects, energy expenditure is traditionally determined through the measurement of oxygen consumption. This approach however requires the subject to wear a face mask and either to carry a portable spirometer, or to walk on a treadmill so he can be connected directly to gas analysis equipment. These techniques represent a significant interference with the subject's freedom of movement and are likely to change the usual pattern of gait. This is particularly the case in paraplegics who rely on watching their foot position to help overcome their proprioceptive problems. Some workers have therefore preferred to use heart rate measurements as an indicator of energy expenditure during able-bodied walking (e.g., MacGregor, 1979) and paraplegic crutch walking (Bowker et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993).

With regard to the measurement of oxygen consumption, there are two distinct terms for energy studies concerning locomotion. These are "oxygen consumption" and "energy cost". The first denotes the amount of energy utilised per unit of time and the other indicates the amount of energy utilised to transverse a unit distance. It has long been known that oxygen consumption and heart rate are linearly related at submaximal effort in able-bodied subjects (Astrand & Rodahl, 1986). Therefore, physical effort such as walking in able-bodied subjects can, as another alternative, be assessed by means of the physiological cost index. Physiological cost index is defined as increase of heart rate (i.e., heart rate while walking minus heart rate at rest) divided by walking speed. It is expressed as beats per metre and indicates the efficiency of subject's ambulation.

Physiological cost index has also been used in paraplegics during crutch walking (Bowker, et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993) based on a linear relationship between heart rate and oxygen consumption during seated arm crank ergometry (Bar-On & Nene, 1990). However, according to several studies (Hjeltnes, 1977; Kinzer & Convertino, 1989; Hopman, et al. 1992b; Hopman et al., 1993c), which have recently shown diminished venous return during arm exercise in paraplegics, cardiovascular responses to seated arm activity might not be the same as in upright arm exercise (e.g., crutch walking) in paraplegics.

In the subsequent sections of this chapter, the circulatory regulation of spinal cord injured patients during exercise, and also physiological responses to both seated and upright arm exercise in paraplegic as compared to able-bodied subjects, are reviewed from the literature.

1.2.2 PATHOPHYSIOLOGY OF SPINAL CORD INJURY

This section provides a brief overview of how circulation is regulated and controlled during exercise and how this control is affected by spinal cord injury.

1.2.2.1 CIRCULATORY REGULATION DURING EXERCISE IN ABLE-BODIED

During exercise, blood flow increases in exercising muscles, cardiac muscle and (after the temperature has risen) skin. The increase of blood flow through the vascular beds of these three organs are the result of vasodilatation of arterioles in them. In both skeletal and cardiac muscles, vasodilatation is mediated mainly by local metabolic factors, but dilatation in the arterioles of skin is achieved by a decrease in the sympathetic outflow to the skin. At the same time arteriolar constriction occurs in the inactive organs e.g., inactive muscles, due to increased activity of their sympathetic nerves.

The increase in heart rate during exercise is caused by greater sympathetic activity and less parasympathetic activity to the heart. Cardiac output (heart rate × stroke volume) can be increased to high levels (5-8 times resting, in trained subjects) if venous return to the heart is sufficient. The following factors can have important effects on venous return i.e., i) muscle pump, ii) respiratory pump, iii) venoconstriction in active organs and iv) constriction of arterioles in less active organs.

1.2.2.2 CIRCULATORY REGULATION IN SPINAL CORD INJURY DURING EXERCISE

During world war II, the study of the circulatory problems in paraplegics was begun by Guttman. The findings of Hjeltnes (1977) later showed high arterio-venous oxygen differences and blood lactate concentrations relative to those in able-bodied subjects, during submaximal seated arm crank ergometry in paraplegics. According to these results, it was concluded that the blood flow in the arm muscles of paraplegics is not adequate during arm crank ergometry. The results of Davis & Shephard (1988) have also shown that the cardiac output for a given oxygen consumption is less in paraplegics related to able-bodied subjects during arm exercise. Therefore, both studies used the term "hypokinetic circulation" in paraplegics. They suggested that diminished venous return in paraplegics decreases end-diastolic ventricular volume and thus causes lower stroke volume and cardiac output.

Kinzer & Convertino (1989) investigated cardiovascular responses during arm exercise in able-bodied subjects, paraplegics and amputees i.e., people having normal, paralysed or no leg musculature, respectively. They used the plethysmograph to measure the fluid accumulation in the legs of paraplegic and able-bodied subjects. Their results showed a fluid accumulation in the inactive legs of paraplegics compared to able-bodied subjects during arm exercise. They also showed higher heart rates and lower stroke volume in paraplegics compared with able-bodied subjects and *amputees*. They explained this, too, as being due to fluid accumulation in the paraplegics' legs. Hopman, et al. (1993c) also recorded the leg volume changes, measured by plethysmograph, during arm crank ergometry in paraplegic and able-bodied subjects. Their results showed that the rates of calf volume decrease during exercise were significantly lower in paraplegics than able-bodied subjects. They concluded that the paraplegic subjects are unable to redistribute fluid effectively below the spinal cord lesion during arm exercise.

Various attempts have recently been made to improve venous return below the spinal cord lesion, in order to increase end-diastolic ventricular volume and to enhance stroke volume during arm exercise. Two types of studies were carried out to increase venous return in the legs of spinal cord injuries.

Functional electrical stimulation of the paralysed lower limb muscles was used by Davis et al. (1990) to stimulate and activate the leg muscle pump in paraplegics: stroke volume and cardiac output during arm exercise were increased by the functional electrical stimulation. Another study (Hopman, et al., 1992a) investigated the application of an anti-gravity suit, which puts external pressure on legs and abdomen. This diminishes venous blood pooling below the level of lesion and increases the venous return in paraplegics during arm exercise. Hopman et al. (1992a) reported significantly diminished heart rate and increased stroke volume after applying the external positive pressure. Therefore, both studies strongly suggested the existence of venous pooling below the level of lesion in paraplegics.

This has been explained by two main factors. Firstly, impaired sympathetic innervation, which causes the lack of venoconstriction and vasoconstriction below the level of lesion (Hopman et al., 1992b; Hopman et al., 1993c); and secondly, the inability of paraplegics to activate their muscle pump due to muscular paralysis in the lower extremities (Kinzer & Convertino, 1989; Hopman et al., 1992b).

Therefore, the diminished venous return of blood during exercise is a marked feature in the pathophysiology of paraplegia.

1.2.3 COMPARISON OF PHYSIOLOGICAL RESPONSES TO ARM EXERCISE BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

Owing to the circulatory problems in paraplegics, the physiological responses to arm exercise are likely to differ in detail between paraplegic and able-bodied subjects, especially when they are upright. In the following sections, the physiological responses to arm exercise are compared between paraplegic and able-bodied subjects during both seated (i.e., wheelchair propulsion and seated arm crank ergometry) and upright (i.e., crutch walking and upright arm crank ergometry) arm exercise, separately.

1.2.3.1 SEATED ARM ACTIVITY

Seated arm activity is the most common form of physical activity in paraplegic individuals. In this section, time course of heart rate, relationship between heart rate and oxygen consumption, and physiological responses to seated arm activities e.g., wheelchair propulsion or seated arm crank ergometry, are discussed.

1.2.3.1.1 TIME COURSE OF HEART RATE CHANGES

Steady heart rate during seated arm crank ergometry in paraplegics has been reported by Jehl et al. (1991) and Hopman et al. (1992b) up to 75% and 60% of $\dot{V}_{O_2}$max, respectively.

1.2.3.1.2 RELATIONSHIP BETWEEN HEART RATE AND OXYGEN CONSUMPTION

In some types of activities the measurement of oxygen consumption is difficult (Rodahl et al., 1974). Therefore, according to the linear relationship between heart rate and oxygen consumption during submaximal work rates, heart rate was used for assessment of energy expenditure in able-bodied subjects (Malhotra et al., 1962).

On the other hand, sympathetic outflow to the cardiovascular system arises from thoracic spinal cord level. Therefore, heart rate might not linearly follow the oxygen consumption in paraplegics, particularly when their lesions are high, and this makes the reliability of heart rate responses to exercise doubtful in such patients.

Nevertheless, linear relationships between heart rate and oxygen consumption in paraplegics (with lesions below T_3) have recently been reported by Bar-On & Nene (1990). They have suggested that the cardiac response to an increased demand in physical exercise may be controlled by some other mechanism, which is not clear. This finding provides an important basis for the use of heart rate as an indicator for assessment of energy expenditure during exercise in paraplegics.

1.2.3.1.3 PHYSIOLOGICAL RESPONSES TO SUBMAXIMAL SEATED ARM ERGOMETRY

Similar oxygen consumption in paraplegic and able-bodied subjects has been reported during submaximal arm crank ergometry (at 40 watts) by Hjeltnes (1977). However, several researchers (Hjeltnes, 1977; Kinzer & Convertino, 1989, Hopman et al., 1992b) observed low stroke volumes, compensated by higher heart rates, in paraplegics compared with able-bodied subjects at the same oxygen consumption. At higher exercise intensities, however, heart rate is unable to compensate for the lower stroke volume and cardiac output decreases (Hopman et al., 1992b).

Hjeltnes (1977) and Davis & Shephard (1988) reported lower cardiac output in paraplegics compared with able-bodied subjects at the same oxygen consumption. By contrast, some other researchers (Sawka et al., 1980; Kinzer & Convertino, 1989 and Hopman et al., 1992b) have found the same cardiac output in paraplegic and able-bodied subjects. This discrepancy can partly be explained by different levels of lesions, varying completeness of spinal cord injuries and different subjects' training level (Hopman et al., 1992b).

To my knowledge there is no study which compares the perceived exertion between paraplegic and able-bodied subjects during submaximal arm crank ergometry, although perceived exertion was used during comparisons of maximum oxygen consumption (Eriksson et al., 1988).

1.2.3.2 UPRIGHT ARM ACTIVITY

In upright paraplegics, as already noted circulatory problems are likely to be enhanced due to influence of gravity. In this section, time course of heart rate and physiological responses to crutch walking are compared between paraplegic and able-bodied subjects.

1.2.3.2.1 TIME COURSE OF HEART RATE CHANGES

Annesley et al. (1989) found stable heart rates during crutch walking in able-bodied subjects but some other workers reported non-steady heart rate (Patterson & Fisher, 1981; Hinton & Cullen, 1982; Bhambani & Clarkson, 1989). The reason for this discrepancy is not clear.

In paraplegics, even if heart rate stabilises during crutch walking, the time it takes to do so is important since most patients can not continue walking for long. Although some researchers (Nene & Jenning, 1992; Bowker at al., 1992 & Merkel et al., 1984 & 1985) have reported that the heart rate achieved steady condition in their experiments, no study has shown the time course of heart rate during crutch walking and stated the period of time taken for the stabilisation of heart rate.

In addition, in spite of the use of heart rate based methods for assessment of energy expenditure in paraplegics, no data is available that compared the heart rate responses to crutch walking between paraplegic and able-bodied subjects.

1.2.3.2.2 PHYSIOLOGICAL RESPONSES TO CRUTCH WALKING

Crutches have been used to assist the walking of people with the lower limb disability from very ancient times (Ghosh et al., 1980). Axillary crutch walking is often prescribed for patients with injured lower extremities (Bhambani & Clarkson, 1990). Bhambani & Clarkson (1989) have reported that the physiological demand of crutch walking is considerably higher than that of walking and running in able-bodied subjects. Patterson & Fisher (1981) also recommended the use of the wheelchair in patients with cardiovascular problems because of the extreme physiological responses to crutch walking.

Waters et al. (1985) pointed out that a common reason for paraplegics preferring to use a wheelchair is the slow speed and high energy cost of crutch walking. This high cost has also been shown by other researchers (Cerny et al., 1980; Chantraine et al., 1984; Merkel et al., 1984 & 1985). However, to my knowledge no study has compared the energy cost of crutch walking between paraplegic and able-bodied subjects under the same conditions.

Reduction of the energy cost of walking with orthoses is now an important goal for clinicians and rehabilitation therapists (Nene & Patrick, 1990), and measurement of energy expenditure is an important parameter in the assessment of orthotic devices. The most commonly used physiological parameter for assessment of energy expenditure has been the measurement of oxygen consumption. Heart rate has also been shown to increase linearly with oxygen consumption in seated paraplegics (Bar-On & Nene, 1990; Jehl et al., 1991; Hopman et al., 1992b) as it does in able-bodied subjects (e.g., Astrand & Rodahl, 1986). According to this, heart rate and walking speed measurements were reported as feasible for assessment of energy expenditure (Rose et al., 1991), particularly in the paraplegics, who had problems using the Douglas bag method due to impaired proprioceptive sensation (Winchester et al., 1993).

However, physiological cost index (a heart rate based method) was initially used for assessment of energy expenditure during walking i.e., lower limb activity, and cardiorespiratory responses during arm and leg exercise are different even in able-bodied subjects (e.g., Sawka et al., 1986). Furthermore, due to circulatory problems during arm exercise in paraplegics (Kinzer & Convertino, 1989; Hopman et al., 1993c), cardiovascular responses to seated arm activity might not relate similarly to those during upright arm exercise in paraplegics as compared with able-bodied subjects. There are not, to my knowledge, any publications studing physiological cost index during crutch walking in able-bodied subjects or comparing cardiorespiratory responses to seated and upright arm activities between paraplegic and able-bodied subjects.

1.2.4 SUMMARY

Locomotion is an essential component of most daily activities, and many disabled people are dependent upon manually operated wheelchairs for their locomotion. Clinicians and engineers involved in the rehabilitation of paraplegics with thoracic level of lesions, have long wished to achieve a more effective form of locomotion.

Wheelchair bound paraplegics spend a sedentary life in their wheelchairs and also, for many patients, wheelchair locomotion may elicit cardiorespiratory stresses. Thereby, the demand for physiological data and better understanding of physiological responses to arm exercise in spinal cord injury has arisen.

On the other hand, regulation of cardiovascular adjustments is necessary to support muscular work effectively and previous studies investigated the cardiovascular responses to arm exercise in paraplegics and reported that the redistribution of blood during arm exercise is disturbed. As we have already seen, it is suggested that the lack of sympathetic innervation below the level of the spinal cord lesion and the inability to activate the muscle pump in the legs, result in lower venous return to the heart in the paraplegics compared to able-bodied subjects. In addition, this circulatory problem can be increased in the upright position due to the influence of gravity.

On the other hand, the physiological benefits gained from standing and crutch walking by paraplegics i.e., the prevention of contractures, pressure sores and osteoporosis, and the improvement of kidney and bowel function, are known and documented (Nene, 1994). These factors partly prompted the establishment of the orthoses for paraplegic locomotion.

In other words, walking has always been a dream for most paraplegics and many paraplegics say that, the wish to walk is a high priority for them. Nevertheless, most of them do not use their orthoses after leaving the hospital, probably partly due to the inefficiency of the available types of walking aids. Estimation of energy expenditure therefore becomes important for assessing the efficiency of walking aids. In recent years, there has also been an increased interest in the assessment of energy expenditure during exercise in paraplegics.

The use of oxygen consumption for assessment of energy expenditure is inconvenient in paraplegics, because of their lack of proprioceptivesensation. The use of heart rate based methods (e.g., physiological cost index) therefore becomes more tempting, if it can be reliable. For using heart rate, two factors should be considered in paraplegics i.e., the stabilisation of heart rate and the period of time needed for steady heart rate to be achieved. It is well known that paraplegics are not able to walk for a long period of time due to their low exercise capacity. In this study, the time course of changing heart rates has therefore been compared between paraplegic and able-bodied subjects during crutch walking. The effects of posture on the time course of heart rate responses have also been considered during arm ergometry in paraplegic and able-bodied subjects.

1.3 AIMS AND OBJECTIVES

The aim of this project therefore is the assessment of energy expenditure during seated and upright arm crank ergometry and crutch walking in paraplegic and able-bodied subjects. It can be divided into three parts.

i. Assessment of energy expenditure during seated arm crank ergometry in paraplegic and able-bodied subjects.

ii. Comparison of the effect of posture on the variables, used for assessment of energy expenditure (oxygen consumption, heart rate and perceived exertion), between paraplegic and able-bodied subjects.

iii. Assessment of energy expenditure during crutch walking in paraplegic and able-bodied subjects.

CHAPTER 2

METHODOLOGY

(SUBJECTS, PROCEDURES AND CALIBRATION OF INSTRUMENTS)

2.1 METHODS

2.1.1 SUBJECTS

Twenty able-bodied and 12 paraplegic subjects were studied in the course of the project. Their characteristics are summarised in tables 2.1 and 2.2. The hospital files of paraplegics were consulted for past medical history; their levels of lesions and causes of injuries also indicated in table 2.2.

The main group of paraplegic subjects were members of the West-Scotland and Inverclyde Colts wheelchair basketball teams. They were visited in the Bellahouston (in Glasgow) and Linwood sport centres prior to the start of experiments. At this visit, the aims of experiments were described to them and they were familiarised with the equipment to be used in the physiological testing.

The objective in concentrating on these sports-active paraplegic subjects, for the main part of the project, was in the hope that they would be free of physiological characteristics due simply to unfitness - a complication which seems likely to have been present in many of the paraplegic subjects of previous studies.

What was not foreseen was that this sports-active group would also prove the most willing to volunteer. Only 2 patients who were not members of the basketball team actually got to the laboratory during the period of some 6 months, near the end of the project which had been set aside for "ordinary paraplegic" controls.

Paraplegic subjects	Age (years)	Mass (kg)	Height (cm)	Age of lesion (years)	Cause of injury (years)	Level of lesion
1 †	44	75	160	25	TB of spine	T_5,incomplete
2 †‡	35	73	178	15	Traumatic(sports)	T_5,complete
3 †‡*	45	83.5	172.5	8	Traumatic(traffic)	T_6, complete
4 †‡	21	70	165	21	Spina bifida	T_8, complete
5	27	95	170	8	Traumatic(sports)	T_8, complete
6 †‡*	23	59.6	186	3.5	Traumatic(sports)	T_{11},complete
7 †	19	90	165	19	Spina bifida	L_1,incomplete
8 †‡*	37	81.6	175	11	Traumatic(industry)	L_1,incomplete
9 †‡*	37	80.5	161	37	Spina bifida	L_2,incomplete
10 †‡*	47	58	157.7	45	Polio.	L_2,incomplete
Mean	33.5	76.6	168.7	19.3		
SE	3.2	3.8	2.9	4.2		
11 †‡^	30	94	190	14	Operation	C_6,incomplete
12 †^	23	63	165	6	Traumatic(traffic)	L_2,incomplete

† Volunteer in seated arm crank ergometry
‡ Volunteer in upright arm crank ergometry
*** Volunteer in crutch walking**
^ Non-sports paraplegic

TABLE 2.1

Clinical details of paraplegic subjects

Able-bodied subjects	Age (years)	Mass (kg)	Height (cm)
1 †‡*	27	62	172
2 †‡*	22	73.8	182
3 †‡*	21	65.3	176.3
4 †‡*	24	84.4	186
5 †‡*	31	82.4	184
6 †‡*	29	90	188
7 †‡*	35	84.8	186
8 †‡*	27	66.7	174.2
9 †‡*	27	93.5	191.9
10 †‡*	24	70.7	177
11 †‡	25	67.5	178.9
12 †‡	24	57.5	166
13 †‡	21	71	182
14 †‡	20	75	183
15 †‡	20	97.5	191
16 †‡	22	67.3	175.4
17 †‡	20	62.5	171
18 †‡	27	82	189
19 †‡	24	61.8	172.1
20 †‡	27	79.5	177.2
Mean	24.8	74.8	180.2
SE	0.89	2.56	1.62

† Volunteer in seated arm crank ergometry
‡ Volunteer in upright arm crank ergometry
* Volunteer in crutch walking

TABLE 2.2
Details of able-bodied subjects

2.1.2 PROCEDURE OF ARM CRANK ERGOMETRY EXPERIMENT

1. The arm crank ergometer used was a standard, friction-belt bicycle ergometer of the weight-loaded type (Monark), on which the pedals had been replaced by cylindrical handles (figure 2.1). This ergometer was mounted on rigid frames with different heights (for seated and upright arm crank ergometry, separately). A standard wheelchair was used, and its height was adjusted for each individual. In upright arm crank ergometry, the heights of the volunteers were also adjusted relative to the ergometer. Our criterion for this adjustment (in the vertical plane) in both seated and upright arm crank ergometry was to set the ergometer crank shaft level with subject's shoulder joint. The ergometer position could also be adjusted in the horizontal plane (figure 2.2). The ergometer's internal dynamic friction at 50 rpm was evaluated (see section 2.2.4.2 for calibration details).

For upright arm crank ergometry, the paraplegics used their own callipers and were supported by a frame using straps around the knee and waist. The frame was built in such a way that it would not hinder the movement of the arms during arm cranking. The same equipment was adapted for able-bodied subjects except that no callipers were used.

2. Each subject fasted (food, alcohol and tobacco) at least 2 hours before data collection and took no medication. The temperature in the laboratory was 20-22°C. Before the experiment, the weight and the height of subjects were measured. The paraplegic subjects were weighed in the seated position on a hospital scale.

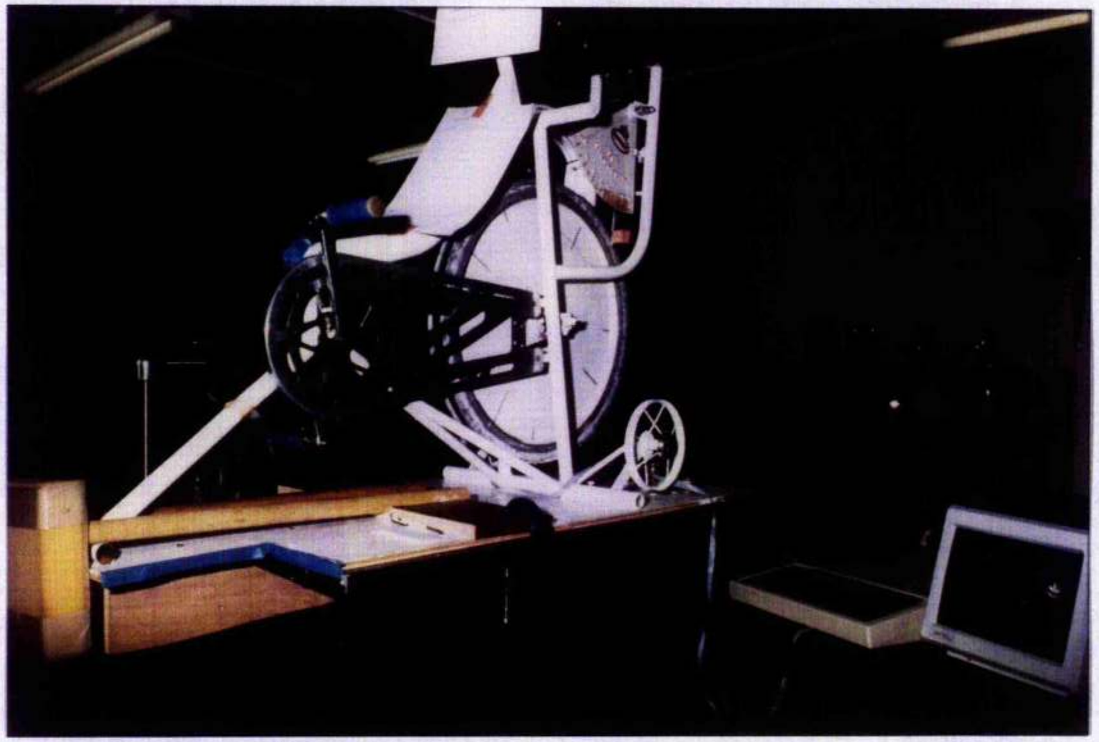

FIGURE 2.1

Arm crank ergometer

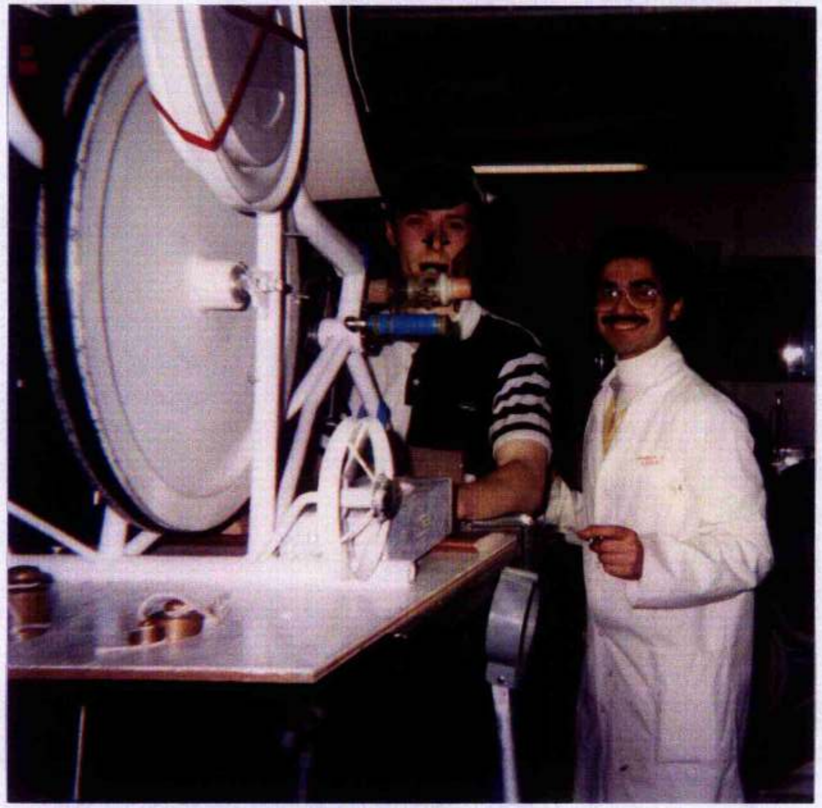

FIGURE 2.2

A paraplegic volunteer (T_{11}) during uptight arm crank ergometry

3. Oxygen consumption was measured by the Douglas bag technique. Collected volumes were measured by a Parkinson Cowan gas meter; O_2% was measured by a paramagnetic oxygen analyser (Servomax type 570A) and CO_2% by infrared absorption (Beckman L3-2). The calibration of both gas analysers and the ergometer was checked before and after every session. On no occasion throughout the project was there a shift in work rate settings which the eye could detect. Therefore the standard error of work rate was lower than the diameter of the graphical symbols. The volume meter was regularly calibrated against a Tissot spirometer. Heart rates were measured by PE 3000 Sports Tester with 15 seconds sampling rate and perceived exertion was evaluated by standard 6-20 Borg scale.

4. Before each exercise test, subjects familiarised themselves with the equipment by arm cranking for 2 minutes against zero load, in the body position in which the test was to be performed. Then, after a few minutes, they rested for 5 minutes in the same position. Heart rate was recorded throughout this rest period and during the last 2 minutes expired air was collected for analysis.

5. An incremental series of 3 work rates was then completed. The real work rates adopted were 16, 28 and 40 watts. These were achieved in the presence of the 4.7 watts internal friction load, by setting the adjustable belt load to 11-12, 23-24 and 35-36 watts, respectively. The chosen work rates were found in a pilot study to be within the capacity of all subjects. Cranking rate was standardised at 50 rpm, monitored by a counter, and the number of crank revolutions per minute was displayed continuously. Each work rate was maintained for 5 minutes. Heart rate

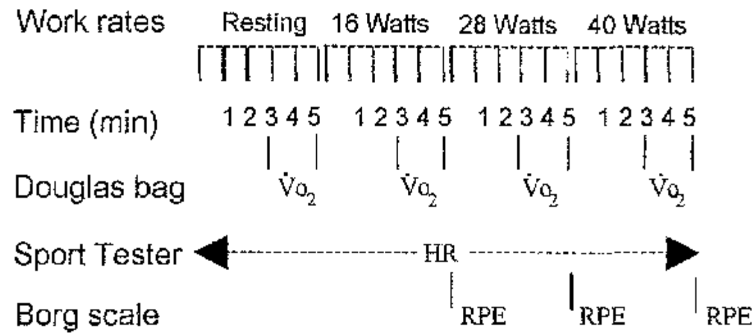

FIGURE 2.3
The procedure of arm crank ergometry experiments

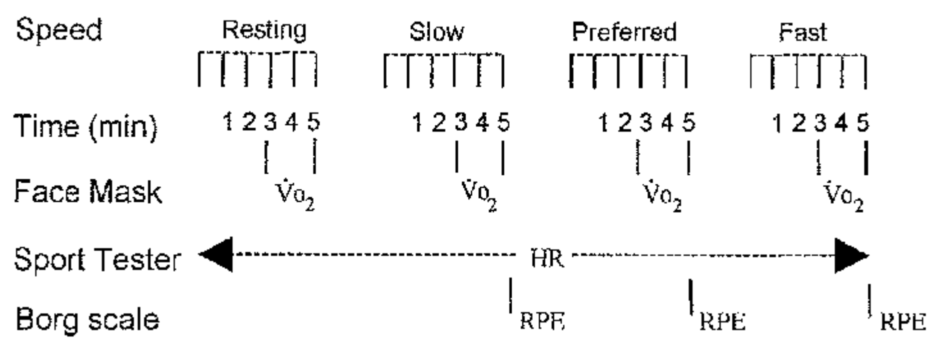

FIGURE 2.4
The procedure of crutch walking experiments

was continuously measured. At the last 2 minutes of each exercise stage, expired air was again collected for analysis. At the end of each work rate, subjects were also shown a standard "6-20" Borg scale and asked to state their perceived exertion (figure 2.3).

6. Agreed conditions for termination of exercise were:

i) The subject requesting that it be stopped,

ii) The subject complaining of any pain or discomfort, and

iii) The subject appearing to be in any distress

2.1.3 CRUTCH WALKING EXPERIMENTS

Ten able-bodied subjects (table 2.2, subjects 1 to 10) and 5 paraplegics (table 2.1, subjects 3, 6, 8 ,9 and 10) participated in this mode of exercise.

2.1.4 PROCEDURE OF CRUTCH WALKING

1. Each able-bodied subject was asked to don an adjusted knee-ankle-foot orthosis with locked knee and ankle; then to rest in the wheelchair. The paraplegic volunteers used their own callipers whenever possible. These were all knee-ankle-foot orthoses. The mass of the callipers was similar in all experiments (3.2±0.6 kg [mean±SE] for paraplegics and 3.5 kg for able-bodied subjects).

2. Standard axillary crutches were used in most instances. Subjects were initially familiarised with the swing-to crutch walking. The crutches were adjusted

to a height approximately 5 cm below the subject's axilla with the hand grip positioned to allow approximately 25° of elbow flexion. Two paraplegics did not use axillary crutches; one of these used a walking frame (no. 3) and the other (no. 9) used tripod crutches.

3. Subjects wore a face mask system for the measurement of oxygen consumption (detail of calibration in section 2.3). The face mask was introduced to the subjects before sampling, for familiarisation purposes. Oxygen consumption was measured during 2 last minutes of each stage of crutch walking. Heart rate was continuously monitored throughout the experiment by PE 3000 Sport Tester with 15 seconds sampling rate. Rating of perceived exertion was evaluated at the end of each experiment (figure 2.4).

4. Each subject walked at his self selected preferred speed on a figure of eight track (Figure 2.5). Able bodied subjects also walked at slower and faster speeds than their preferred speed.

i) Preferred speed: the subject's own most convenient way of walking.

ii) Slow (relaxed) speed: a pattern of walking chosen while strolling in a park or passing leisure time.

iii) Fast (hurried) speed: as when hurried for a train.

5. Before exercise, the resting value of heart rate was continuously monitored for 5 minutes and, during the last two minutes, oxygen consumption

FIGURE 2.5

A paraplegic volunteer (L_2) during crutch walking was measured. Prior to the start of the experiment, the subject stood still until his heart rate stabilised.

6. Each subject moved continously around a 13.5 m figure of eight track. Each exercise period lasted 5 minutes.

7. Time and distance were recorded simultaneously to calculate velocity.

8. Between each of the 3 tests on able-bodied subjects, each person was allowed to sit down for resting, until their heart rate returned to pre-test level, before participating in the other test.

2.1.5 STATISTICAL METHODS

Repeated measure analysis of variance was used to investigate the effect of type of subject (paraplegic or able-bodied), posture (seated or upright arm crank ergometry) and power output (16, 28 and 40 watts) on each measured variable (oxygen consumption, heart rate and perceived exertion) during arm crank ergometry.

Paired *t*-tests were used to investigate the significance of the effects of a change in speed (e.g., 18 to 41 m . min^{-1}) on the energy cost and physiological cost index during crutch walking in able-bodied subjects.

2.2 CALIBRATION OF ERGOMETER

2.2.1 INTRODUCTION

The cycle ergometer is one of the basic instruments in any exercise physiology laboratory. Arm cranking has also been used to assess physiological function in disabled individuals such as paraplegics (e.g., Glaser, 1989). The arm crank ergometer (figure 2.6) was modified from a standard Monark bicycle ergometer and the aim of this subsection is to describe calibration.

2.2.2 DEFINITION

Power is the rate of doing work measured as the amount of work done per second - the watt.

$P = W / t = (F \times D) / t = (F \times (S \times rpm)) / t$

P : power (watts)

W : work (joule)

t : time (second)

F : force (newton)

D : distance (meter)

S : circumference of flywheel (0.52 m)

rpm : revolution per minute of flywheel

rpm of flywheel = 52/14 × rpm of crank

Thus, to generate a certain power output, either the magnitude of the applied force or the rate of application can be adjusted.

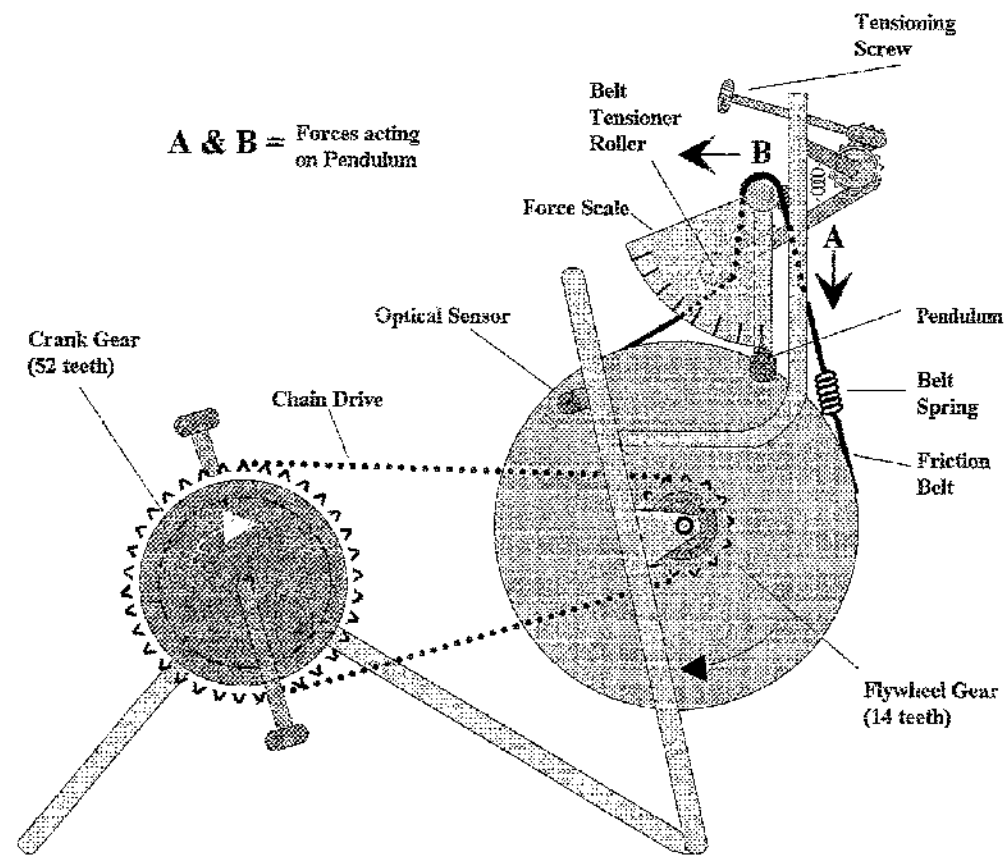

FIGURE 2.6
The arm crank ergometer modified from a Monark cycle ergometer

2.2.3 CALIBRATION PROCEDURE

The calibration of the ergometer's force scale is based on the displacement of a pendulum (Doblen, 1956). Calibration of the force scale requires the application of known forces (i.e., A in figure 2.6) to the pendulum while marking the corresponding new position of the pendulum on the scale. This part of the procedure is carried out in the absence of belt friction forces (i.e., B in figure 2.6).

During exercise, the subject is asked to maintain a constant cranking rate (e.g., 50 rpm). Friction is applied to the pendulum by adjusting the belt tensioning screw and the resultant of forces A and B is displayed on the force scale.

2.2.4 EXPERIMENTS

To prepare the ergometer for the arm cranking experiment, the following precautions were used.

1. The mechanical balance of the flywheel was tested by rotating the pedals for a randomly selected numbers of revolutions. When the flywheel stopped, the "twelve o'clock" position was recorded. This experiment was repeated 130 times.

In the second part of this experiment, the circumference of the flywheel was divided into twelve equal areas. For all twelve marked points of the flywheel, the static friction was measured and each test was repeated 3 times.

The results of both parts showed unequal distribution of the mass of flywheel. Plasticine was applied to the lighter part of the flywheel to obtain balance. The plasticine was then weighed, and its mass (43.33 g) was replaced by a lead weight plus fixing screws. After attaching the lead weight, the balance of the flywheel was checked again, and found free of detectable bias.

2. Measurement of dynamic internal friction of the ergometer was a difficult problem and this was undertaken as a short research project of the Department of Mechanical Engineering (Thoresen, 1993). His procedure can be explained in two parts.

a) An electric motor with a rubber-rimmed pulley was connected to the flywheel and the motor speed adjusted, to obtain a flywheel speed of 186 rpm, measured by a strobe lamp. This flywheel speed was equivalent to a hand crank speed of 50 rpm.

b) The electric motor was removed from the flywheel and connected to a friction brake. The motor was run using the same speed *setting* as in part "a". The brake load was then increased until the actual running speed (checked by the strobe lamb) was exactly as the first part. The friction load therefore represented the dynamic friction of the ergometer.

The internal friction on this ergometer at 50 rpm was found to absorb power at the rate of 4.7±0.2 watts.

3. To increase the sensitivity of the ergometer pendulum (at the low work rates required), the 9.8 N pendulum bar was replaced by one weighing 1.96 N.

4. A small movement of the tension screw produced a large increase in the work rate. Sudden increases in the work rate resulted in a rapid increase in the heart rate. This problem was solved by replacing the friction belt spring with one of a lower elastic modulus. This reduced the sensitivity of the adjusting screw, and

according to the following formula, allowed more movement for each increment of load.

$F = K \times X$

F : force acting from a spring

K : modulus of the spring

X : displacement of the spring

5. Due to a small amount of slack in the tension roller gear unit, adjustment of the tension screw sometimes failed to produce the required friction adjustment. This problem was overcome by attaching a small spring between the roller arm and the gear unit strut.

6. An optical sensor, attached to the ergometer frame, produced a pulse for each complete revolution of the flywheel. The pulses were then applied to the computer digital interface circuit. The computer program counted each pulse received by the interface circuitry and, by calculating the time between consecutive pulses, determined the flywheel speed. Using the crank to flywheel gear ratio 1:3.7, the program then calculated the crank speed in rpm. The computer continually updated and displayed the number of crank revolutions on the screen. While operating the arm crank, the subject could use the visual feedback from the monitor screen to maintain a constant cranking rate.

7. The parallax error when reading the force scale (work rate) of the ergometer was overcome by placing a mirror behind the scale. By aligning a small

hole in the pendulum with its reflection in the mirror, the correct position of the pendulum on the scale was obtained.

8. The freeplay of chain (10-25 mm in the middle) was adjusted according to the Monark catalogue.

9. All moving parts were thoroughly greased. These consisted of the pedal axis, the gears of tensioning screw, the chain, the pendulum bearing, the crank bearing and the tension roller.

10. The flywheel surface was regularly cleaned and a standard Monark belt was used.

2.3 CALIBRATION OF FACE MASK

2.3.1 INTRODUCTION

Oxygen consumption is measured to determine the energy cost of physical activities. A standard method for the measurement of oxygen consumption is to analyse the expired air collected in a Douglas bag.

The Douglas bag equipment has a restraining effect during walking, on especially paraplegics (Winchester et al., 1993), and also limits the movements of any subjects during physical activities (Kawakami et al., 1992). Nene & Patrick (1989) consequently used a face mask apparatus for the analysis of energy cost of crutch walking.

The purpose of the work described in this section was to check and calibrate a portable face mask apparatus for measurement of oxygen consumption.

2.3.2 APPARATUS (FACE MASK)

The apparatus used in this study is shown in figure 2.7 and consisted of the following parts: turbine, rubber face mask, O_2 cell (CTL standard type c/s oxygen sensor), O_2 container as a mixing chamber, electrical component (interface electronics), data logger and computer software.

A turbine for measuring the volume of inspired air is attached to the side of the face mask. The expired air passes through the oxygen cell in the mixing chamber and then to the atmosphere. During inspiration, a photo-detector device measures the velocity of revolutions of the turbine and, from this velocity, the flow volume was calculated. The signals from the turbine and the O_2 cell were amplified

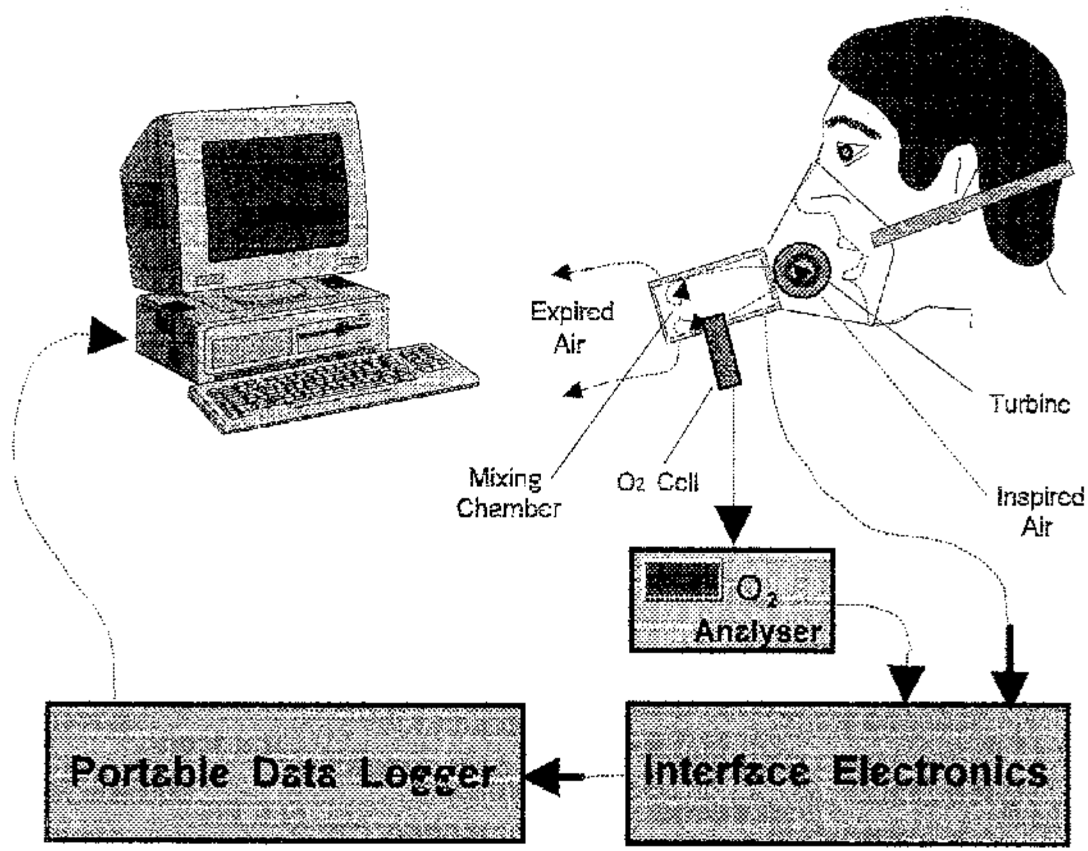

FIGURE 2.7
The face mask used for the measurement of oxygen consumption

and presented to the data logger. At the end of the experiment, the portable data logger was connected to the computer and the data down-loaded for analysis by the software. The oxygen consumption at standard temperature and pressure, dry, was calculated from the volume of inspired air and the percentage concentration of oxygen in expired air.

The total mass of the apparatus was 3.15 kg.

2.3.3 EXPERIMENTS

The procedures were carried out in several separate experiments.

2.3.3.1 CALIBRATION OF O_2 CELL

The aim of this part was to find out the accuracy of the O_2 cell of the face mask. Gas samples were made by mixing 100% O_2 and N_2 in a Douglas bag (the percentage concentration of O_2 was measured by a Servomex 570A O_2 analyser; this was calibrated before and after each experiment, and each sample was randomly made). The head of the face mask oxygen cell container (mixing chamber) was connected to the valve head of the Douglas bag by a tube (10 cm length and the same diameter (4 cm) as the oxygen cell container). The gas was then smoothly pushed through the O_2 cell. By applying different gas samples, the corresponding oxygen concentrations were recorded.

The ranges of O_2 samples were from 10 to 30 percent concentration of O_2. The results showed a highly significant correlation (r=0.993) between percent concentration of O_2 indicated by the standard gas analyser and the voltage output of the O_2 cell of the face mask.

2.3.3.2 TURBINE CALIBRATION

The reason for conducting this experiment was to check the response of the face mask turbine to different flow rates. The gas volume was measured simultaneously by the standard gas meter and the turbine. A Parkinson Cowan gas meter was used and it was calibrated by a Tissot spirometer. In this experiment, room air was passed through the turbine and the gas meter. The temperature of the gas and its humidity were then the same for both measuring systems. The face mask turbine was connected to the gas meter, which was connected to a pump (Figure 2.8). Twenty two different flow rates (10-180 l . min^{-1}) were applied in random sequence. Simultaneously, the relationship between the flow rates and the pressure differences between the inside and the outside of the face mask turbine was measured. For measuring the pressures, water manometers were used.

A significant correlation ($r=0.999$) emerged, between the flow rates measured by the standard gas meter and the face mask turbine (Fig. 2.9). Figure 2.10 shows that there is not a linear relationship between the flow rates and the pressure differences across the turbine face mask. This confirmed the results of a previous study (Dal Monte et al., 1989).

2.3.3.3 COMPARISON OF VOLUME MEASUREMENT METHODS

The aim of this experiment was to compare the Douglas bag and the face mask measurements of the volumes produced by a standard syringe (Figure 2.11). The piston of a six litre syringe was moved at a constant speed within the range of 2 to 10 strokes per minute.

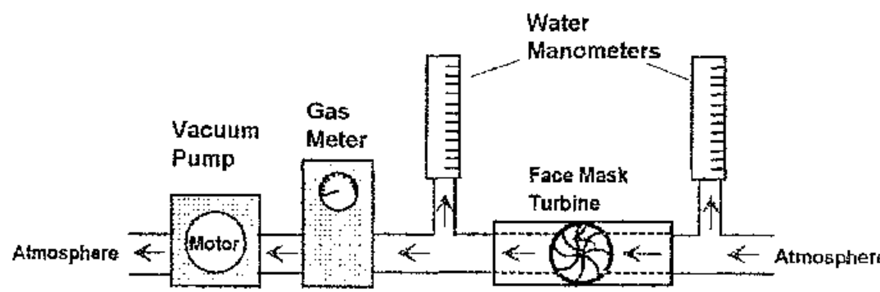

FIGURE 2.8
The pressure drops across the face mask turbine were measured by manometer and simultaneously, the flow rates were measured by both the gas meter and the face mask turbine

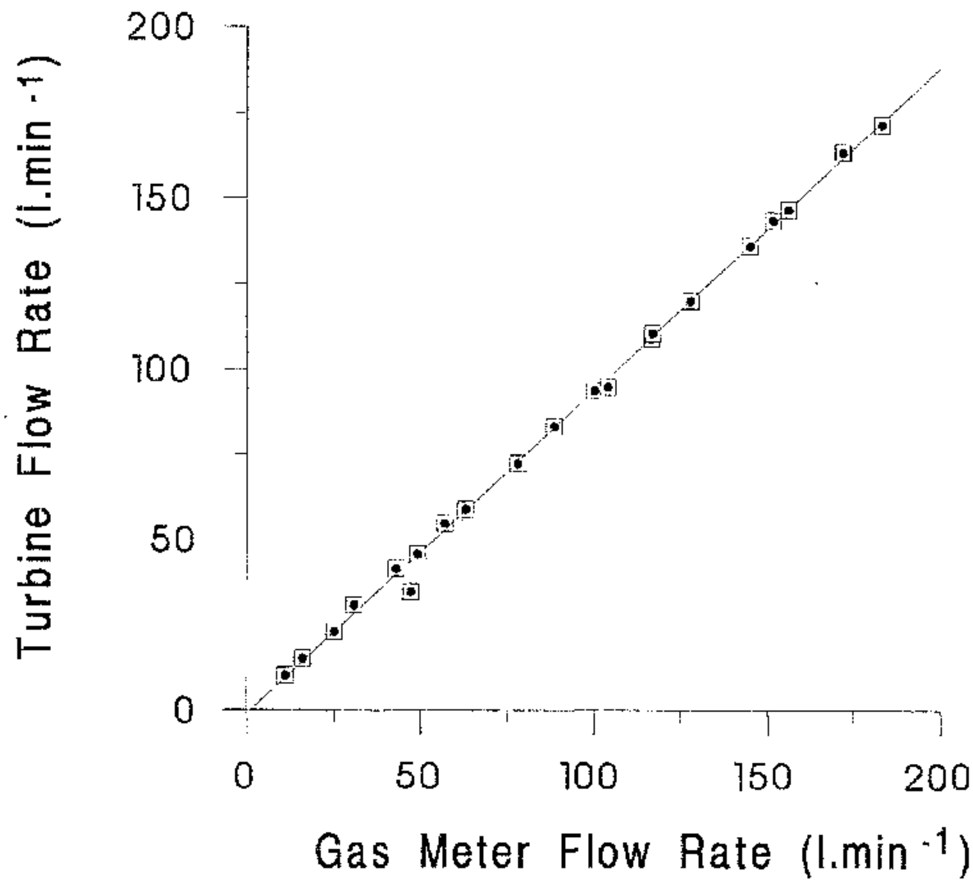

FIG. 2.9

The flow rate measured by the face mask turbine plotted against the flow rate measured by the Parkinson gas meter

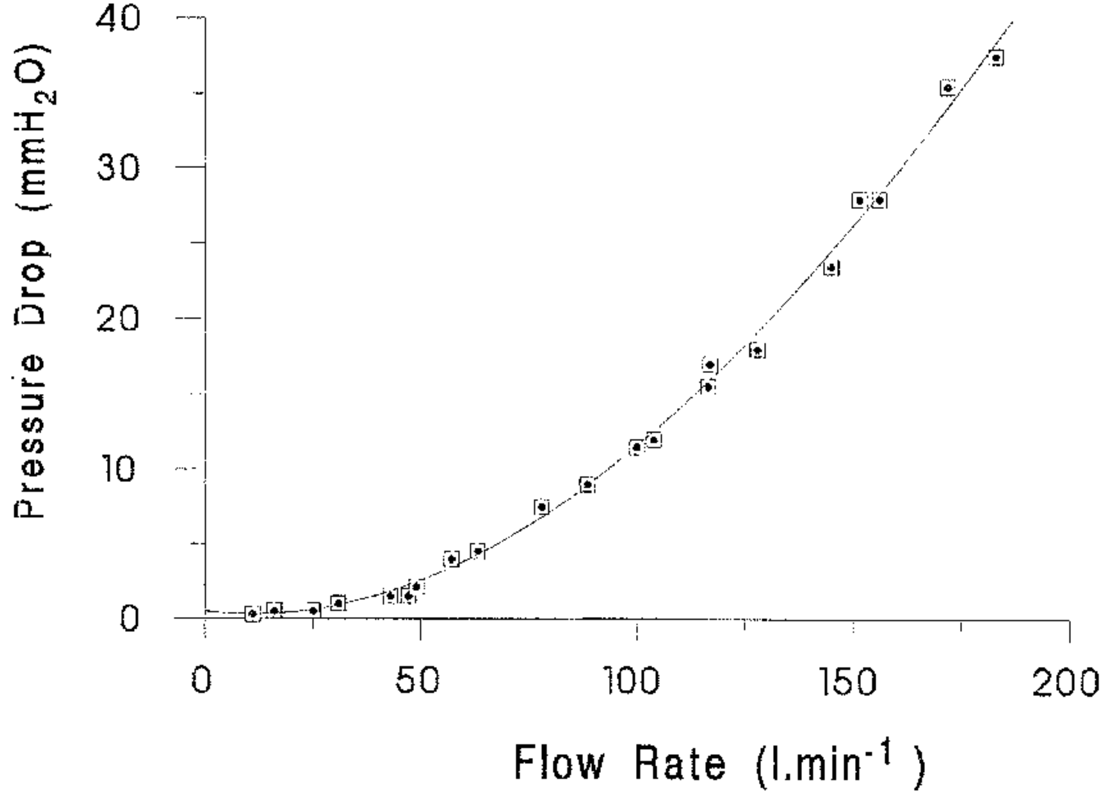

FIG. 2.10

The pressure drops through the face mask turbine were plotted against the flow rates.

Paired *t*-tests did not show any significant differences between standard syringe / face mask; standard syringe / Douglas bag and face mask / Douglas bag reading. The percentage differences (i.e., (($V_{face\ mask}$ or $V_{Douglas\ bag}$ - $V_{syringe}$) / $V_{syringe}$) × 100) between volumes of the Douglas bag and the syringe (10.8%, 0.55%, 1.7% and 1%) and the volumes of the face mask and syringe (2.3%, 1.8%, 1.4% and 2.7%) for all four tested volumes (12, 18, 28 and 30 l . min^{-1}, respectively) were acceptable. There were also highly significant correlations between the volumes produced by the standard syringe and the volumes recorded by both the face mask (r=0.993) and the Douglas bag (r=0.989).

2.3.3.4 COMPARISON OF HUMAN VENTILATION MEASUREMENT METHODS

The aim of this part was to compare directly the measurement of human ventilation during exercise between the face mask turbine and the Douglas bag techniques (fig. 2.12). In this set up, a dual port one-way valve was used. The entrance port of the valve was connected to the face mask turbine. The exit port of the valve was connected to the Douglas bag. The inspired air was then recorded by the face mask turbine and the expired air collected by the Douglas bag.

The subjects performed arm crank ergometry at the known work rates. In the last two minutes of exercise during the steady condition, samples of inspired air (with the face mask) and expired air (with the Douglas bag) were collected. Eight experiments carried out with a rest period, during which heart rate returned to pretest level, between each experiment. The temperature and barometric pressure

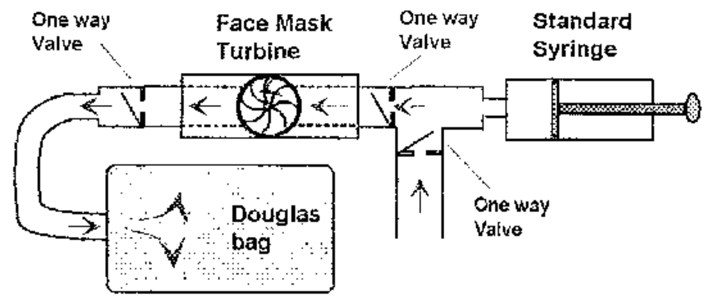

FIGURE 2.11
The calibrated volume of air from the syringe was measured by both the face mask turbine and the Douglas bag

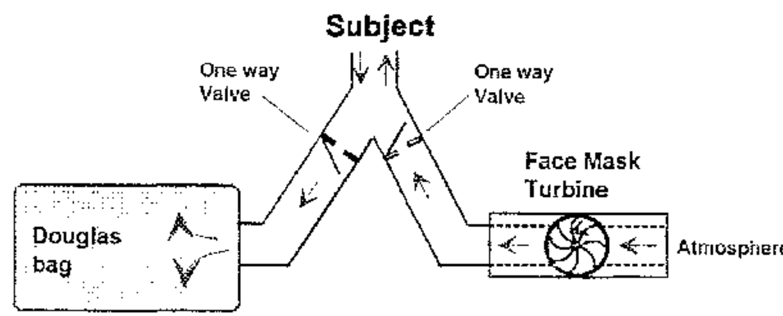

FIGURE 2.12
The volume of inspired air was measured by the face mask turbine and the volume of expired air was measured by the Douglas bag technique during exercise in able-bodied subjects

were recorded and the volumes of both systems were corrected to standard temperature and pressure, dry.

The range of ventilation rates was about 8 to 40 l . min^{-1}. The paired *t*-test showed that there were not significant differences between the flow rates measured by the face mask turbine and those of the Douglas bag. In addition, the mean percentage differences between the flow rates measured by the face mask and the Douglas bag were small (less than 3%). There was also a highly significant correlation between the flow rates measured by the Douglas bag and the face mask ($r=0.997$).

2.3.3.5 COMPARISON OF HUMAN OXYGEN CONSUMPTION MEASUREMENT METHODS

i) The aim of these experiments was to compare the measurement of oxygen consumption obtained with the face mask and the Douglas bag. As in previous experiment (section 2.3.3.4), during steady conditions, expired air was collected directly from the face mask into the Douglas bag (fig. 2.12). During the experiments, the oxygen cell failed to work properly; at such times a portable O_2 analyser (i.e., Instrumentation Laboratory oxygen analyser model 408) was employed instead. This O_2 analyser was calibrated with the known values of O_2 concentration and each gas sample was used several times. The results showed that the O_2 analyser was accurate and reliable. The O_2 cell of the analyser was inserted through a small hole in the side of the face mask mixing chamber.

ii) The aim of these experiments was to compare the oxygen consumption measurement produced by the face mask and a reference system (Douglas bag). Five healthy male subjects volunteered to take part in the study. Each subject performed the test twice. The subjects carried out arm crank ergometry with discontinuous protocol using the 3 work rates (16, 28 and 40 watts). With each subject experiments were carried out, each at 3 work rates. The duration of each work rate was 9 minutes. After the first 3 minutes of arm crank ergometry, used as a warm up and to obtain a steady state, 2 minutes Douglas bag (4th and 5th minutes), 2 minutes face mask (6th and 7th minutes) and again 2 minutes Douglas bag (8th and 9th minutes) were used for measuring oxygen consumption. To compare the oxygen consumption measurements by the face mask and the Douglas bag (i.e., the mean of two samples of Douglas bag), paired *t*-test, correlation coefficient and percentage differences were calculated between the oxygen consumptions measured by two methods .

The oxygen consumption and ventilation of Douglas bag and face mask are shown in table 2.3. Paired *t*-test did not show a significant differences between oxygen consumption measured by Douglas bag and face mask; and the mean of percentage differences between them was less than 5% for all three work rates (4.6%, 3.5% and 3.9% for 16, 28 and 40 watts, respectively). There was a significant correlation (r=0.945) between oxygen consumption measured by two methods (figure 2.13). Correlations between the power output and oxygen consumption measured with both the face mask (r=0.88) and the Douglas bag (r=0.90) were also significant.

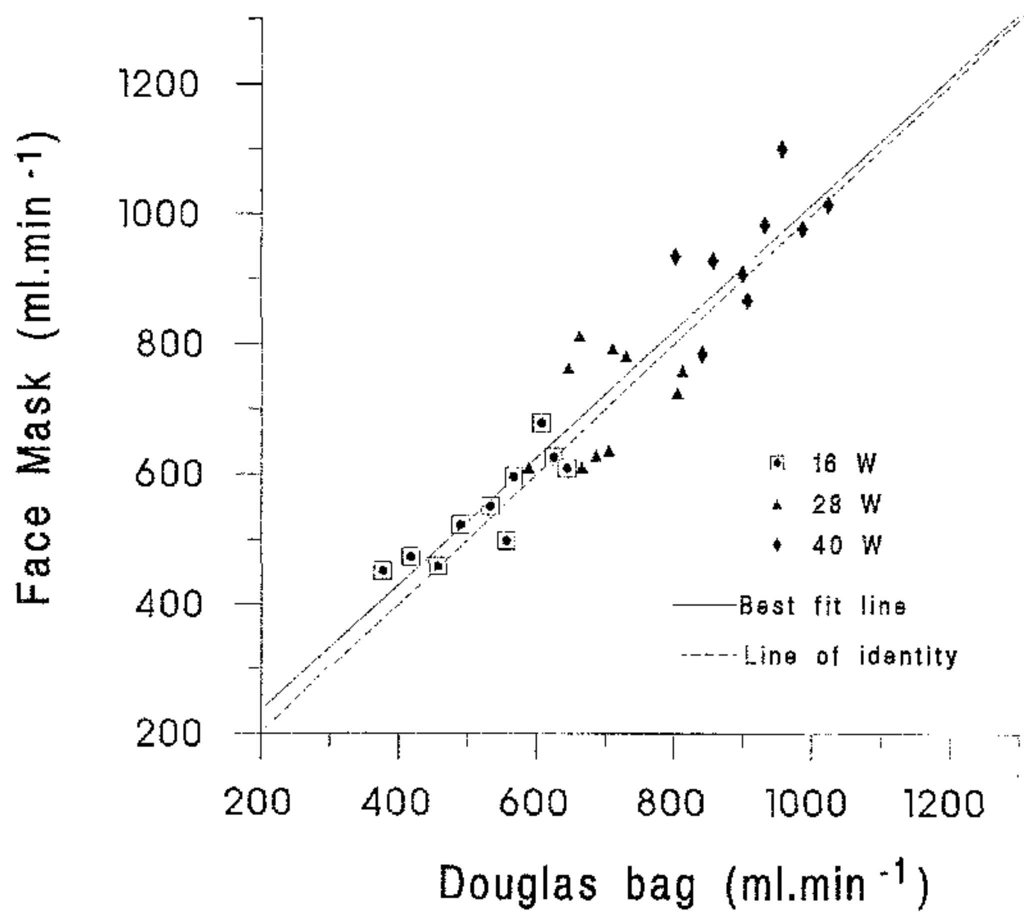

FIG. 2.13

The oxygen consumption measured by the face mask plotted against the oxygen consumption measured by the Douglas bag

Variables	16 W	28 W	40 W
$\dot{V}O_2$ (l . min^{-1}); DB	0.527±0.029	0.701±0.022	0.911±0.024
$\dot{V}O_2$ (l . min^{-1}); FM	0.547±0.025	0.722±0.031	0.946±0.030
$\dot{V}e$ (l . min^{-1}); DB	12.287±0.502	16.147±0.382	21.286±0.624
$\dot{V}e$ (l . min^{-1}); FM	12.287±0.440	16.029±0.661	20.880±0.764

TABLE 2.3

Mean+SE of oxygen consumption and ventilation measured by the face mask (FM) and Douglas bag (DB) methods during arm crank ergometry at three work rates (16, 28 and 40 watts)

2.3.4 SUMMARY

This section has described different methods of checking and calibration of the face mask apparatus. Each part of the face mask was checked. It was then compared with a standard syringe and Douglas bag technique, both separately and together using different gas volumes. Face mask and Douglas bag comparisons were also made during arm crank ergometry at the different work rates.

The results show that the face mask is a suitable instrument for the measurement of oxygen consumption.

CHAPTER 3

RESULTS

The results of this study are divided into four separate sections, i.e.:

i) Assessment of energy expenditure during seated arm crank ergometry

ii) The effect of posture on oxygen consumption, heart rate and perceived exertion

iii) A comparison of assessment of energy expenditure between sports and non-sports persons with spinal cord injury during arm crank ergometry

iv) Assessment of energy expenditure during crutch walking

3.1 ASSESSMENT OF ENERGY EXPENDITURE DURING SEATED ARM CRANK ERGOMETRY

3.3.1 INTRODUCTION

Oxygen consumption, heart rate and perceived exertion responses during seated arm crank ergometry were studied in paraplegic and able-bodied subjects. The means ± SE of these variables during seated arm crank ergometry and the resting period in both 10 sports-active paraplegics and 20 able-bodied subjects are indicated in table 3.1. In both paraplegics and able-bodied subjects, the following are considered:

i) the time course of heart rate responses to seated arm crank ergometry

ii) the reproducibility of measured variables

iii) relationship between measured variables

iv) the comparison of these responses to seated arm crank ergometry between two groups of subjects.

3.1.2 TIME COURSE OF HEART RATE CHANGES

In figure 3.1, mean ± SE of heart rates are plotted for each minute in both paraplegic and able-bodied subjects at rest and all 3 work rates. Heart rate stabilised within about 2 minutes in both groups of subjects. The difference in heart rate between 4th and 5th minutes was less than 2 beats.min^{-1}. This figure shows that an essentially - stable heart rate is achieved during 5 minutes of seated arm crank ergometry in both groups of subjects at all 3 work rates.

			Work rate	
	Rest	16 W	28 W	40 W
Paraplegic subjects				
$\dot{V}O_2$ (l . min^{-1})	0.282±0.010	0.579±0.022	0.745±0.022	0.931±0.024
HR (beats . min^{-1})	72±5	87±4	100±5	114±6
RPE		9.00±0.71	11.80±0.51	13.60±0.48
Able-bodied subjects				
$\dot{V}O_2$ (l . min^{-1})	0.314±0.012	0.567±0.012	0.737±0.012	0.934±0.010
HR (beats . min^{-1})	72±2	82±2	91±3	105±4
RPE		9.41±0.37	11.65±0.40	13.76±0.48

TABLE 3.1

The mean+SE of oxygen consumption, heart rate and rating of perceived exertion at resting condition and all three work rates, for both 10 sports-active paraplegics and 20 able-bodied subjects

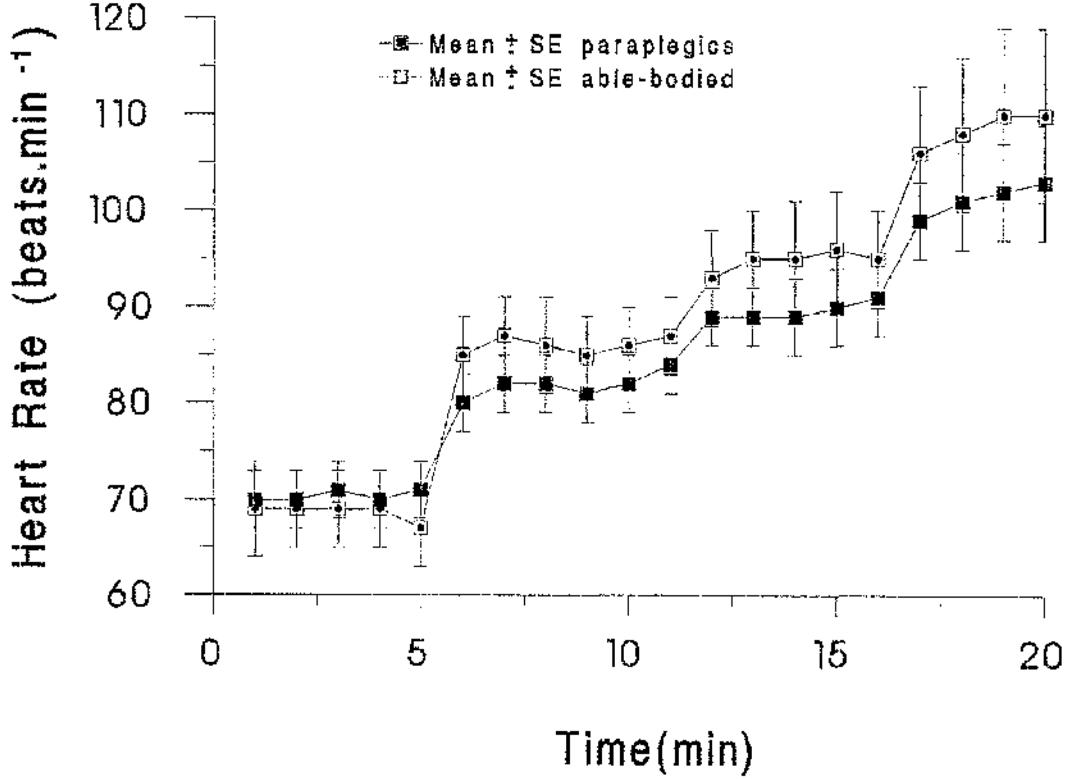

FIG. 3.1

The mean ± SE of time course of heart rate plotted in 10 sports-active paraplegics and 20 able-bodied subjects during seated arm crank ergometry

The fact that heart rate does rise until the second minute of the medium and high work rates is due to the period (approximately 30 seconds) between work-bouts during which the subjects were asked to rate their perceived exertion; because work stopped during these periods, heart rates fell substantially below those at the end of the preceding bout.

3.1.3 REPRODUCIBILITY

For evidence concerning the reproducibility of the measured variables during arm crank ergometry, 6 paraplegics and 6 able-bodied subjects repeated the experiment within 2 weeks (tables 3.2 and 3.3).

The reproducibility of the variable was estimated by a paired *t*-test and correlation coefficient. This was also reported in terms of percentage differences (i.e., ((test-retest) / ((test + retest) / 2)) × 100).

The percentage differences between test and retest trials are indicated in table 3.4 and 3.5. These differences are less than 5% for both groups of subjects for all 3 variables; none are significant (paired *t*-test). There were also strong correlations between test-retest trials of all 3 variables in both paraplegic and able-bodied subjects (table 3.6).

In figure 3.2, retest heart rate is plotted against the test heart rate in 6 paraplegics at 3 work rates. The line of identity and the best-fit line are drawn. This figure visually illustrates the very high reproducibility of heart rate measurements during seated arm crank ergometry in paraplegics. By contrast, as can be deduced from table 3.6, the lowest correlation was found for perceived exertion in these subjects.

	Work rate		
	16 W	28 W	40 W
Test trial			
$\dot{V}O_2$ (l . min^{-1})	0.614±0.021	0.761±0.015	0.963±0.012
HR (beats . min^{-1})	77±3	84±3	91±5
RPE	10.00±0.52	12.33±0.42	13.83±0.65
Re-test trial			
$\dot{V}O_2$ (l . min^{-1})	0.590±0.035	0.757±0.022	0.977±0.021
HR (beats . min^{-1})	75±2	84±3	92±4
RPE	9.83±0.60	11.83±0.54	13.67±0.88

TABLE 3.2

The mean±SE of oxygen consumption, heart rate and rating of perceived exertion during test - retest trials at all three work rates show for 6 able-bodied subjects during seated arm crank ergometry

	Work rate		
	16 W	28 W	40 W
Test trial			
$\dot{V}O_2$ (l . min^{-1})	0.586±0.021	0.752±0.027	0.915±0.037
HR (beats . min^{-1})	88±6	97±7	107±8
RPE	9.50±0.85	12.00±0.45	13.33±0.61
Re-test trial			
$\dot{V}O_2$ (l . min^{-1})	0.577±0.013	0.746±0.023	0.944±0.022
HR (beats . min^{-1})	87±5	97±7	107±9
RPE	9.17±0.79	11.50±0.72	14.17±0.75

TABLE 3.3
The mean±SE of oxygen consumption, heart rate and rating of perceived exertion during test - retest trials at all three work rates, for 6 sports-active paraplegics during seated arm crank ergometry

	Work rate		
	16 W	28 W	40 W
$\dot{V}O_2$ (l . min^{-1})	5.13±3.12	0.49±2.29	-1.55±2.86
HR (beats . min^{-1})	1.74±2.08	-0.28±1.55	-1.05+1.68
RPE	1.67±2.88	4.08±2.71	1.40±3.29

TABLE 3.4
The mean±SE of percentage differences between trials (test and retest) at all three work rates, for the 6 able-bodied subjects

	Work rate		
	16 W	28 W	40 W
$\dot{V}O_2$ (l . min^{-1})	1.25±1.98	0.60±2.72	-3.66±3.51
HR (beats . min^{-1})	0.63±1.23	0.12±0.96	0.18±0.98
RPE	2.10±6.46	3.45±7.46	1.40±3.29

TABLE 3.5
The mean±SE of percentage differences between trials (test and retest) at all three work rates, for the 6 paraplegics

Test - retest trials	**Paraplegic subjects**	**Able-bodied subjects**
	r-value	r-value
$\dot{V}O_2$ (l . min^{-1})	0.903*	0.960*
HR (beats . min^{-1})	0.990*	0.940*
RPE	0.739*	0.919*

* P<0.01

TABLE 3.6
The correlation coefficients between test and retest trials for all three variables indicate for both paraplegic and able-bodied subjects during seated arm crank ergometry

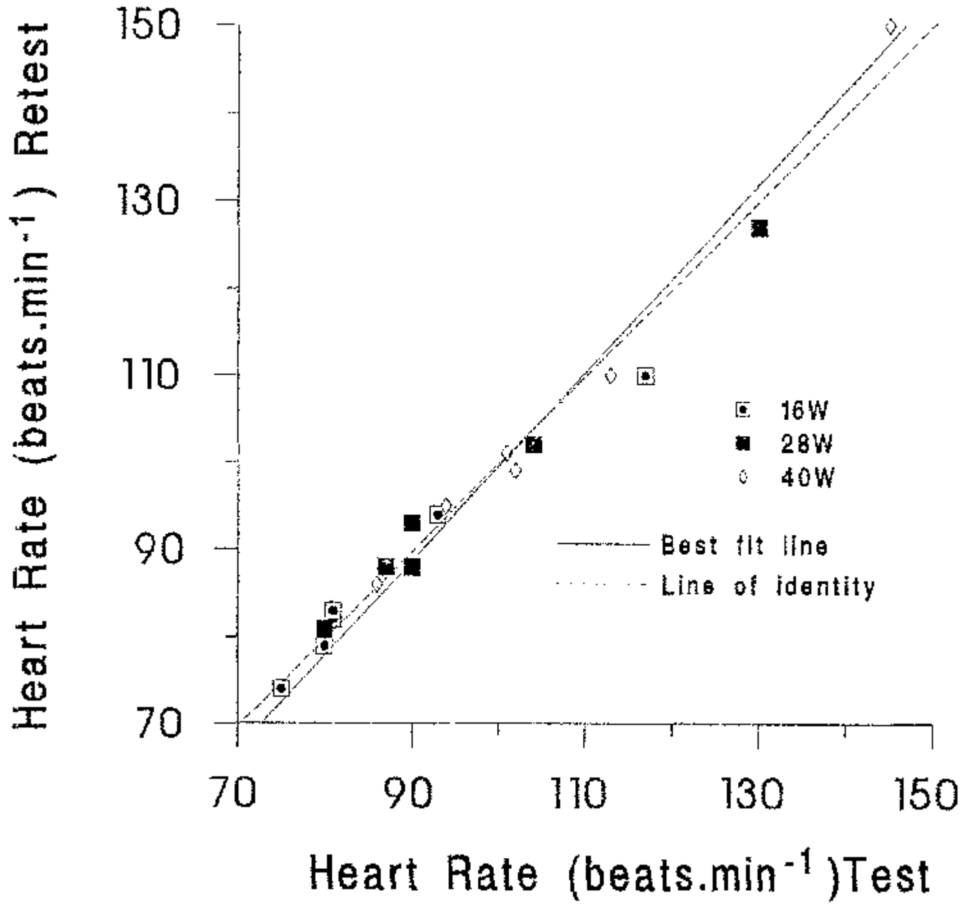

FIG. 3.2

Retest heart rate plotted against the test heart rate in 6 paraplegics during seated arm crank ergometry

3.1.4 RELATIONSHIP BETWEEN VARIABLES

The correlation coefficient between different variables in the full groups of both sports-active paraplegic and able-bodied subjects are indicated in table 3.7. All three measured variables correlated significantly with power output (oxygen consumption most strongly, heart rate least); the variables also correlate one with another. Correlations with power output are illustrated further in figures 3.3 - 3.5.

3.1.5 COMPARISON OF THE MEASURED VARIABLES BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

In figure 3.4, oxygen consumption responses to seated arm crank ergometry are plotted against the power output for both groups of subjects under resting condition and all 3 work rates. Analysis of variance with repeated measurement did not indicate any significant differences of physiological variables between paraplegics and able-bodied subjects at any of the chosen work rates.

	Paraplegic subjects	Able-bodied subjects
	r - value	r - value
Power output / $\dot{V}O_2$	0.864*	0.941*
Power output / HR	0.504*	0.516*
Power output / RPE	0.747*	0.726*
$\dot{V}O_2$ / HR	0.611*	0.466*
$\dot{V}O_2$ / RPE	0.679*	0.668*
HR / RPE	0.532*	0.615*

* $P<0.01$

TABLE 3.7
The correlation coefficients between measured variables indicate for both sports-active paraplegic (n=10) and able-bodied subjects (n=20) during seated arm crank ergometry

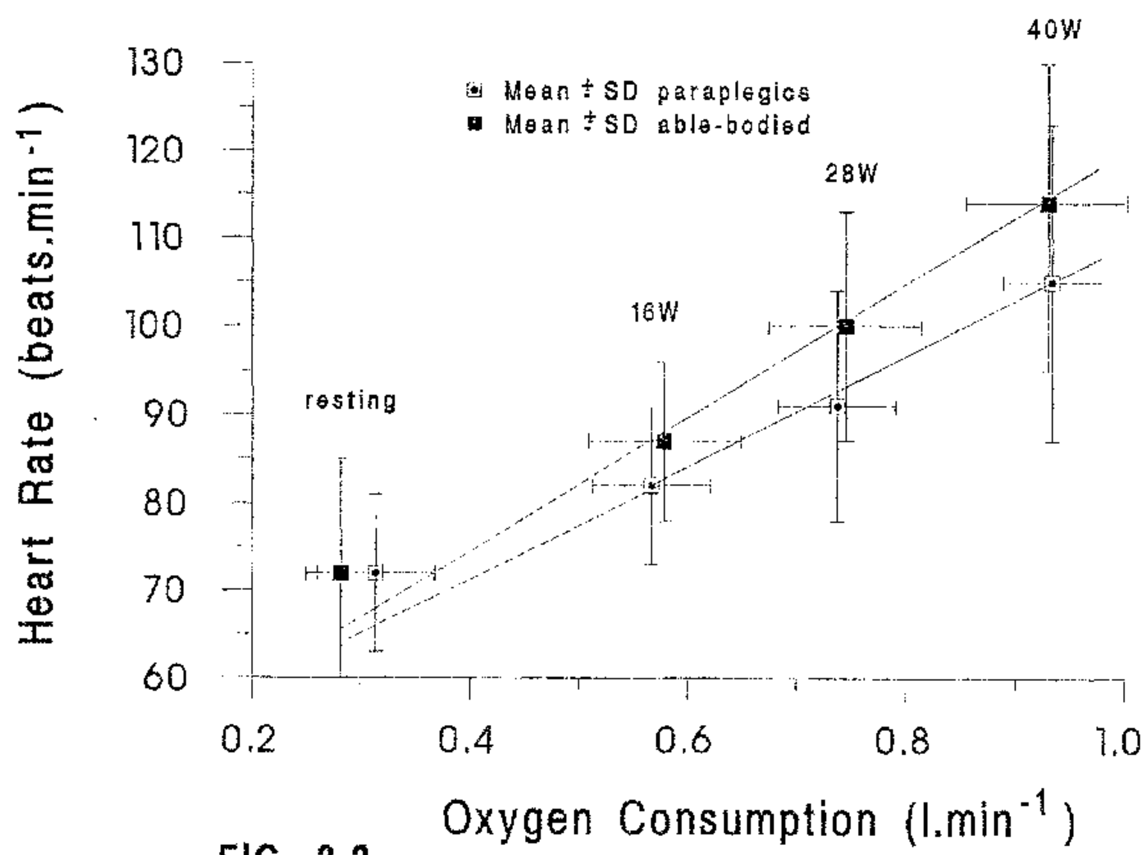

FIG. 3.3

The mean ± SD of heart rate plotted against oxygen consumption during seated arm crank ergometry at resting condition and three work rates (16, 28 and 40 watts) in 10 sports-active paraplegics and 20 able-bodied subjects

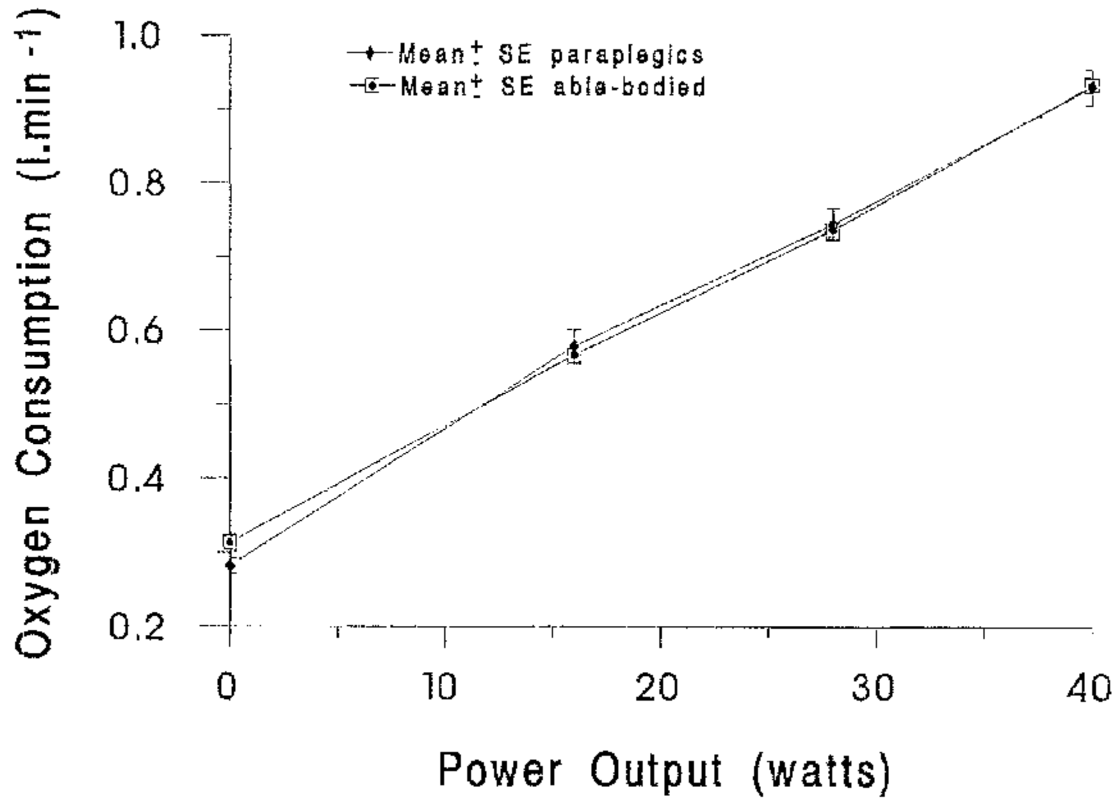

FIG. 3.4

The mean ± SE of oxygen consumption plotted against the power output during seated arm crank ergometry in 10 sports-active paraplegics and 20 able-bodied subjects

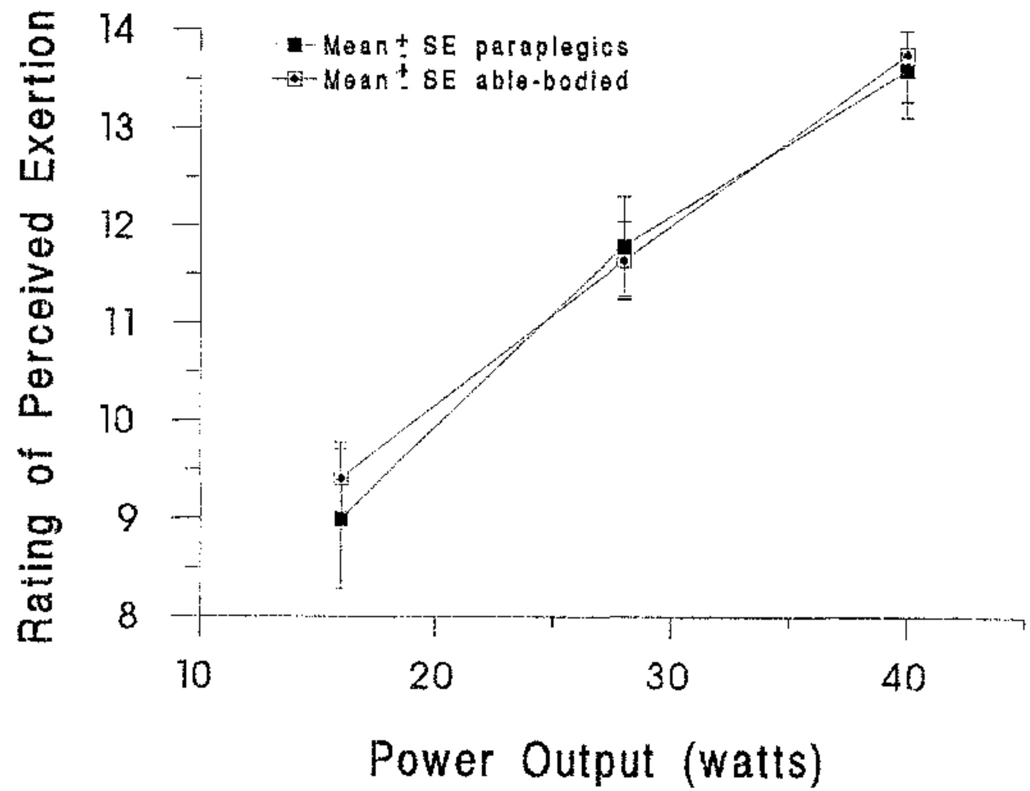

FIG. 3.5
The mean ± SE of rating of perceived exertion plotted against the power output during seated arm crank ergometry in 10 sports-active paraplegics and 20 able-bodied subjects

3.2 THE EFFECT OF POSTURE ON THE OXYGEN CONSUMPTION, HEART RATE AND PERCEIVED EXERTION

3.2.1 INTRODUCTION

In this part, the physiological responses to seated and upright arm crank ergometry are compared between 7 sports-active paraplegics and 20 able-bodied subjects. The mean ± SE of oxygen consumption, heart rate and perceived exertion in both 7 paraplegics and 20 able-bodied subjects during seated and upright arm crank ergometry are indicated in table 3.8.

3.2.2 TIME COURSE OF HEART RATE CHANGES

The mean ± SE of heart rate responses to seated arm crank ergometry in the 7 paraplegics and 20 able-bodied subjects are indicate in figure 3.6. This figure shows that heart rate becomes effectively stable at all 3 work rates in both groups of subjects (neither the apparently continuing upward trend with time nor the difference between paraplegic and able-bodied subjects is statistically significant).

However, during upright arm crank ergometry, the heart rate responses were different between the 2 groups of subjects. In figure 3.7, the heart rate for each of the individual paraplegics during upright arm crank ergometry is compared with the mean ± SE for the 20 able-bodied subjects. A trend is displayed for high-lesion paraplegics (T_5 and one of the two T_8s). Furthermore, in one subject (a T_8 lesion) heart rate did not stabilise during the 5 minutes of exercise at the highest

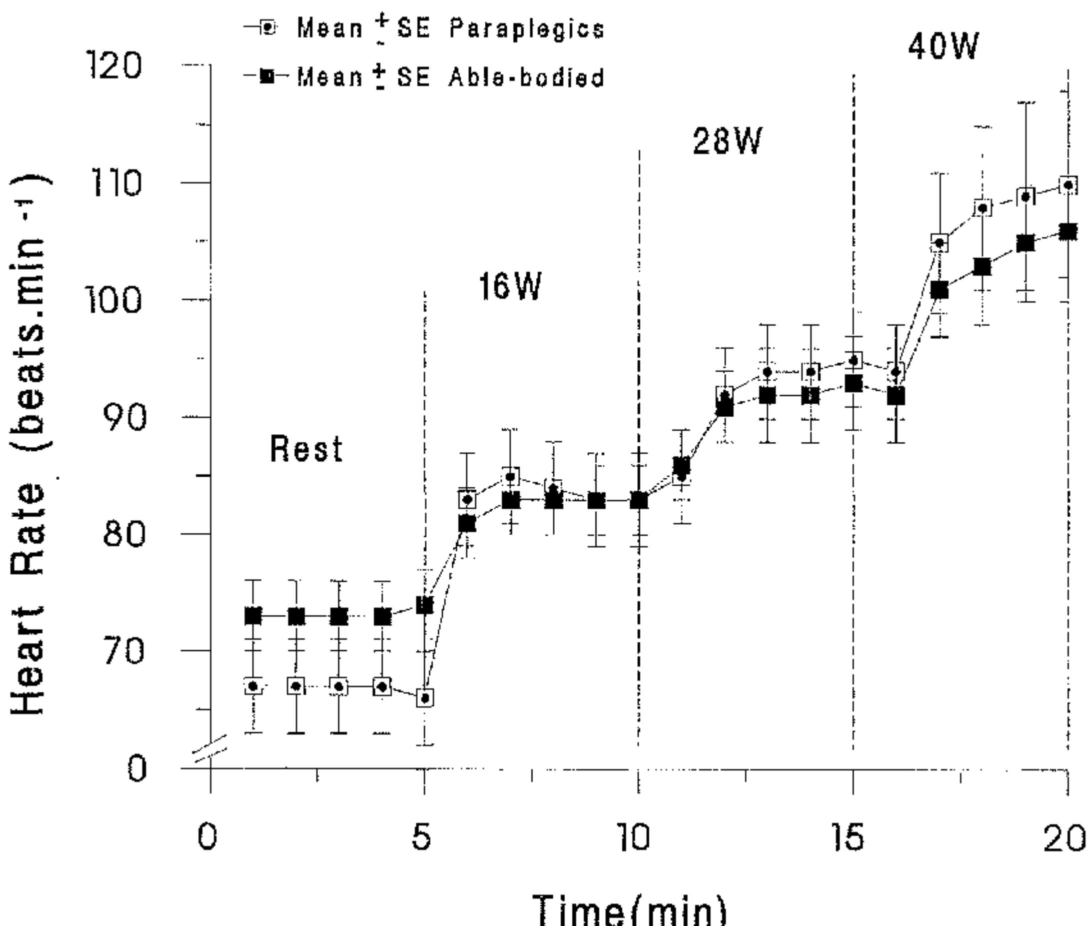

FIG. 3.6

Time course of heart rate plotted in 7 sports-active paraplegics and 20 able-bodied subjects during seated arm crank ergometry

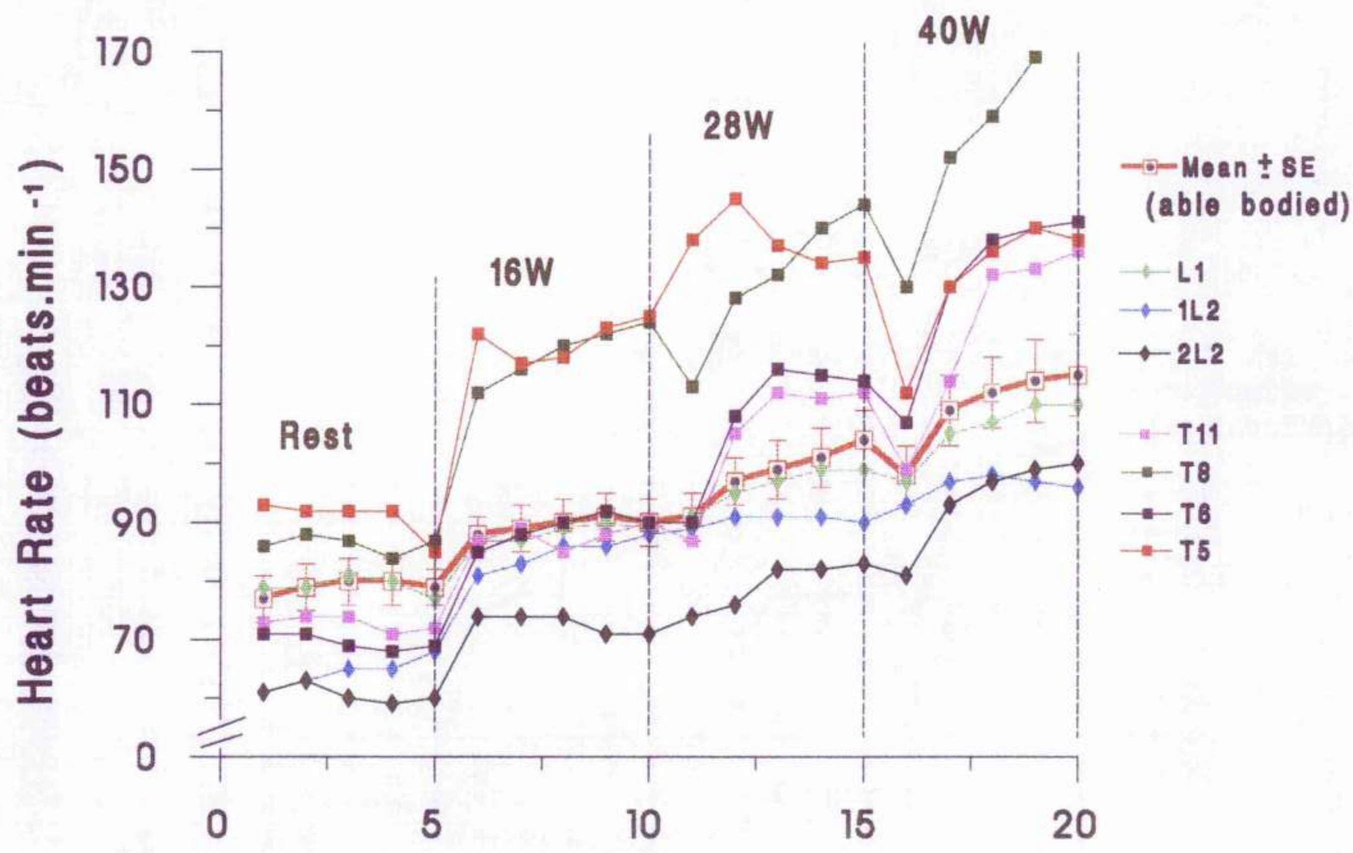

FIG. 3.7

Time course of heart rate plotted in sports-active paraplegics and able-bodied subjects during upright arm crank ergometry

Seated ACE			**Work rate**		**Upright ACE**		**Work rate**	
	Rest	16 W	28 W	40 W	Rest	16 W	28 W	40 W
Paraplegic subjects								
$\dot{V}O_2$ (l . min^{-1})	0.282±0.009	0.584±0.009	0.743±0.009	0.941±0.009	0.280±0.010	0.593±0.031	0.746±0.027	0.920±0.026
HR (beats . min^{-1})	67±4	83±4	95±5	110±8	76±4	97±4	112±8	126±10
RPE		8.43±0.78	11.14±0.51	13.00±0.49		9.43±0.61	12.00±0.44	13.86±0.55
Able-bodied subjects								
$\dot{V}O_2$ (l . min^{-1})	0.314±0.012	0.567±0.012	0.737±0.012	0.934±0.010	0.314±0.012	0.602±0.019	0.790±0.018	0.989±0.014
HR (beats . min^{-1})	72±2	82±2	91±3	105±4	77±3	89±3	100±4	113±5
RPE		9.41±0.37	11.65±0.40	13.76±0.48		9.35±0.417	11.71±0.28	13.82±0.38

TABLE 3.8

The mean±SE of oxygen consumption, heart rate and rating of perceived exertion at resting condition and all three work rates show for both 7 sports-active paraplegics and 20 able-bodied subjects during seated and upright arm crank ergometry work rate (40 watts): it rose by 20 beats during the last 2 minutes. In the T_5 paraplegic, however, heart rate appeared to have reached its limit at 28 watts. By contrast, in seated arm crank ergometry at the same work rate, a stable heart rate was achieved within 2 minutes by all paraplegics at all 3 work rates (fig. 3.6).

3.2.3 COMPARISON OF MEASURED VARIABLES BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

The design used to investigate the effect of the 3 important factors (i.e., posture, power output and type of subjects), is graphically illustrated below.

Subjects	P						AB					
Posture of ACE	S			U			S			U		
Power output (Watts)	16	28	40	16	28	40	16	28	40	16	28	40

P : Paraplegic
AB : Able-bodied
S : Seated arm crank ergometry
U : Upright arm crank ergometry

For statistical analysis, because of the incremental protocol and the fact that all combinations of posture and power output are measured on the same subject, a repeated measure analysis of variance was used. Thereby, the effect of posture (seated and upright arm crank ergometry), power output (16, 28 and 40 watts) and type of subject (paraplegic and able-bodied) on the measured variables (oxygen consumption, heart rate and perceived exertion) were studied.

The effect of the power output on all 3 variables was significant; however, the type of subject, the posture and the interaction between these did not significantly affect oxygen consumption.

Heart rate however, was significantly higher in upright than seated arm crank ergometry in both types of subject and there was also an interaction between the type of subject (paraplegic and able-bodied) and the effect of posture on the heart rate.

Figure 3.10 displays the equivalent data for subjects' perceived exertion. Clearly, there was no postural effect in normal subjects. There is a suggestion that paraplegics found work at any given rate harder upright than seated, attain the accepted level of statistical significance ($P<0.07$).

In figures 3.8, 3.9 and 3.10, oxygen consumption, heart rate and perceived exertion are respectively plotted against the power output.

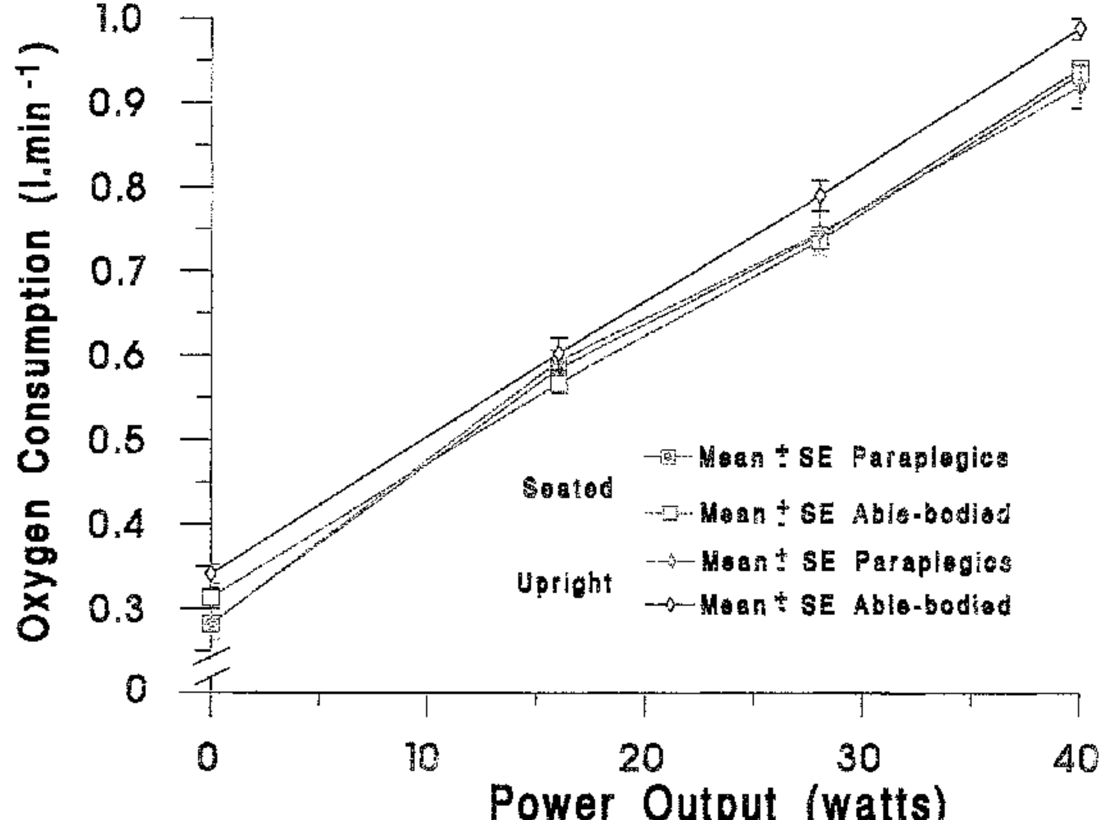

FIG. 3.8

Mean ± SE of oxygen consumption plotted against the power output during seated and upright arm crank ergometry in 7 sports-active paraplegics and 20 able-bodied subjects

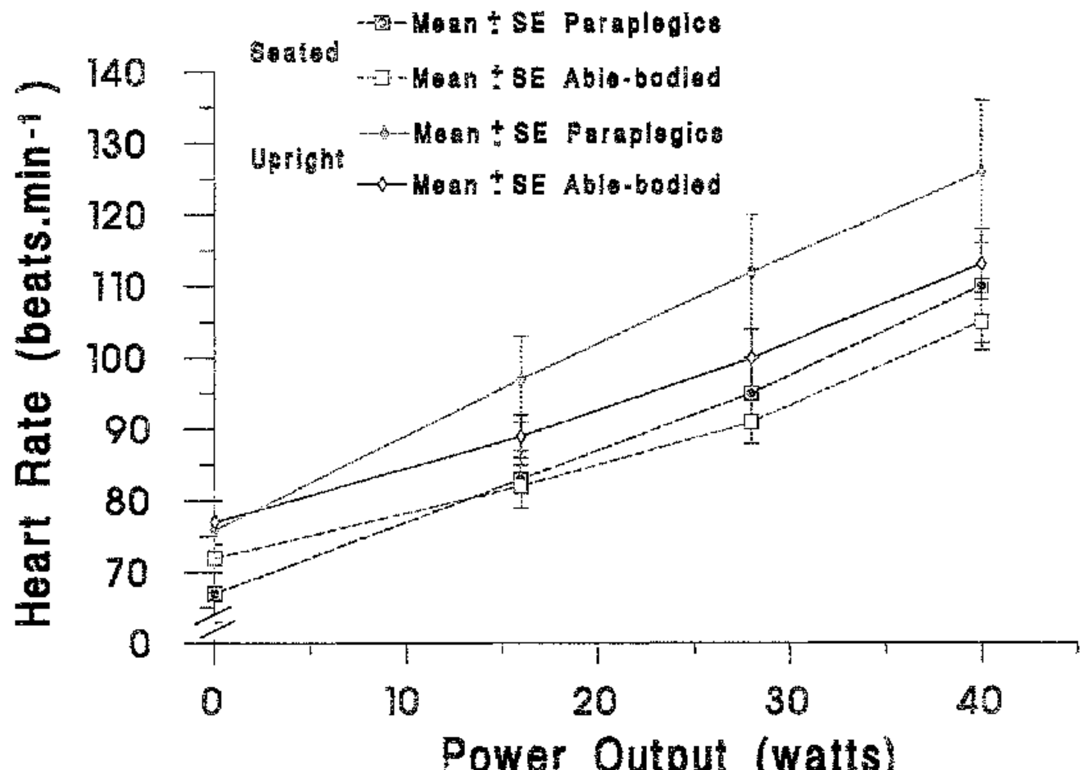

FIG. 3.9

Mean ± SE of heart rate plotted against the power output during seated and upright arm crank ergometry in 7 sports-active paraplegics and 20 able-bodied subjects

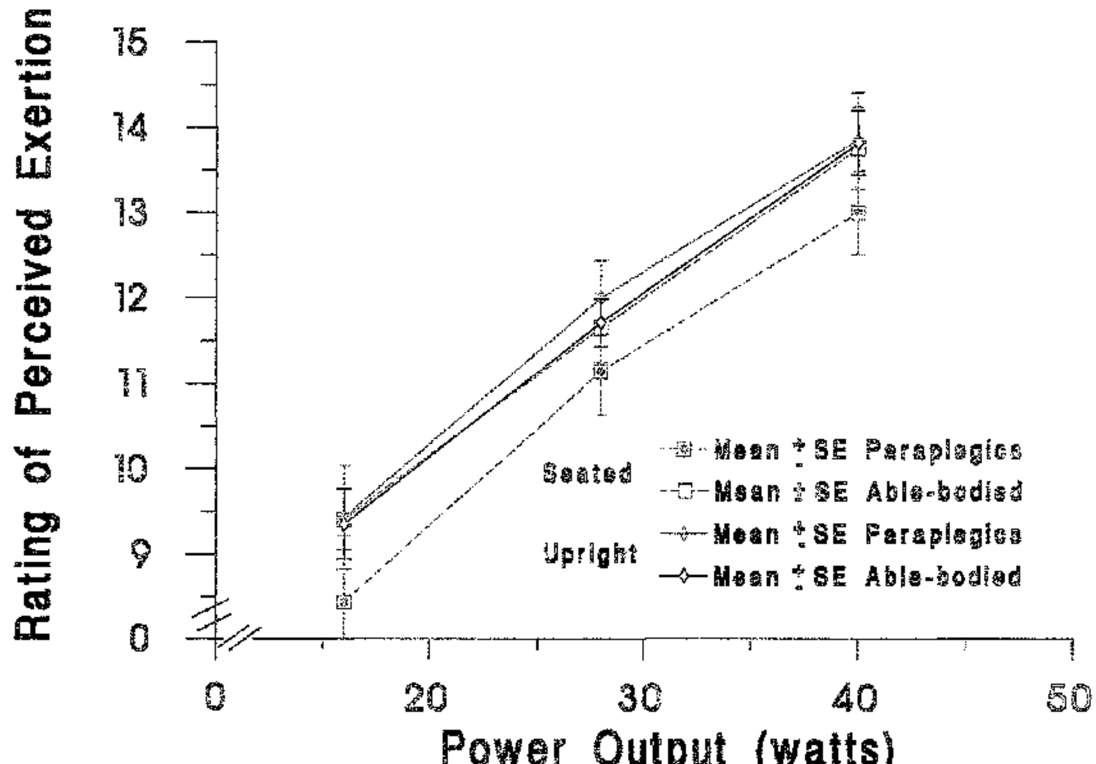

FIG. 3.10

Mean ± SE of rating of perceived exertion plotted against the power output during seated and upright arm crank ergometry in 7 sports-active paraplegics and 20 able-bodied subjects

3.3 A COMPARISON OF ASSESSMENT OF ENERGY EXPENDITURE BETWEEN SPORTS-ACTIVE AND NON-SPORTS PERSONS WITH SPINAL CORD INJURY DURING ARM CRANK ERGOMETRY

In the following section, the results of 10 sports-active paraplegics who participated for seated and upright arm crank ergometry are compared with the data of the 2 non-sports paraplegics who were available and volunteered to perform the arm crank ergometry.

The seated arm crank ergometry were carried out by all 12 paraplegics. However, only 1 of the non-sports and 7 sports-active paraplegics were able to perform upright arm crank ergometry. The results from the non-sports paraplegic shows the unstable heart rate (fig. 3.11) at the last work rate of 40 watts during upright arm crank ergometry. His heart rate rose each minute by 3 beats from second to fifth minutes. This is similar to the result of a complete T_8 sports-active paraplegic shown earlier in fig. 3.7.

The small number of non-sports paraplegics participating in this experiment made a statistical analysis impractical.

Although seated oxygen consumption and heart rates during exercise at 16 watts show no apparent differences between the two groups (table 3.9), heart rates at 28 and 40 watts and perceived exertion at all work rates show that the non-sporting paraplegics felt the work easier than did the sports active paraplegics. Heart rate in the upright non-sports paraplegic was within the range of the sports-active paraplegics, but the former used less oxygen.

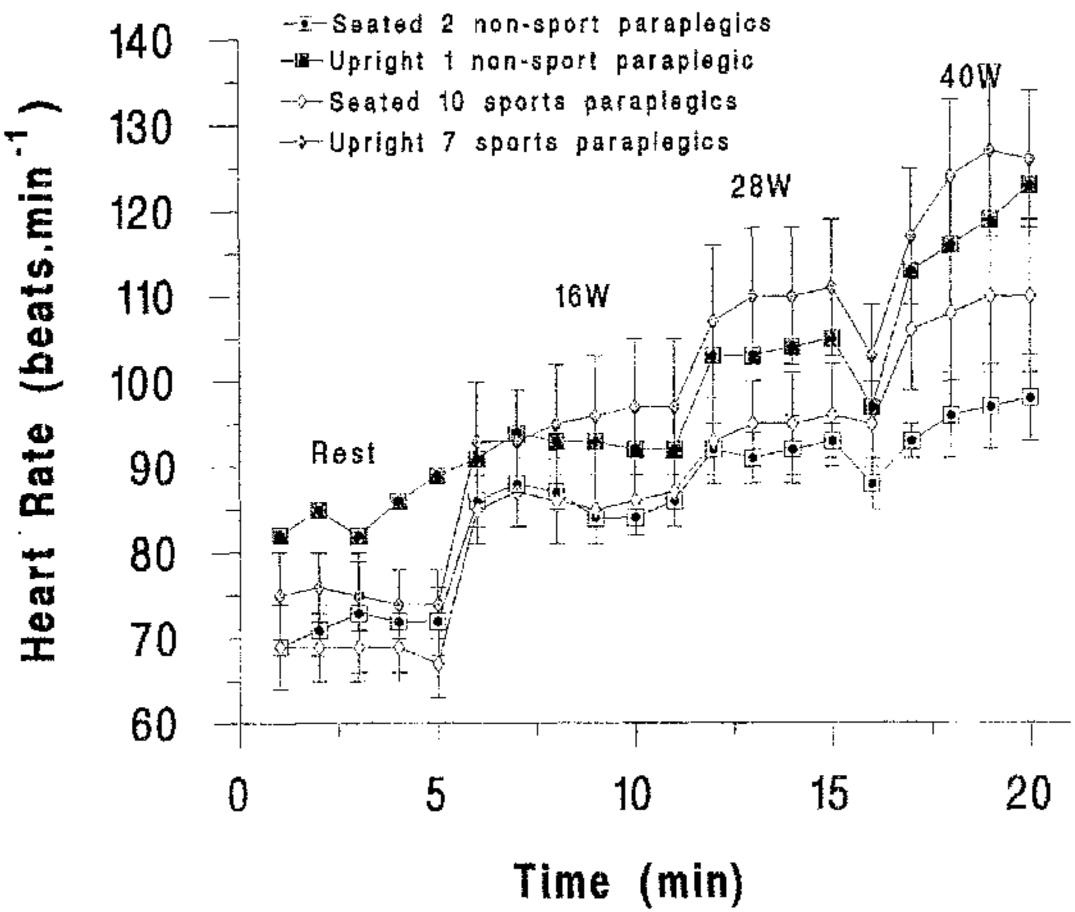

FIG. 3.11

The mean+SE of time course of heart rate plotted in sports-active and non-sports paraplegics during seated and upright arm crank ergometry

Type of paraplegic	Arm Ergometry	OXYGEN CONSUMPTION ($l.min^{-1}$) Rest	16W	28W	40W	HEART RATE ($beats.min^{-1}$) Rest	16W	28W	40W	PERCEIVED EXERTION 16W	28W	40W
Non-sport 2 paraplegics	Seated	0.235± 0	0.583± 0.001	0.781± 0.009	0.904± 0.002	71± 4	84± 2	93± 2	98± 5	7± 0	7.5± 0.5	10.5± 0.5
Non-sport 1 paraplegic	Upright	0.272	0.521	0.712	0.885	85	92	105	123	8	11	11
Sport-active 10 paraplegics	Seated	0.282± 0.010	0.579± 0.022	0.745± 0.022	0.931± 0.024	72± 5	87± 4	100± 5	114± 6	9± 0.71	11.80± 0.51	13.60 ±0.48
Sport-active 7 paraplegics	Upright	0.280± 0.010	0.593± 0.031	0.746± 0.027	0.920± 0.026	76± 4	97± 6	112± 8	126± 10	9.43± 0.81	12± 0.44	13.86 ±0.55

TABLE 3.9

A comparison of variables (mean±SE) between 10 sports-active and 2 non-sporting paraplegics during seated and upright arm crank ergometry

At the same power output, the percentage difference of heart rate was 6.5% lower in non-sports paraplegics than in the sports-active paraplegics. This percentage difference is the mean of all exercise periods at all intensities in both seated and upright experiments. A similar analysis of perceived exertion shows the non-sports active paraplegics rated the work 20% easier to perform than did the sports-active paraplegics.

Clearly, these trends are counter to one's expectation, on the basis of training. It should be considered that the physiological responses to exercise in paraplegics are related to several parameters e.g., nature, level and completeness of injury plus regularity and intensity of exercise. Because both non-sports paraplegics had incomplete lesions, therefore, it could be suggested that completeness of injury in paraplegics has a more important effect on their physiological responses to exercise than the regularity of their sports participation. However, the tiny sample size obviously permits no firm conclusion.

3.4 ASSESSMENT OF ENERGY EXPENDITURE DURING CRUTCH WALKING

3.4.1 INTRODUCTION

As described under "Method" (section 2.1.4), crutch walking with knee-ankle-foot orthoses was carried out by 5 sports-active paraplegics and 10 able-bodied subjects on a figure of eight track. All subjects walked at their individually preferred speed; able-bodied subjects also walked at slower and faster speeds. Two methods were used to express energy efficiency of walking. These methods were based respectively on the oxygen consumption and heart rate.

The following sections considers:

i) time courses of heart rate responses to crutch walking in both groups

ii) physiological responses to crutch walking at 3 speeds in able-bodied subjects

iii) comparison of energy cost of crutch walking between paraplegic and able-bodied subjects.

3.4.2 TIME COURSE OF HEART RATE CHANGES

Figure 3.12 illustrates the mean ± SE of heart rate obtained from the 10 able-bodied subjects for each of 5 consecutive minutes at rest and the 5 minutes at each of the 3 speeds of crutch walking. In these subjects, stable heart rates were reached within about 2 minutes at the 2 lower speeds: slow (18 ± 1.6 m . min^{-1}) and preferred (28 ± 1.5 m . min^{-1}). However, at fast speed (41.1 ± 2.2 m . min^{-1}) heart

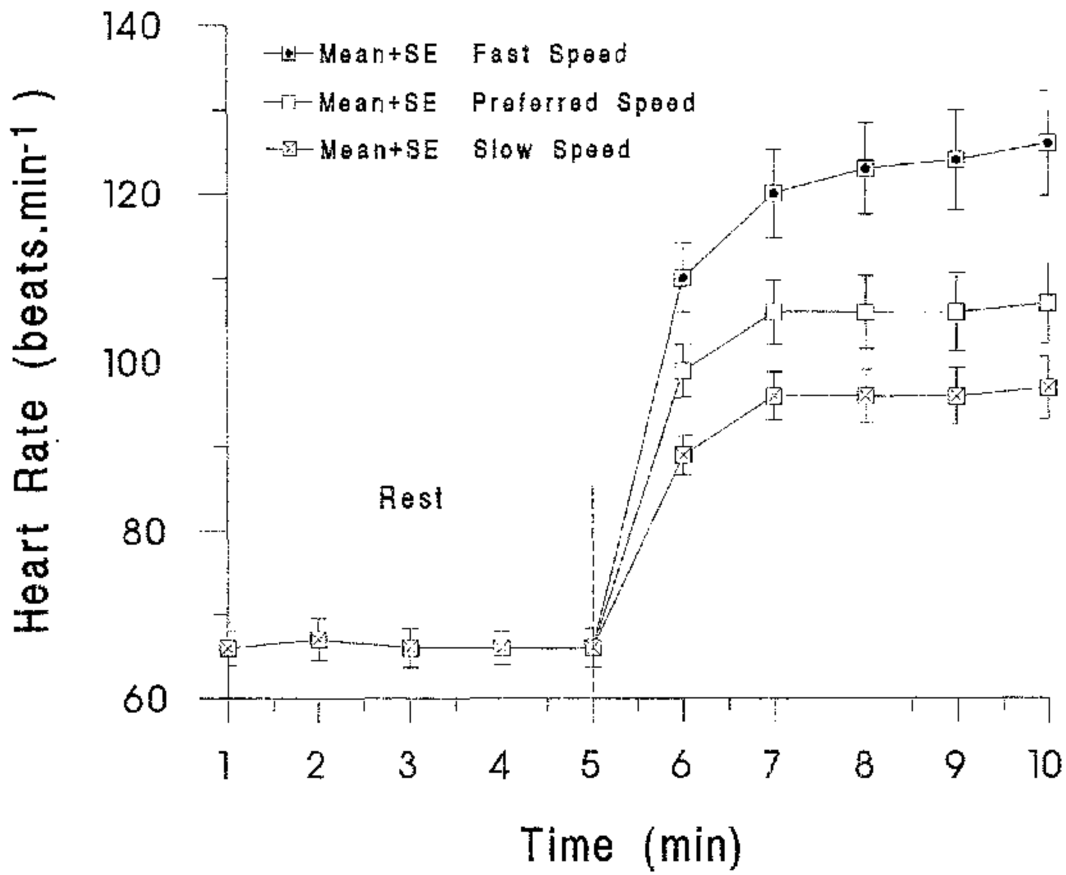

FIG. 3.12

Time course of heart rate during crutch walking at three speeds in able-bodied subjects rates showed a continuing upward trend, rising progressively one beat each minute from the 2nd to the 5th minute.

Figure 3.13 shows the mean ± SE of heart rate during crutch walking at preferred speed in paraplegics (with the data for able-bodied subjects repeated for comparison). Heart rates in the 3 lumbar paraplegics became stable during 5 minutes exercise, as in able-bodied subjects. It should be recognized that the actual values of the heart rates cannot be directly compared between groups, because the speeds of crutch walking are not the same.

The two thoracic paraplegics responded differently from those with lower lesions. In the T_6 volunteer, whose speed was about 1/3 that of lumbar paraplegics and able-bodieed subjects (table 3.12), heart rate rose progressively each minute until the end of 5 minutes of exercise and the difference in heart rate between 4th and 5th minutes was 6 beats . min^{-1}. In the T_{11} subject, heart rate increased rapidly up to about 180 beats . min^{-1} and it is to be noted that he was unable to complete the 5th minute of exercise, despite the fact that his speed was less than 1/5 that adopted preferentially by the lumbar paraplegics and able-bodied subjects.

3.4.3 COMPARISON OF MEASURED VARIABLES IN ABLE-BODIED SUBJECTS

Table 3.10 and figures 3.14 - 3.16 show the mean ± SE of oxygen consumption, heart rate and perceived exertion of the 10 able-bodied subjects under resting conditions and during crutch walking at 3 speeds with knee-ankle-foot orthoses. Table 3.11 shows the correlations between all 3 variables and speed; all are significant.

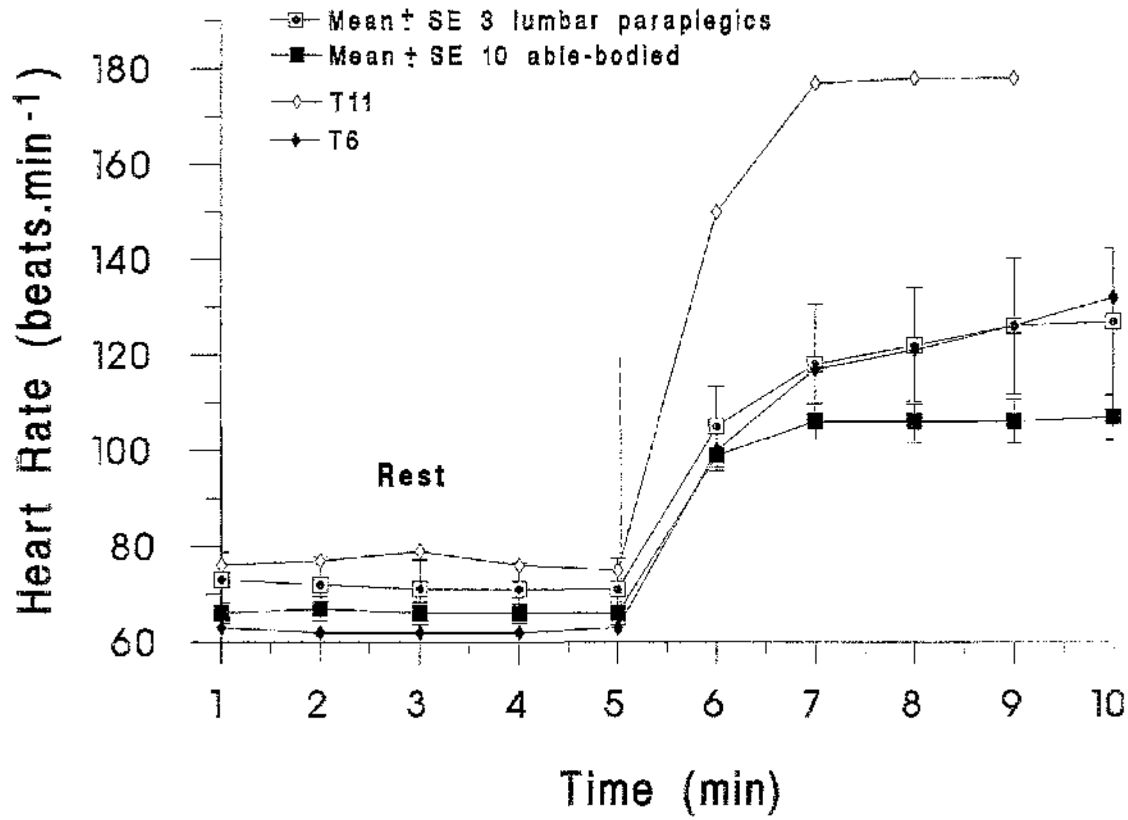

FIG. 3.13

Time course of heart rate during crutch walking at preferred speed in paraplegic and able-bodied subjects

Speed	HR (beats.min^{-1})	$\dot{V}O_2$ (ml.kg^{-1}.min^{-1})	Energy Cost (ml.kg^{-1}.m^{-1})	PCI (beats.m^{-1})	Speed (m.min^{-1})	RPE
Rest	67±2.1	3.31±0.25				
Slow	97±3.7	11.91±1.06	0.67±0.04	1.74±0.17	18.04±1.64	11±0.8
Preferred	107±5.0	14.58±0.84	0.52±0.03	1.46±0.11	28.05±1.46	13±0.6
Fast	126±6.7	22.45±1.37	0.57±0.05	1.47±0.14	41.07±2.17	16±0.7

TABLE 3.10

Mean±SE of heart rate, oxygen consumption, energy cost, physiological cost index, speed and rating of perceived exertion during crutch walking in able-bodied subjects

Variables	r-value (correlation coefficient)
Oxygen consumption / Speed	0.736*
Heart Rates / Speed	0.684*
Rating of Perceived Exertion / Speed	0.733*
Oxygen consumption / Heart Rate	0.633*
Energy Cost / Physiological Cost Index	0.649*

* $P<0.01$

TABLE 3.11
The correlation coefficients between variables during crutch walking at three speeds in 10 able bodied subjects

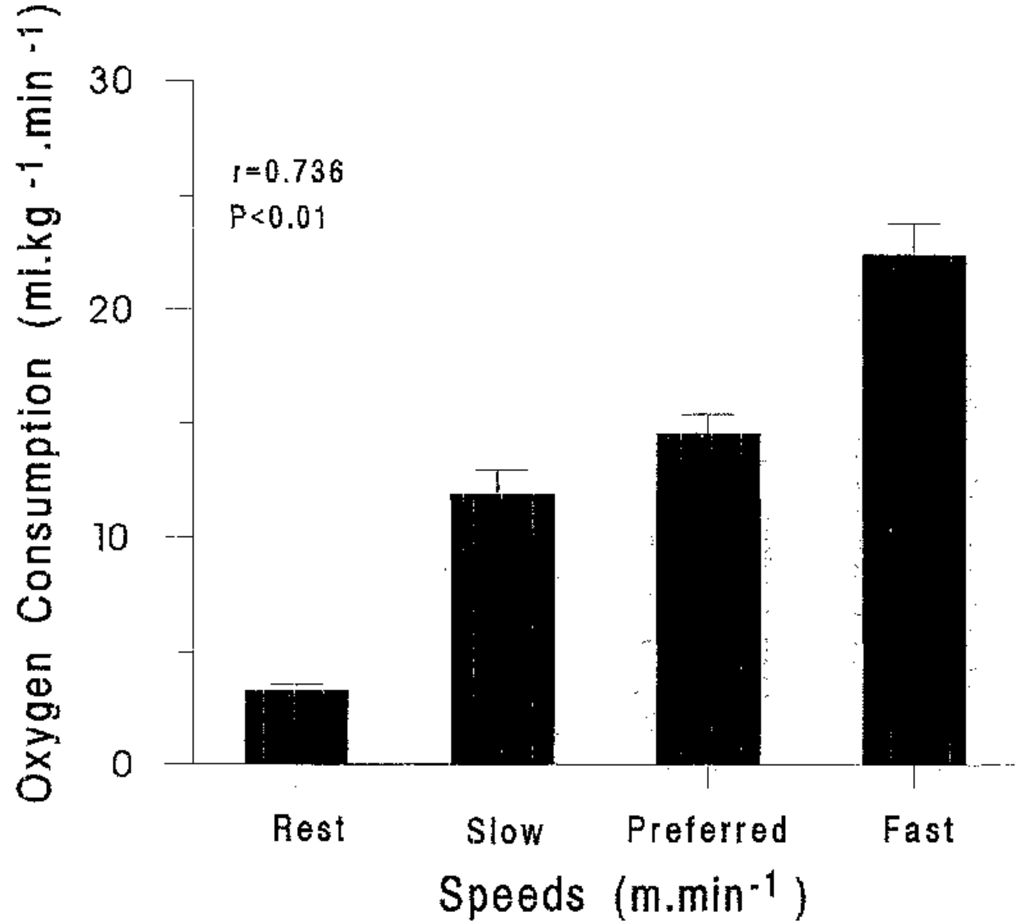

FIG. 3.14
The mean+SE of oxygen consumption at rest and during crutch walking at three speeds, in 10 able-bodied subjects

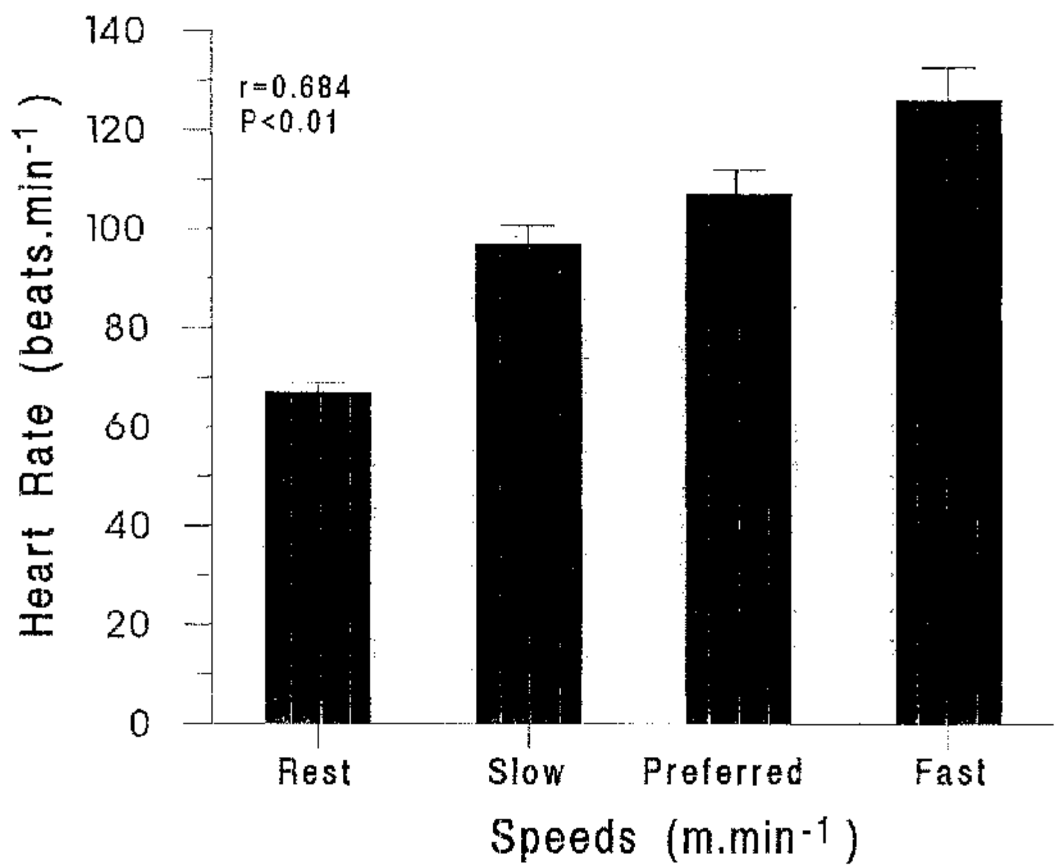

FIG. 3.15

The mean+SE of heart rate at rest and during crutch walking at three speeds, in 10 able-bodied subjects

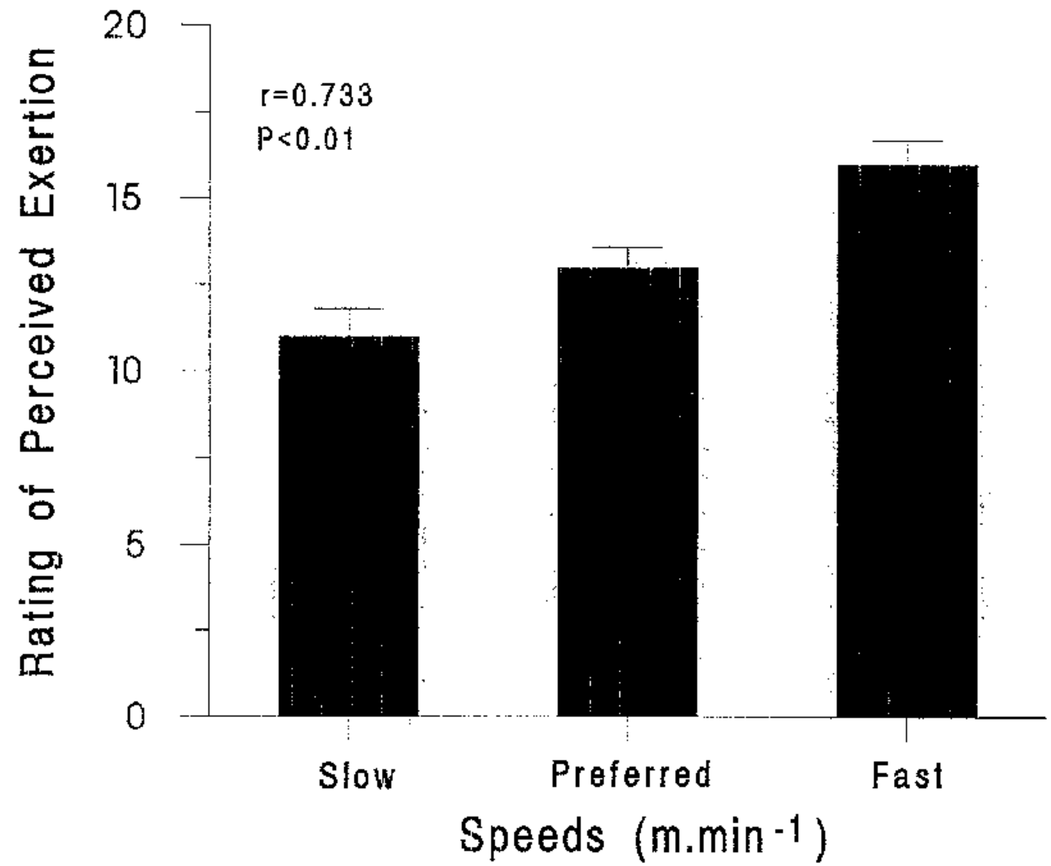

FIG. 3.16

The mean+SE of rating of perceived exertion during crutch walking at three speeds in 10 able-bodied subjects

Figures 3.17 and 3.18 show the mean ± SE of energy cost and physiological cost index of crutch walking at the 3 speeds in able-bodied subjects; and table 3.11 indicates that the two measures did correlate, though not particularly highly. The paired *t*-test did not show differences between preferred and fast speeds for either measure, but these speeds were more efficient for crutch walking than the slow speed.

3.4.4 COMPARISON OF MEASURED VARIABLES BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

Figures 3.19, 3.20 and 3.21 respectively indicate the energy cost, physiological cost index and perceived exertion, and mean ± SE for able-bodied subjects and lumbar paraplegics. Data are also tabulated separately for each of the five paraplegics in table 3.12.

As already noted, preferred speed of crutch walking was lower in thoracic paraplegics than in lumbar and able-bodied subjects; nevertheless energy cost and physiological cost index, even at these *speeds*, were higher (figures 3.19 and 3.20). However, perceived exertion was higher in only one of the two thoracic-lesion patients (fig. 3.21).

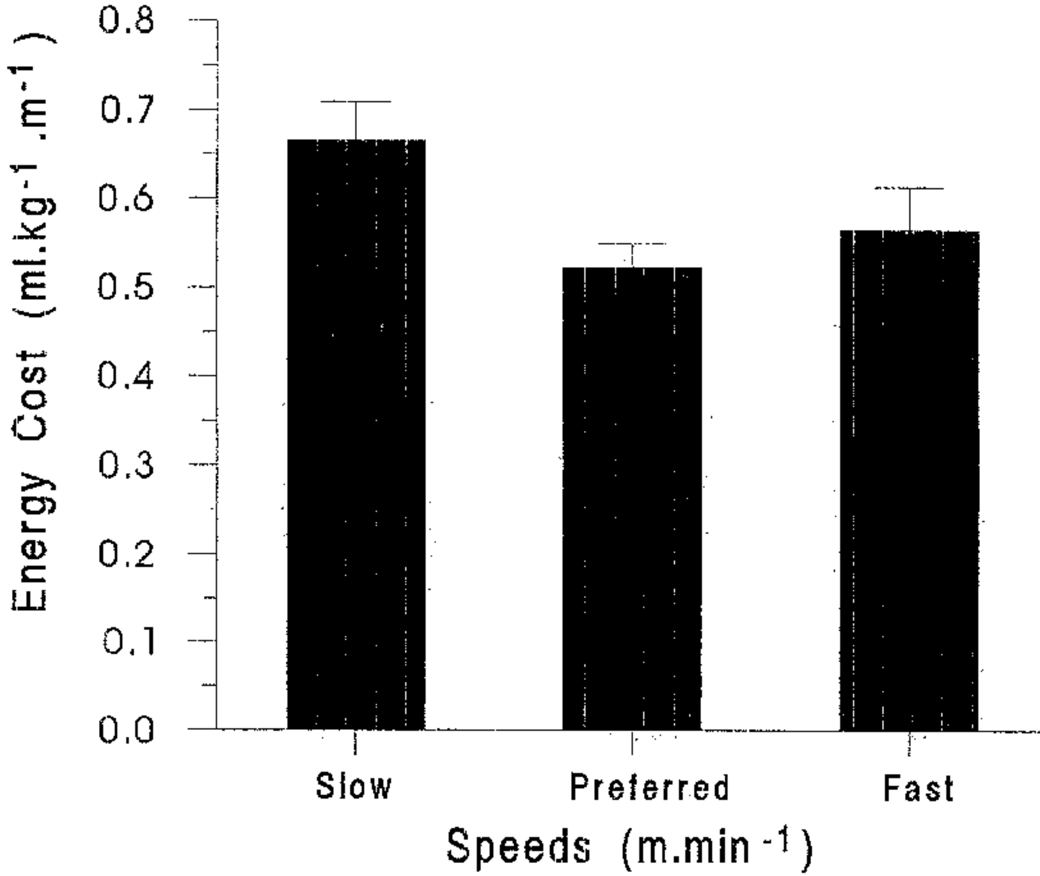

FIG. 3.17

The mean+SE of energy cost during crutch walking at three speeds in 10 able-bodied subjects

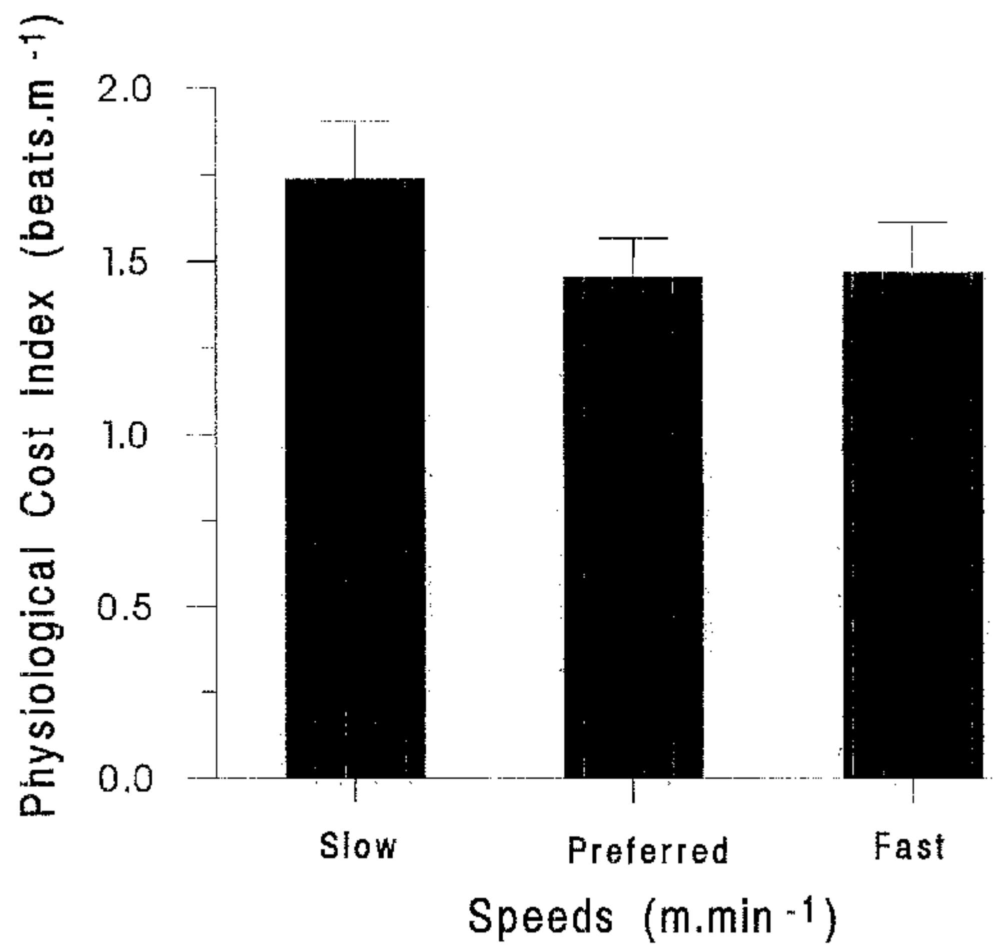

FIG. 3.18

The mean+SE of physiological cost index during crutch walking at three speeds in 10 able-bodied subjects

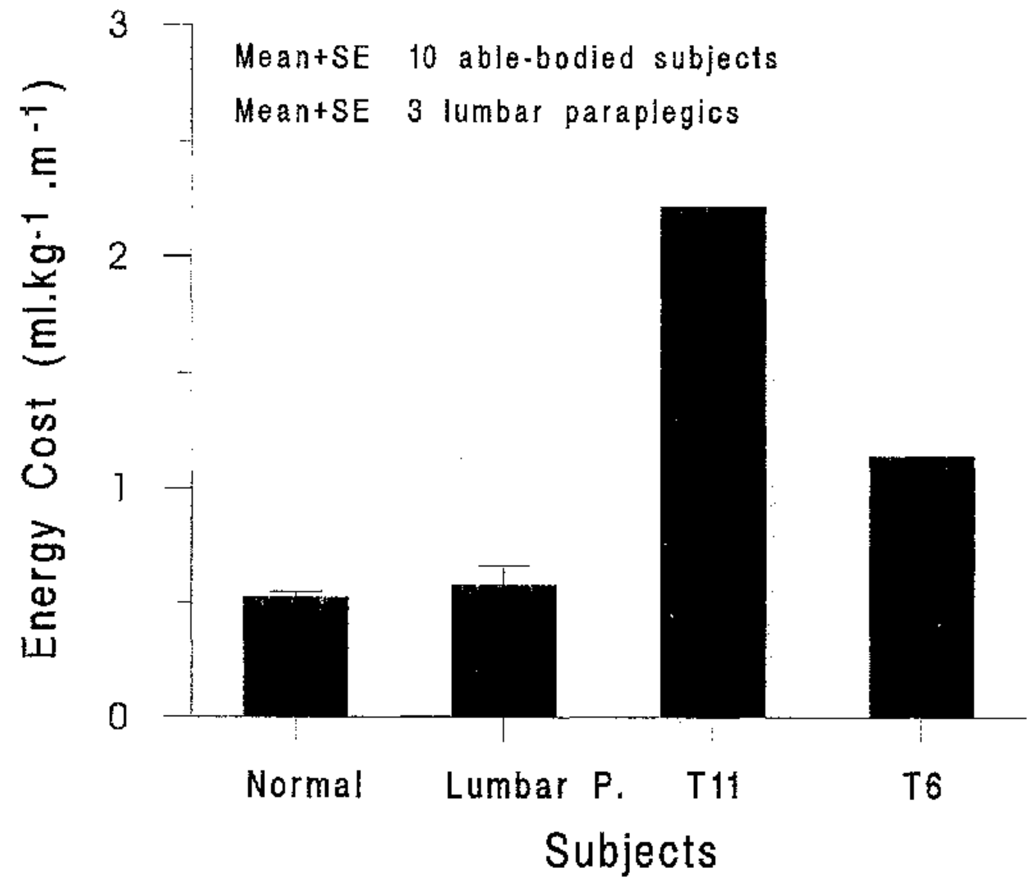

FIG. 3.19

The mean+SE of energy cost in 10 able-bodied subjects, 3 lumbar paraplegics and energy cost of thoracic paraplegics during crutch walking at preferred speed

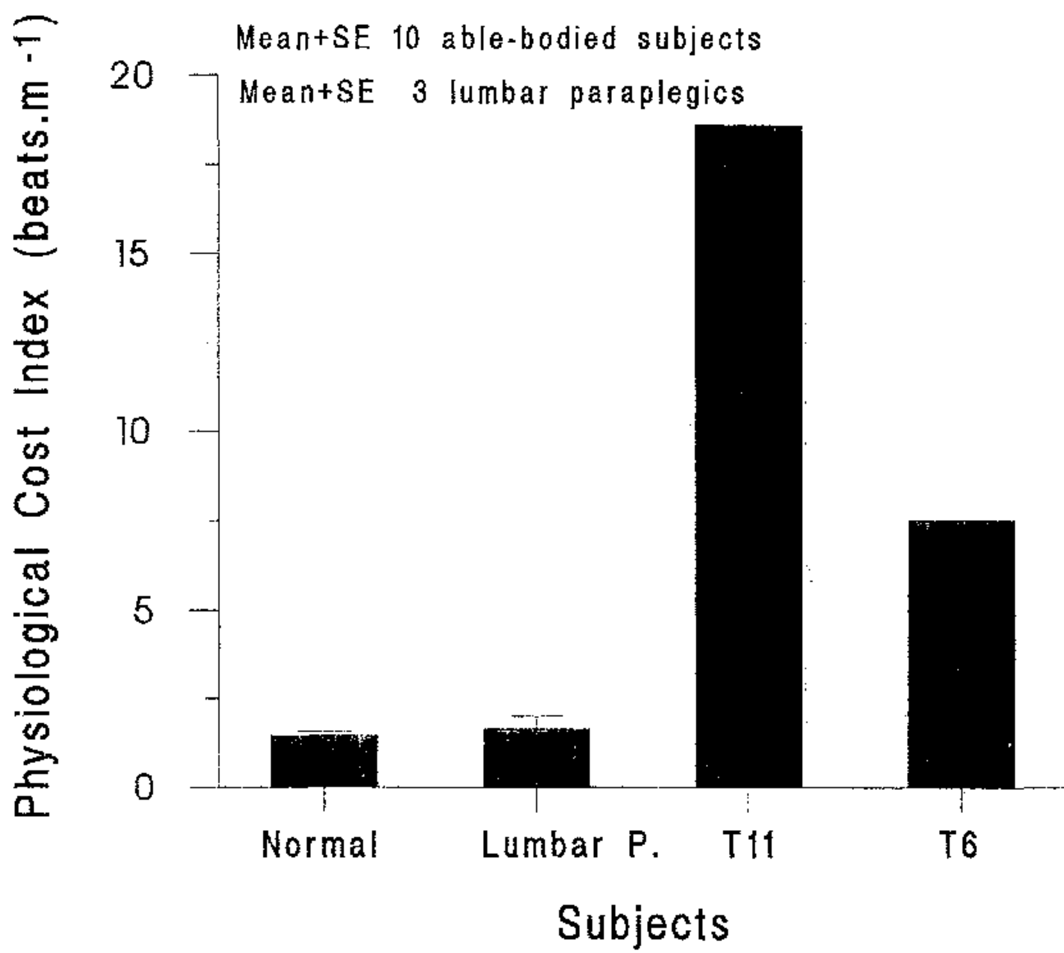

FIG. 3.20

The mean+SE of physiological cost index in 10 able-bodied subjects, 3 lumbar paraplegics, and cost index of two thoracic paraplegics during crutch walking at preferred speed

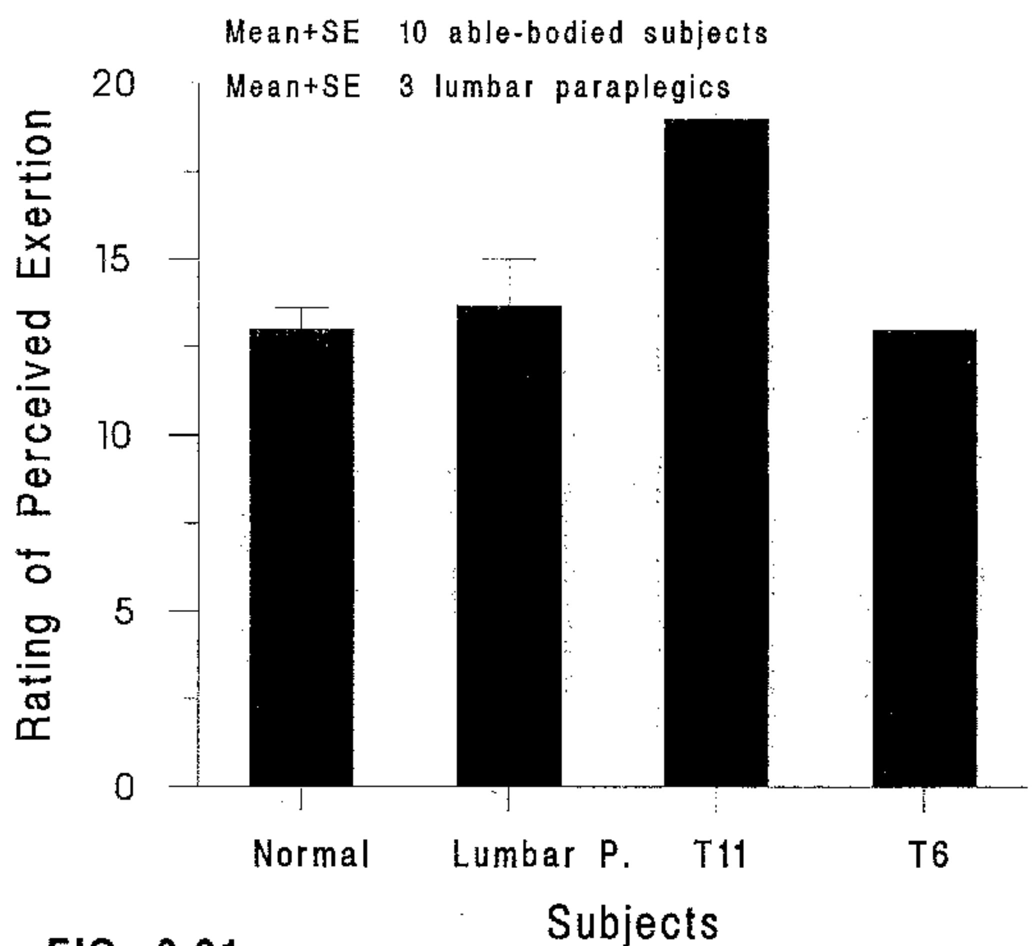

FIG. 3.21

The mean+SE of rating of perceived exertion in 10 able-bodied subjects, 3 lumbar paraplegics, and perceived exertion of two thoracic paraplegics during crutch walking at preferred speed

Volunteers	Level of lesion	HR (beats.min^{-1})	$\dot{V}O_2$ (ml.kg^{-1}.min^{-1})	Energy Cost (ml.kg^{-1}.m^{-1})	PCI (beats.min^{-1})	Speed (m.min^{-1})	RPE
DT	T_6	132	10.60	1.14	7.53	9.301	13
RO	T_{11}	178	12.06	2.22	18.60	5.43	19
MG	L_1	156	21.60	0.49	2.08	43.85	15
MM	L_2	103	19.43	0.49	0.94	39.55	11
CD	L_2	123	15.32	0.75	1.96	20.45	15
Lumbar paraplegics		127±15	18.78±1.87	0.58±0.09	1.66±0.37	34.62±7.33	13.7±1.4
Able-bodied subjects		107±5.0	14.58±0.84	0.52±0.03	1.46±0.11	28.05±1.463	13±0.6

TABLE 3.12
Heart rate, oxygen consumption, energy cost, physiological cost index, speed and rating of perceived exertion during crutch walking at preferred speed in thoracic and lumbar paraplegics. Mean±SE of these variables in lumbar paraplegics and able-bodied subjects are also indicated.

CHAPTER 4

DISCUSSION

4.1. INTRODUCTION

"Individuals with lower limb paralysis, such as those disabled by spinal cord injury, usually depend on upper body exercise (e.g., arm cranking, wheelchair propulsion) for daily locomotion" (Davis et al., 1990). For designing the wheelchairs and walking aids, the assessment of energy expenditure becomes important in spinal cord injuries e.g., paraplegics during arm exercise.

During exercise in able bodied subjects, a redistribution of blood takes place to increase cardiac output and to supply the exercising muscles with blood (Powers & Howley, 1990). However, when paraplegics perform arm exercise, impaired sympathetic innervation below the level of lesion and non-functional lower limb skeletal muscles disturb the redistribution of blood (Hopman et al., 1992b). Venous pooling in the lower limbs of paraplegics has been reported (Hjeltnes, 1977; Davis & Shephard, 1988; Kinzer & Convertino, 1989; Davis et al., 1990; Hopman et al., 1992a; Hopman et al., 1993b; Davis et al., 1993). Thus, the physiological responses to seated and upright arm exercise cannot be the same between paraplegic and able-bodied subjects.

This project compares responses to different modes of arm activity between paraplegic and able-bodied subjects. In this study, not only were the physiological responses to two modes of arm exercise (i.e., seated arm crank ergometry and crutch walking) compared between the two groups of subjects, but they were also compared during arm crank ergometry in two different postures (seated and upright) at the same power outputs.

The results are discussed separately in the 3 following sections and all is summarised together in the conclusion (section 4.6). The 3 following sections consist of:

i) Assessment of energy expenditure during seated arm crank ergometry

ii) The effect of posture on the oxygen consumption, heart rate and perceived exertion during arm crank ergometry

iii) Oxygen consumption, heart rate and perceived exertion responses to crutch walking in paraplegic and able-bodied subjects.

4.2. ASSESSMENT OF ENERGY EXPENDITURE DURING SEATED ARM CRANK ERGOMETRY

4.2.1. INTRODUCTION

The testing of cardiorespiratory responses to arm exercise is necessary whenever subjects cannot use their legs, such as the case with paraplegics and amputees (Bar-Or & Zwiren, 1975). However, the responses of the various indicators (oxygen consumption, heart rate and perceived exertion) to arm exercise in paraplegics might not be the same as in able-bodied subjects due to the circulatory problems mentioned earlier (c.f., Bar-On & Nene, 1990).

In the following section, the responses to seated arm crank ergometry are compared between paraplegic and able-bodied subjects.

4.2.2. TIME COURSE OF HEART RATE CHANGES

There is no published data, to my knowledge, showing the time course of heart rate during seated arm crank ergometry in paraplegics. Some workers (Jehl et al., 1991; Hopman et al., 1992a & 1992b) mentioned that the heart rate reached to steady state in their arm cranking experiments. Even at power output up to 80% of $\dot{V}_{O_2}$max (about 100 watts), Hopman et al., (1993b) reported that the heart rate stabilised within 4 minutes in complete thoracic paraplegics.

In the present project, virtually stable heart rate was achieved within 2 minutes at all 3 work rates, in both paraplegic and able-bodied subjects. These results lend some support to the statement of the above workers, but the paraplegics here reached only half as high a work-intensity. (Recalling that my subjects were sportsmen, yet some of them found 40 watts a rapidly fatiguing load, it would be interesting to know more details of the Hopman subjects).

4.2.3. RELATIONSHIP BETWEEN MEASURED VARIABLES

In paraplegics with lower lesions, a linear relationship between oxygen consumption and power output was reported by Hjeltnes (1977) similar to that in able-bodied subjects (Astrand & Rodahl, 1986). However, as already noted the heart rate responses to arm exercise particularly in high thoracic level paraplegics could be different from normal. Therefore, if the heart rate is to be used as an indicator of exercise intensities in paraplegics generally, linearity of its relationship with oxygen consumption must be established.

Several researchers (Bar-On & Nene, 1990; Jehl et al., 1991; Hooker et al., 1993) have recently reported a linear relationship between heart rate and oxygen consumption in paraplegics during seated arm crank ergometry. My results confirmed these previous studies. In addition, in the present study, the time course of heart rate adaptation is also indicated and the stabilisation found makes the following statement of the previous workers stronger, i.e., "Heart rate can be confidently used as an indicator of aerobic exercise intensity in paraplegics" (Hooker et al., 1993) - provided the investigation is confined to seated work.

4.2.4. COMPARISON OF OXYGEN CONSUMPTION BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

A few experiments were carried out to compare the oxygen consumption response to arm exercise at a similar power output between paraplegic and able-bodied subjects. No significant difference was found. This is in line with previous results (Hjeltnes, 1977; Kinzer & Convertino, 1989).

Futhermore, the average oxygen consumption at 40 watts (0.95 l . min^{-1}) and 35 watts (0.85 l . min^{-1}) in the studies of Hjeltnes (1977) and Kinzer & Convertino (1989) respectively, were near to the average oxygen consumption in the present project (0.930 l . min^{-1} at 40 watts). In addition, in able-bodied subjects, the average oxygen consumption during arm cranking at 50 watts (about 1 l . min^{-1}) and 35 watts power output (0.80 l . min^{-1}) reported by Vokac et al. (1975) and Kinzer & Conventiro (1989) respectively, are comparable with the mean for my able-bodied subjects at 40 watts (0.934 l . min^{-1}).

4.2.5. COMPARISON OF HEART RATE BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

Hjeltnes (1977) found a higher heart rate in 9 paraplegics (T_6 - T_{12}) than in able-bodied subjects during seated arm crank ergometry at 50%, 70% and 90% of their individual $\dot{V}_{O_2}$max values. In addition, he reported lower cardiac output, higher arterial-venous oxygen difference and higher lactate concentration in the paraplegics. Lower venous return in paraplegics, resulting in lower stroke volume, was the suggested explanation.

In 1989, Kinzer & Convertino showed higher heart rate in 5 paraplegics (T_6 - T_{11}) than in 5 amputees and 5 able-bodied subjects during seated arm crank ergometry at 35 watts (about 45% of the $\dot{V}O_2$max). They suggested that three factors could be taken into consideration for this higher heart rate in paraplegics.

i) Poor physical training level of paraplegics. They rejected this factor in their experiments due to the similar $\dot{V}O_2$max, work rate and time of exhaustion at maximal exercise between paraplegic and able-bodied subjects.

ii) Impaired sympathetic innervation and the control of arterial vaso-constriction. Their results showed a similar change of the leg arterial blood flow in both groups of subjects during exercise. They suggested that the sympathetic control of arterial tone during exercise was intact in their paraplegics.

iii) Lack of muscular pumping. This appeared to be the only mechanism not eliminated. Thus, they concluded that the higher heart rate in paraplegics must be explained by the lack of normal muscular pumping action.

It should be pointed out that *veno*constriction is one of the important factors for venous return of blood during exercise and smooth muscles of veins are innervated by sympathetic nerves - a factor which was not explicity considered by Kinzer & Convertino (1989).

The results of Hopman et al. (1992b) showed higher heart rate in 11 paraplegics (T_6 - T_{12}) than in 11 able bodied subjects during seated arm crank ergometry at 40% (1.04 l . min^{-1}) and 60% (1.43 l . min^{-1}) of $\dot{V}O_2$max, though the difference at 20% (0.73 l . min^{-1}) was not significant. They suggested that the higher heart rate at the last 2 work rates in paraplegics was caused by the lack of *both* sympathetic vasomotor regulation *and* the muscular pump.

In another study, Hopman et al. (1993c) did not find a significant difference of heart rate between paraplegic and able-bodied subjects during arm crank ergometry at 50% of $\dot{V}_{O_2}$max. This conflicted with the results of their previous study (1992b). This insignificant difference of heart rate was explained by heterogeneity; "For example, the paraplegic group varied in the level and completeness of the lesions and the control group varied in training status compared with other studies (Hjeltnes, 1977; Davis & Shephard, 1988; Hopman et al., 1992a)" (Hopman et al., 1993c).

In addition, the training level of paraplegics could have a significant effect on their cardiovascular responses to arm exercise. The cardiovascular responses to arm crank ergometry were studied by Davis & Shephard (1988) in active and inactive paraplegics. The active group participated at least 3 times / week in a variety of sports for the disabled, including basketball, swimming and weightlifting. Inactive ones did none of these things. Their results showed that cardiac output and stroke volume were 40% higher in active than in inactive paraplegics. This was explained by i) greater use of middle and lower body movement by the active subjects for the generation of force (assisting venous return), ii) a reduced cardiac afterload in these subjects due to upper body hypertrophy in the active paraplegics (i.e., greater activation of arm muscles, allowing a larger blood flow to the working tissues).

By contrast, I have not found a significant difference of heart rate between paraplegic and able-bodied subjects during seated arm crank ergometry. This can be attributed to the following points:

i) Compared with the untrained subjects of previous studies, the sports-trained paraplegic subjects of my study had higher training level and also, lower levels of lesion (middle and lower thoracic, lumbar).

For example, the average heart rate (±SE) in Kinzer & Convertino's results was 123±2 beats . min^{-1} in paraplegics. Despite the lower average of oxygen consumption (0.85 l . min^{-1}) and power output (35 watts) than in my experiments (oxygen consumption: 0.934 l . min^{-1} at 40 watts), their average heart rate was higher than the average of my study (114±6 beats . min^{-1}) during seated arm crank ergometry by paraplegics.

ii) My project compared the physiological variables between the sports-trained paraplegics, where lower heart rate than non sports-trained paraplegics might be expected, and untrained able-bodied subjects, where higher heart rate than trained able-bodied subjects might be expected.

4.2.6. COMPARISON OF PERCEIVED EXERTION BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

Rating of perceived exertion was previously used by a few authors (Gayle et al., 1990; Eriksson et al., 1988) as an index for the strain of physical activity in paraplegics during maximal arm ergometry. To my knowledge, there is no published data to compare the perceived exertion responses to submaximal arm exercise between paraplegic and able-bodied subjects.

The factors responsible for the perceived exertion are multiple and complex. The individual will however evaluate his perceived exertion during physical work due to at least two factors (Ekblom & Goldbarg, 1971).

i. Local factors: a muscular response for perception of effort as a local factor is based on the mediation of feelings of strain in the exercising muscles. The parameters, which provide sensory input for perceived exertion, may include muscle lactate, Golgi tendon activity and general muscle sensation (Mihevic, 1981). More recent thinking would suggest that other muscle metabolites, and K^+ ions should be considered alongside lactate.

ii. Central factors: they consist of the pulmonary ventilation and circulation and perhaps direct sensory effects of muscle metabolites acting upon receptors located centrally.

In paraplegics, during exercise above the level of lesion (e.g., arm crank ergometry) both local and central pathways are intact, as in able-bodied subjects. In this study, perceived exertion did not show any significant differences between paraplegic and able-bodied subjects during seated arm crank ergometry. This agrees with oxygen consumption and heart rate responses to seated arm crank ergometry.

It should be noted that during exercise below the level of lesion in paraplegics (e.g., electrically induced leg cycle ergometry) the local factors cannot be involved as inputs for perceived exertion because the afferent nerves are not intact. Thus, the perceived exertion responses to exercise below the level of lesion in paraplegics might not be the same as in able-bodied subjects; no published data in this area is available, but from the point of view of fundamental physiology the experiment seems worth doing, since it might help to distinguish between local and central contributions to perceived exertion.

4.2.7. SUMMARY

Therefore, in spite of theoretical expectations based on impaired venous return, and the supporting experimental results of some previous workers (Hjeltnes, 1977; Kinzer & Convertino, 1989; Hopman et al., 1992c), in this study all oxygen consumption, heart rate and perceived exertion responses to seated arm crank ergometry up to 40 watts did not show significant differences between sports-trained paraplegics and able-bodied subjects.

4.3. THE EFFECT OF POSTURE ON THE OXYGEN CONSUMPTION, HEART RATE AND PERCEIVED EXERTION

4.3.1. INTRODUCTION

"While vigorous efforts are made by rehabilitation teams to allow persons with spinal cord injury to return to pre-injury activities, the physiological aspects of the individual's potential are often unknown" (Pare et al., 1993).

From the physiological viewpoint, the venous return is more important for arm activity in the upright than in the seated position, due to the influence of gravity. There are 3 main mechanisms for increasing venous return during exercise, i.e.:

i. Venoconstriction which occurs via a reflex sympathetic constriction of smooth muscle in the veins. This reduces their volume capacity, allowing more blood to occupy other regions of the cardiovascular system and results in an increased flow of blood toward the heart.

ii. The muscle pump, a result of the mechanical action of skeletal muscle contractions. This compresses veins and pushes blood toward the heart. In addition, the one-way valves, located in large veins, prevent blood flowing away from the heart.

iii. Respiratory pump i.e., a rhythmic pattern of breathing which provides a mechanical pump. The decrease of pressure in thorax and the increase of abdominal pressure occur during each inspiration. This causes the blood to move from the abdominal region into the thorax.

In addition, during arm exercise, although there is the vasodilatation of arterioles in active arm skeletal muscles, there is vasoconstriction of arterioles in less active organs.

In paraplegics, because of the paralysis of skeletal lower limb muscles and impaired sympathetic innervation below the level of lesion, venous return mechanisms and arteriolar vasoconstriction in the lower body during arm exercise will be prevented. This circulatory problem is likely to affect the cardiovascular responses to arm exercise.

Various attempts have been made to understand the circulatory problem in paraplegics (Bidart & Maury, 1973; Kinzer & Convertino, 1989; Hopman et al., 1993c) and also, to improve venous return from below the level of lesion (Davis et al., 1990; Hopman et al., 1992a). The effects were recently studied of leg muscle contractions induced by functional electrical stimulation (Davis et al., 1990) and of lower body positive pressure (Hopman et al., 1992a) on the cardiovascular responses to submaximal arm exercise in paraplegics. Both of these experiments investigated if venous return was lower in paraplegics when the interventions were not made:their conclusions are considered in section 4.3.3 below.

Despite the possibility of circulatory problems in paraplegics some recent publications (Bowker et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993) have used a heart-rate based indicator, namely the physiological cost index, for assessment of energy expenditure during crutch walking in paraplegics. Physiological cost index is certainly a convenient method of measuring the physiological cost of walking (MacGregor, 1979); combining speed and heart rate and indicating locomotion efficiency. However, it appears that the use of physiological cost index in paraplegic ambulation was based on the linear relationship between heart rate and oxygen consumption during arm crank ergometry in the *seated* position. In other positions, however, the cardiovascular responses to postural change in paraplegics might not be the same as in able-bodied subjects.

In the following section, the physiological responses to arm exercise in seated and upright postures at the same power output are compared between paraplegic and able-bodied subjects. There were 2 objectives with this experiment:

i) Confirming the circulatory problems in paraplegics reported by previous researchers such as, Davis et al. (1990) and Hopman et al. (1992a)

ii) Comparing the physiological responses to arm activity in different postures as a common point for comparison of seated arm ergometry and crutch walking.

4.3.2. TIME COURSE OF HEART RATE CHANGES

The sympathetic outflow to the heart arises above the spinal level of T_6. Patients with lesion above T_6, may then have abnormal cardiac responses to exercise (Bowker et al., 1992). Hoffman (1986) stated that there is generally an upper limit of heart rate between 110 to 130 beats . min^{-1}, due to the impaired sympathetic cardiac innervation in quadriplegic patients. McLean et al. (1992) studied the relationship between heart rate and oxygen consumption during arm crank ergometry in quadriplegics. Their results displayed the heart rates regressed on oxygen consumption values and demonstrated that heart rate does not reflect exercise intensity in all quadriplegics. Coutts et al. (1983) showed that the maximal heart rate in T_1 to T_5 paraplegics (between 135 and 180 beats . min^{-1}) is lower than that in middle and low thoracic paraplegics. This lower maximal heart rate was also indicated by Hooker et al. (1993) in nine high lesion (T_1 to T_5) paraplegics. This has been explained by partial loss of cardiac sympathetic innervation (Coutts et al., 1983). However, Bar-On & Nene (1990) have reported that there is an *increase* in heart rate up to maximal value in paraplegics with apparently complete T_3 to T_6 lesions. The reason for this discrepancy is not clear.

In the results of the present project, the heart rate responses to upright arm crank ergometry were not the same for all paraplegics. Figure 3.7 illustrates a limitation of heart rate at 135 beats . min^{-1} in the complete T_5 patient during upright arm crank ergometry. This result agrees with the view of those previous workers who found reduced maximum cardiac responses to exercise in high level paraplegics.

In the complete T_8 paraplegic, figure 3.7 shows unstable heart rate at 40 watts power output, and the difference in heart rate between the two last min of 40 watts work rate was 10 beats. However, when the same conditions were applied for seated arm crank ergometry, stable heart rates were observed, and were the same as for able-bodied subjects at all 3 work rates. The used of heart rate, as an indicator for assessment of energy expenditure during upright arm activities (e.g., crutch walking) in thoracic paraplegics, can thus already be questioned on the following grounds:

a) Where heart rate is unstable, when it should be taken is a problem.

b) Where heart rate quickly plateaus or is permanently high, plainly it cannot be use as a measure of anything.

Even where neither of these problems apply, other difficulties may remain. The following subsections illustrate these.

4.3.3. COMPARISON OF POSTURAL EFFECT ON HEART RATE RESPONSE BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

The difference between circulatory responses to arm and leg exercise is well known (e.g., Bevegard et al., 1966; Stenberg et al., 1967). Several investigators have reported a higher heart rate and lower stroke volume at submaximal oxygen consumption during seated arm crank ergometry than during cycle exercise (Sawka, 1986). These results led some authors to the concept that at least a part of the lower stroke volume during arm exercise is explained by there being less venous return (Stenberg et al., 1967). In contrast, reduced venous return during arm exercise in able-bodied subjects was rejected by Kinzer & Convertino (1989). They found similar responses of heart rate and stroke volume during arm crank ergometry (at 45% of $\dot{V}_{O_2}$max) in able-bodied subjects and amputees, who have negligible leg vasculature. They concluded that this insignificant difference was due to muscular pump action even in seated able-bodied subjects.

However, it is thought elsewhere that the inactive lower limb muscles during arm crank ergometry cannot considerably increase venous return. In addition, Hopman et al (1993c) recorded electromyographs from the calf muscles of able-bodied subjects during seated arm crank ergometry, and reported no muscle activity. Thus, the level of amputation (at and above knees) and the low intensities of exercise (45% of $\dot{V}_{O_2}$max of arm exercise) should be pointed to as a part of the reasons for the insignificant difference found by Kinzer & Convertino (1989).

Change of posture causes considerable differences in heart rate and stroke volume even when the same exercise - or none- is considered (Jennett, 1989). With regard to the reduced venous return in the upright able-bodied, Ng et al. (1987) investigated the effect of lower body positive pressure (52 mmHg) on the cardiovascular responses to arm exercise, in twelve able-bodied subjects. They found that stroke volume increased, even in these subjects. Thus, despite intact somatic and autonomic nervous systems, venous return during arm crank ergometry is reduced by static upright posture when external compression is not applied.

On the other hand, Hopman et al. (1992a) investigated the effect of lower extremity compression on the cardiovascular responses in paraplegics during arm crank ergometry. Their experiments showed that the effect of an anti-gravity suit on the heart rate was significantly different between paraplegic and able-bodied subjects; i.e.,. lower heart rates were seen in paraplegics with anti-gravity suits. They therefore concluded that "The significant difference in the effect of the anti gravity suit on heart rate between both groups seems to corroborate the hypothesis of venous blood pooling in persons with paraplegia".

In paraplegics, the venous return mechanisms and vasoregulation are disturbed. Therefore, one would expect a greater reduction in venous return on being raised to an upright position when compared to able-bodied subjects. Two studies have attempted to show the lower venous return from paralysed leg muscles of paraplegics. The first (Hopman et al., 1993c) investigated the volume changes in legs during arm exercise by using plethysmography. They showed that the decrease of leg volume, during arm exercise, was significantly lower in paraplegic than in able-bodied subjects. From this, they concluded that the paraplegics are not able to redistribute fluid effectively below the level of lesion during arm exercise.

The second type of experiment was to examine the effect on venous return of functional electrical stimulation of lower extremities. In 1990, Davis et al. used functional electrical stimulation during arm exercise to increase venous return. Although, they only tested paraplegics without an able-bodied control group, their results showed that electrically-induced muscle contraction increased stroke volume in paraplegics. However, in spite of the theoretical appeal of this result, it is thought that the above experiment cannot be taken to indicate the circulatory problems peculiar to paraplegics, because the use of functional electrical stimulation in able-bodied subjects has also elicited increases in stroke volume and venous return (Rattan et al., 1985).

In line with these previous studies, two different body positions were applied in the present project and then the heart rate responses to arm exercise were compared between paraplegic and able-bodied subjects. The results confirmed the previous studies. As figure 3.9 illustrates, the increase of heart rate in upright compared with seated arm crank ergometry was significantly greater in paraplegics (16±2 beats . min^{-1}, i.e., mean±SD for three work rates) than in able bodied subjects (8±1 beats . min^{-1}, i.e., mean±SD for three work rates).

4.3.4. COMPARISON OF OXYGEN CONSUMPTION RESPONSE TO POSTURAL EFFECT BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

The results of this study confirm the report of Vokac et al, (1975) that, in able-bodied subjects, there is no significant difference in oxygen consumption between seated and upright arm crank ergometry at submaximal power output. However, oxygen consumption has also here been measured during both seated and upright arm crank ergometry in paraplegics.

The effect of posture on the oxygen consumption was not significant in either group of subjects. In addition, there was no significant difference between the oxygen consumption changes (from seated arm crank ergometry to upright arm crank ergometry) between paraplegic and able-bodied subjects.

In parenthesis, it should be recognised that the $\dot{V}_{O_2}$max or maximal power output is not the same between paraplegic and able-bodied subjects during arm crank ergometry. Lower $\dot{V}_{O_2}$max in paraplegics, compared with able-bodied subjects, has been reported during seated arm ergometry. Assuming that the state of upper body training is not less in the paraplegics, the lower $\dot{V}_{O_2}$max can be explained by two main factors (Eriksson et al., 1988) i.e., the smaller active muscle mass and the diminished venous return which (as discussed in 4.2.5 and 4.3.1) limits maximal cardiac output. With regard to these limiting factors, functional electrical stimulation has been reported to enhance $\dot{V}_{O_2}$max due to the increased active muscle mass and consequently, venous return. By contrast, there is no significant effect on the $\dot{V}_{O_2}$max of applying lower body positive pressure (Hopman et al., 1993a) has partly been explained by the influence of the other factors e.g., the functional capacity of lungs, muscle mitochondria and aerobic enzymes on the maximal performance. In addition, unlike functional electrical stimulation, lower body positive pressure does not enlarge the active muscle mass. The insignificant $\dot{V}_{O_2}$max effect of merely reducing venodilation may be compared with account given by Kinzer & Convertino (1989) of their heart rate experiments, which also ignored the venous pool (c.f., page 59). For the purposes of this thesis, however, the most important aspect of the oxygen consumption results, in the prior literature as well as my own experiments, is not the value of $\dot{V}_{O_2}$max but the linear relation between work rate and oxygen consumption at submaximal levels. This was found in all paraplegics, whatever their level of lesion. The contrast with the heart rate and work rate relationship, in high lesion patients, was noted.

4.3.5. COMPARISON BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS OF POSTURAL EFFECT ON PERCEIVED EXERTION RESPONSE

In 1962, Borg proposed perceived exertion as a psychological complement to physiological responses during exercise (Ekblom & Goldbarg, 1971). Stamford (1976) also showed a strong relationship between perceived exertion and work intensity during exercise in able-bodied subjects.

The results of this project in paraplegics also show significant correlation between perceived exertion and heart rate, oxygen consumption and power output during both seated and upright arm crank ergometry. However, figure 3.10 illustrates that the increased perceived exertion in upright compared with seated arm crank ergometry was higher ($P<0.07$) in paraplegics than able-bodied subjects. This might not be considered surprising, because of lack of familiarity of paraplegics with upright arm activity and their difficulties in performing it. However, the perceived exertion is also linked to heart rate responses (Borg, 1982) and paraplegics showed higher increased heart rate to upright arm crank ergometry. These is no evidence upon which to discriminate between these explanations.

4.3.6. SUMMARY

My results have indicated that, despite an insignificant difference in the effect of posture on oxygen consumption, heart rate responses to change of posture were significantly different between paraplegic and able-bodied subjects. The greater response of heart rate, to inceased power output in upright paraplegics can be explained as a compensatory response to reduced stroke volume. The higher perceived exertion ($P<0.07$) during arm crank ergometry in upright than seated paraplegics also correlates with heart rate, though factors explaining this analysis have just been noted.

As a result, firstly, the more steeply increasing heart rate during upright arm activity, tallies with the concept of lower venous return in paraplegics. Secondly, unstable heart rate during upright arm crank ergometry in thoracic paraplegics, made the assessment of energy expenditure by a heart rate based method (i.e., physiological cost index) difficult and unreliable, in this subset of patients.

4.4. A COMPARISON OF ASSESSMENT OF ENERGY EXPENDITURE BETWEEN SPORTS-ACTIVE AND NON-SPORTS PERSONS WITH SPINAL CORD INJURY DURING ARM CRANK ERGOMETRY

The recognition of the importance of sport in the rehabilitation of patients with spinal cord injury has led to the investigations of the physiological benefits of training programs on spinal cord injured individuals. For instance in 1994, Kaprielian et al. compared the effect of lower body positive pressure on cardiovascular responses to seated arm exercise at 50% and 80% $\dot{V}_{O_2}$max in 8 trained and 10 untrained paraplegics with complete lesions between T_6 and T_{12}. They stated that "A lower heart rate and upward trend in stroke volume during submaximal exercise in paraplegics suggests that lower body positive pressure offers positive hemodynamic benefits with the untrained group showing a stronger benefit than the trained group, suggesting a possible venous pooling".

With regard to the results of the comparison of cardio-respiratory responses between sports and non-sports paraplegics during both seated and upright arm crank ergometry in the present study (section 3.3), the very small sample of non-sporting paraplegics, however, gave no support to the assumption, embodied in my initial design, that sports-active people would be fitter than other paraplegics. But it did not contradict it either. The clearest message is that only patients with equivalent lesions (equivalent in level and severity) can usefully be compared.

4.5. RESPONSES OF OXYGEN CONSUMPTION, HEART RATE AND PERCEIVED EXERTION TO CRUTCH WALKING: COMPARISON BETWEEN PARAPLEGIC AND ABLE-BODIED SUBJECTS

4.5.1. INTRODUCTION

This part of the study is discussed in two separate subsections i.e.:

i. Oxygen consumption, heart rate and perceived exertion responses to crutch walking in able-bodied subjects

ii. Comparison of oxygen consumption, heart rate and perceived exertion responses to crutch walking between paraplegics and able-bodied subjects

4.5.2. OXYGEN CONSUMPTION, HEART RATE AND PERCEIVED EXERTION RESPONSES TO CRUTCH WALKING IN ABLE-BODIED SUBJECTS

4.5.2.1. INTRODUCTION

The purpose of assessment of energy expenditure in able-bodied subjects during crutch walking is to compare their data as a control group with paraplegics. The same conditions, as far as possible, have then been applied for both groups of subjects. Because the paraplegics use the calipers to stabilise their paralysed legs during crutch walking, the same type of caliper was also used for able-bodied subjects.

In the following subsections, time courses of heart rate, relationship between measured variables, energy cost, physiological cost index, perceived exertion and crutch walking speeds in 10 able-bodied subjects are respectively discussed.

4.5.2.2. TIME COURSE OF HEART RATE CHANGES

Patterson & Fisher (1981) reported a slow rising of heart rate during crutch walking at speeds of 30 to 70 m . min^{-1} in able-bodied subjects. By contrast, Annesley et al. (1990) in the same type of experiment at the speed of about 40 m . min^{-1} showed steady heart rates. Three main factors were considered for rising heart rate and its magnitude during exercise, i.e., power output (Broman & Wigertz, 1971), environmental conditions (Maxfield, 1971) and the activated muscle groups (Vokac et al., 1975).

In the present project the difference in heart rates between the two last minutes of exercise was only about 2 beats per minute, and these virtually stable heart rates were achieved in less than 2 minutes at all three speeds. It may be concluded that no significant recruitment of further motor units or muscle groups occurred during these observations.

4.5.2.3. RELATIONSHIP BETWEEN MEASURED VARIABLES

The results of this study showed that there are significant correlations between the different variables investigated during crutch walking in able-bodied subjects (table 3.11). The significant correlation between heart rate and oxygen consumption during axillary crutch walking agrees with the previous study of Ghosh et al. (1980).

4.5.2.4. ENERGY COST OF CRUTCH WALKING

"Axilla crutches have been in use for nearly 5000 years" (Sankarankutty et al., 1979). Efficiency of crutch walking has been measured in terms of energy used per unit body weight per unit distance travelled. This number increases as efficiency decreases and vice versa.

The energy costs of axillary crutch walking, found in previous studies, are indicated in table 4.1. The energy costs in the present study (0.523 ml . kg^{-1} . m^{-1} at preferred speed) are higher than those of previous studies (0.344 to 0.439 ml . kg^{-1} . m^{-1} is the range of energy cost from 4 studies i.e., Patterson & Fisher, 1981; Hinton & Cullen, 1982; Kathrins & Osullivan, 1983; Bhambani & Clarkson, 1989, all of which used axillary crutches without calipers). This could be explained by the use of calipers, as it has been reported that crutch walking with a cast has a higher energy cost than without cast (Imms et al., 1976). In addition, it should be considered that the small figure of eight track can play a role in diminishing the speed of crutch walking and increasing its energy cost. However, MacGregor (1979) stated that, "I believe the tightness of our track gives a more realistic view of walking than the usual unencumbered circumstances of the very large laboratory. Fairly sharp turns are also typical of most homes and workplaces".

4.5.2.5. PHYSIOLOGICAL COST INDEX DURING CRUTCH WALKING

In able-bodied subjects heart rate has been shown to increase linearly with power output and oxygen consumption at submaximal levels (Astrand & Rodahl, 1986). Furthermore, heart rate and walking speed measurements are found to provide possible means for assessment of energy expenditure (Rose et al., 1991). The method embodying these two variables (i.e., physiological cost index) was primarily established by MacGregor (1979) for assessment of energy expenditure during walking. However, crutch walking is upper body activity and cardiovascular responses to lower and upper body exercise are different. Several workers (e.g., Secher et al., 1974) compared cardiorespiratory responses during arm and leg activity. These studies were summarised by Sawka (1986) and the following differences were stated.

i. Maximal cardiac output and peak oxygen consumption are approximately 30% lower in arm crank ergometry than cycling.

ii. During upright body exercise, reduced skeletal muscle pump activity might fail to facilitate venous return and thereby lead to a decrease in the ventricular end-diastolic volume.

iii. There is a higher heart rate and lower stroke volume at a given oxygen consumption during arm cranking rather than cycling.

In accord with these generalisations, Bhambani & Clarkson (1989) compared the physiological responses during 3 modes of exercise i.e., axillary crutch walking, normal walking and running. Their results lead them to conclude that "Axillary crutch walking was the most stressful activity".

Some workers described unstable heart rates during crutch walking in able-bodied subjects (Bahambani & Clarkson, 1989; Patterson & Fisher, 1981). This made the assessment of energy expenditure from heart rate measurement during crutch walking difficult.

The present project has, however shown steady heart rate in able-bodied subjects (at slow and preferred speeds, if not at high). Mean physiological cost index in this study at preferred speed was 1.46 beats . m^{-1}. However, there are to my knowledge no published data, which used physiological cost index during crutch walking with or without callipers in able-bodied subjects, for comparing with these results.

4.5.2.6. PERCEIVED EXERTION DURING CRUTCH WALKING

"People primarily seek medical care because they feel ill, not for treatment of a special disease. Medical assistance is most frequently sought by a patient who has noted a severe decrease of their physical working capacity and a subsequent strain. In my opinion, perceived exertion is the single best indicator of the degree of physical strain" (Borg, 1982).

The perceived exertion quantified subjective feelings of effort during physical activity (Bhambani & Clarkson, 1989). However, despite the studies about the physiological responses to crutch walking in able-bodied subjects, no researchers have published experiments examining the perceptual responses during crutch walking. Rating of perceived exertion, recorded during crutch walking, supports the physiological evidence in my experiments and, at the same oxygen consumption and heart rate, it has shown a significant correlation with walking speeds.

4.5.2.7. SPEED OF CRUTCH WALKING

The preferred speed of crutch walking with calipers in this study (28 $m.min^{-1}$) was slower than in previous studies (Bhambani & Clarkson, 1981; Hinton & Cullen, 1982) which dealt with crutch walking without calipers (39 and 43 $m.min^{-1}$). The lower speed can be explained by the use of orthoses and the small size of the walkway available in our laboratory.

Figures 3.16 and 3.17 show that energy cost and physiological cost index at preferred and fast speeds are lower than at slow speed. This agrees with the previous studies (Ganguli et al. 1974; Ghosh et al., 1980). Ganguli et al. (1974), stated that the energy cost of walking is higher at slow speed because of the "uncomfortable speed". What this means physiologically is less than clear, but in the present study it was observed that, during crutch walking at lower speeds, the crutch users faced difficulties in maintaining balance and rhythm.

4.5.2.8. SUMMARY:

The energy cost of crutch walking observed in this project, including the high cost of the slow speed (Ganguli et al. 1974) is comparable with that found in previous studies (Patterson & Fisher, 1981; Hinton & Cullen, 1982; Kathrins & Osullivan, 1983; Bhambani & Clarkson, 1989), if allowance is made for the use of calipers (knee-ankle-foot orthoses).

Because these orthoses were used in the present study, the information obtained permits direct comparison of the cardiorespiratory responses to crutch walking between paraplegic and able-bodied subjects. Such a comparison does not seem to have been made before. Having talked about "able-bodied", it seems very asymmetrical not to talk directly about paraplegics, before comparing the two.

4.5.3. OXYGEN CONSUMPTION, HEART RATE AND PERCEIVED EXERTION RESPONSES TO CRUTCH WALKING IN PARAPLEGICS AND THEIR COMPARISONS WITH FINDINGS FROM ABLE-BODIED SUBJECTS

4.5.3.1. INTRODUCTION

It was noted in the overall introduction to this thesis that, while considerable efforts have been made by rehabilitation teams to develop and supply walking aids for paraplegics, it is known that many paraplegics do not use their orthoses after leaving the hospital. The high energy cost of paraplegic crutch walking may be responsible.

To date, several researchers (tables 4.1 and 4.2) have studied the energy cost of crutch walking in both paraplegic and able-bodied subjects. The stress of crutch walking activities compared with normal walking in able-bodied subjects as reported by Bhambani and Clarkson (1989) and also Cerny et al. (1980) have suggested that there is likely to be a high energy cost of crutch walking related to wheelchair propulsion in paraplegics. However, no study has compared both paraplegic and able-bodied subjects during crutch walking in the same conditions i.e., using of calipers for the able-bodied subjects. In addition, techniques for measuring energy cost are not standardised, as noted on page 65.

In the following part of this study, time course of heart rate change, energy cost, physiological cost index and perceived exertion in crutch walking with axillary crutches and knee-ankle-foot orthoses are compared between paraplegic and able-bodied subjects.

4.5.3.2. TIME COURSE OF HEART RATE CHANGES

The time course of heart rate change during crutch walking in paraplegics can be considered under two headings, i.e.:

i. The stabilisation of heart rate. This is necessary if an indicator based on heart rate is to be used for assessment of energy expenditure.

ii. The period of time needed for the stable heart rate to be achieved. Paraplegics are not able to continue the crutch walking for a long period of time, due to their low exercise capacity (Eriksson, et al., 1988), .

Some researchers (Merkel et al., 1984 & 1985; Nene & Jenning, 1992; Bowker et al., 1992) state that stable heart rate was achieved during crutch walking in their experiments, but do not say how long this took.

The results of the present project confirm that stable rates are achieved normally within 2 minutes in lumbar paraplegics, but in the two thoracic paraplegics (fig. 3.12), the heart rate responses were different.

The T_{11} patient stabilised at so high a rate that his maximal work capacity and endurance must both have been seriously diminished. Heart rates in the T_6 patient did not stabilise during the 5 minutes test, though they remained lower than the T_{11} patient's figures. This seemingly paradoxical difference between the lower and higher thoracic patients could have an unrecognised physical explanation, but the possibility of an essentially psychological one should perhaps not be ruled out. In this connection, it should be pointed out that the person with the T_6 injury had several years experience of walking with callipers and crutches: his T_{11} counterpart did not.

In addition, all lumbar paraplegics in this study were accustomed to crutch walking and long leg brace use. Chantraine et al. (1984) have already reported that paraplegics accustomed to crutch walking averaged speeds five times greater than unaccustomed patients, yet their cardiorespiratory responses were clearly lower.

Therefore, the different response of heart rate between lumbar and thoracic paraplegics in this study could be partly explained by less frequent use of crutches. The thoracic paraplegics presumably also suffer from reduced venous return.

Two conclusions follow. Firstly (in agreement with Eriksson, et al., 1988) the duration of crutch walking in thoracic paraplegics is limited by their low exercise capacity. Secondly, heart rate is an unsatisfactory indicator of energy expenditure in some, if not all, thoracic paraplegics.

4.5.3.3. ENERGY COST

In 1956, Gordon & Vanderwalde measured the energy cost of paraplegic locomotion. For expressing the energy cost of ambulation, they used the "Metabolic Unit" i.e., walking oxygen uptake per minute plus oxygen dept per minute, divided by basal metabolic rate. Later, it was found that a simpler figure, the oxygen consumption per metre walked, is an excellent measurement of the energy cost of crutch walking (Merkel et al. 1984). It was called "energy cost" or "orthopaedic efficiency index". Using this method, the aerobic capacity of paraplegics during exercise was measured by several researchers (Cerny et al., 1980; Chantraine et al., 1984; Merkel et al., 1984 & 1985; Nene & Patrick, 1989;

Edwards & Marsolais, 1990; Winchester et al., 1993). Table 4.2 quotes energy cost of ambulation in paraplegics, reported by these investigators.

However, in short duration or exhaustive exercises the steady rate of aerobic metabolism cannot be attained. This is the case in some thoracic paraplegics, who are not able to continue the crutch walking for a long period of time. In these conditions, oxygen consumption does not represent the metabolic rate and investigators should try to take advantage of the extra oxygen consumed by the body during recovery (oxygen debt), to estimate the total oxygen cost of exercise.

The difference between the study of Edwards & Marsolais (1990) and most of the previous studies was that they also measured the O_2 debt for calculation of the anaerobic work in paraplegic crutch walking. It must be recognised that, if the aim is to compare the different type of crutches and the duration of exercise is not enough to reach the aerobic balance, the measurement of O_2 debt could be a useful adjunct to that of aerobic cost for the selection of the most efficient walking aids.

In the results of the present project, energy cost of crutch walking was lower in lumbar than thoracic paraplegics. The results in lumbar paraplegics, who were accustomed to using the long leg brace, were comparable with those of Chantraine (0.61 ml . kg^{-1} . m^{-1}), who tested 7 accustomed paraplegics (T_9-L_1) with long leg braces during crutch walking. Energy cost of crutch walking was also close between lumbar paraplegics (0.557 ml . kg^{-1} . m^{-1}) and able-bodied subjects (0.523 ml . kg^{-1} . m^{-1}) in this project. This is not surprising; indeed Chantraine et al (1984) reported that accustomed paraplegics had an oxygen consumption which was actually lower than that of able-bodied subjects during the same crutch walking exercise. Therefore, they concluded that the energy cost of crutch walking is inversely related to the regular use of long leg braces (Box 4.1).

Box 4.1:

Effect of different factors on the energy cost of crutch walking in paraplegics

Gordon & Vanderwalde (1956) suggested that the magnitude of energy cost of crutch walking at a given speed is reflected by several factors, e.g., motor loss, age, weight of subject, spasticity and type of brace. In addition, the results of Chantraine (1984) showed that another important factor is "the regular use of the walking aid".

In the last decade considerable attention has been directed towards the development of devices that would enable paralysed people to achieve efficient movement (Winchester et al., 1993). A brief description of the new types of orthoses can be useful for understanding the effect of different types on the energy cost of crutch walking. This has been explained by Edwards & Marsolais (1990) i.e.:

i. "Reciprocal gait orthoses employ a pelvic band and trunk bracing, that allow paraplegic subjects to maintain a nearly erect walking posture; thus, reducing weight on the arms and contractions of the latissimus dorsi, compared to long leg braces. In addition, reciprocal gait orthoses have a mechanical cable system that gives a forward assist to the swing leg during ambulation".

ii.. "Parawalker orthoses do not have the cable system of reciprocal gait orthoses, but the bracing system is similar".

iii. "Hybrid systems combines parawalker orthoses with functional electrical stimulation".

Nevertheless, the only type of calipers regularly available for paraplegics are long leg braces, because the new types of orthoses are expensive (Rowley & Edwards, 1987).

As to the thoracic paraplegics, the energy costs of crutch walking found in the present study (i.e., 2.222 and 1.139 $ml \cdot kg^{-1} \cdot m^{-1}$ in T_{11} and T_6, respectively) lie within the range of literature values for simple orthoses collected in table 4.2. Clearly, crutch walking is a high energy cost activity in thoracic paraplegics. Their walking energy costs were between 2 and 5 times those of able-bodied subjects, crutch walking with the long leg brace, in spite of the fact that the physiological stress associated with axillary crutch walking is already high in able-bodied subjects (Bhambani & Clarkson, 1989).

The extra energy cost or inefficiency of crutch walking in the thoracic paraplegics using the long leg brace can probably be explained by the lack of muscular support for their lower limbs and trunks. It has been shown that the much more expensive reciprocal gait orthoses, which provide paraplegics with more lower limb and trunk support, increase the efficiency of upright movements (Edwards & Marsolais, 1990).

4.5.3.4. PHYSIOLOGICAL COST INDEX

As noted in the overall introduction, heart rate varies linearly with oxygen consumption at submaximal effort in normal subjects. Therefore, due to several advantages of the use of heart rate (i.e., low cost, can be used for long period of time, does not limit the range of activity, needs neither a well equipped laboratory nor an expert technician), MacGregor (1979) used physiological cost index (based on heart rate) for assessment of energy cost of walking in able-bodied subjects.

Furthermore, Bar-On & Nene (1990) reported a linear relationship between heart rate and oxygen consumption in paraplegics with level of lesion up to T_3. Therefore, the use of physiological cost index has recently became popular for assessment of energy expenditure in paraplegic locomotion (Bowker et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993). Some of the above workers report having found stable heart rates during crutch walking, in apparent conflict with the results of the present experiments, in that stable work-related heart rates were not achieved in thoracic paraplegics with long leg braces. The probable explanation is that the findings of stable rates have involved more sophisticated support systems (reciprocal gait orthosis or parawalker), which both (though to a different degree) probably assist venous return from the legs. The ranges of physiological cost index in the above reports were extremely wide between 1 and 5 beats . m^{-1} (table 4.2).

The physiological cost index in the lumbar paraplegics of the present study was near to that of the able-bodied subjects. The physiological cost index, as best it could be calculated, was 18.60 beats . m^{-1} in the T_{11} and 7.527 beats . m^{-1} in the T_6 subject. These higher physiological cost indices in thoracic paraplegics compared with the above studies could be explained by the following points:

i. The effect of the type of orthosis: Edwards & Marsolais (1990) showed that, in spite of a lower speed of crutch walking with the long leg brace than with other orthoses (e.g., reciprocal gait orthosis, parawalker), the energy cost during crutch walking was higher with the long leg brace in their thoracic paraplegic.

ii. The speed of crutch walking: the speeds of crutch walking in our T_{11} and T_6 subjects were 5.43 and 9.301 m . min^{-1}, respectively. They were lower than the range of speeds in the above authors (13 - 25 m . min^{-1}). In addition to the type of orthosis, the small figure of eight track could be a further explanation for this.

iii. The regular use of crutches: as noted on page 84 the regular use of the crutches is one of the main factors for decreasing the energy cost of crutch walking (Chantraine et al., 1984). Some workers cited in table 4.2 used a training program before collecting the data e.g., Edwards & Marsolais (1990) trained their subject 3 times per week for one month with the reciprocal gait orthosis; also, he had walked with a long leg brace for 5 years, previously. It will be recalled that the T_6 subject in this study used crutches at home; and his physiological cost index was less than 40% of that found for the other thoracic paraplegic, even though the latter's lesion was lower.

iv. Finally, the resting heart rate can also play an important role in changing the physiological cost index. In our experiment for all subjects (i.e., thoracic and lumbar paraplegics as well as able-bodied subjects) resting heart rates were between 60 and 77 beats . min^{-1}; the above studies reported 80 to 90 beats . min^{-1}.

The high physiological cost indices of the thoracic paraplegics in the present project can be explained by contributions from each of the above factors. Nevertheless, the problems associated with this measure, in high-lesion patients, must be continually borne in mind.

4.5.3.5. RATING OF PERCEIVED EXERTION

In any ambulatory study (with or without assistive devices), walking speed must be taken into consideration, because there is a linear relationship between energy expenditure and velocity (Hinton & Cullen, 1982). Both energy cost (ml . kg^{-1}. m^{-1}) and physiological cost index (beats . m^{-1}) are standardised, by dividing by speed; then they are used as indicators for comparison among different individuals during crutch walking. This has not been done for perceived exertion. In spite of clearly higher energy cost and physiological cost index and also lower preferred speed of crutch walking in T_6, compared with mean of those in able-bodied subjects and lumbar paraplegics, perceived exertion showed only slight differences between groups (figure 3.21). Though more high lesion patients should obviously be studied, the occurrence of even this one cause strongly suggests that perceived exertion is not a suitable indicator for comparison among subjects whose preferred walking speeds are different. For comparison between individuals, crutch walking at a given speed on the treadmill, it could have value.

4.5.3.6. SUMMARY:

This study has shown that crutch walking with knee ankle foot orthoses has a high energy cost in thoracic paraplegics. This may contribute to their reluctance to use orthoses.

In addition, the unstable heart rate, during crutch walking with the long leg brace, conflicted with the use of physiological cost index for assessment of energy expenditure in the thoracic paraplegics. If its use is restricted to those subjects who show stabilised heart rates, varying monotonically with work rate or walking speed, it will undoubtedly be more reliable than perceived exertion; but oxygen consumption (supplemented, where necessary, by oxygen debt), must remain the "gold standard" of energy-expenditure assessment, despite the practical limitations on its widespread use.

4.6 CONCLUSION

4.6.1 INTRODUCTION

The physiological responses to exercise are limited in paraplegic individuals, compared to able-bodied subjects (Glaser, 1992). This can mainly be explained by the following factors.

i. Restriction of heart rate and contractility, which limit stroke volume, may occur during exercise in high level paraplegics (because the sympathetic outflow to the heart arises above the spinal level of T_6).

ii. Immobilisation of the lower limbs (which results from muscular paralysis), can decrease venous return due to the lack of normal muscular pump.

iii. During exercise in able-bodied subjects, a redistribution of blood takes place to increase blood flow to exercising muscles. Synergistic with vasodilation of exercising muscles, sympathetic reflexes diminish blood flow to the inactive organs. In addition, they facilitate venous return from active regions by venoconstriction. These reflexes are reduced to varying degrees in most spinal cord injured individuals (Glaser, 1992).

For many patients who move by wheelchairs or walking aids, the high energy expenditure of locomotion and reduction (by the above factors) of exercise capacity combine to limit activities. Energy expenditure therefore is an important parameter in the assessment of usefulness of wheelchairs and walking aids for paraplegics. Despite widespread recognition of the problems associated with currently available mobility systems for persons with spinal cord injuries, development of more efficient systems has been slow. The lack of a suitable and simple technique for assessment of energy expenditure in paraplegics has perhaps contributed to this slowness.

In able-bodied subjects, oxygen consumption, heart rate and perceived exertion are used for assessment of energy expenditure in arm crank ergometry and wheelchair propulsion; and in crutch walking, energy cost (based on oxygen consumption) and physiological cost index serve these functions.

The task of comparing the energy cost of locomotion using the various forms of orthoses, and of monitoring the progress of training can be made much simpler by monitoring heart rate (Bar-On & Nene, 1990). However, it was evident at the outset of this project that the reliability of heart rate responses to exercise in paraplegics is doubtful as a means of monitoring the subject's physical activity, due to the physiological limitation to exercise in paraplegics. In addition, there clearly would be more circulatory problems in paraplegics in the upright position due to the influence of gravity (Davis, et al., 1993).

Thus, the aim of this project was to assess the energy expenditure during both seated and upright upper body exercise modes in paraplegic and able-bodied subjects.

4.6.2 ASSESSMENT OF ENERGY EXPENDITURE DURING DIFFERENT MODES OF UPPER BODY EXERCISE

The results of this project have shown similar physiological responses to *seated* arm crank ergometry between sports-trained paraplegics and able-bodied subjects. The lack of significant difference of heart rate between the two groups of subjects superficially conflicts with some previous publications (Hjeltnes, 1977;

Kinzer & Convertino, 1989; Hopman et al., 1992b). These researchers have reported the existence of venous pooling in paraplegics, even during seated arm crank ergometry, due to impaired venous return mechanisms and vasoregulation, which results in lower stroke volume compensated by increasing heart rate. However, the insignificant difference of heart rate between normals and paraplegics in this study can be explained by i) relatively high training level of subjects and ii) low levels of lesions of most of the paraplegics.

Crutch walking is an entirely different mode of upper body activity. In spite of there being no difference between the measured variables during seated arm crank ergometry at the same power output between the two group of subjects, the results of crutch walking have shown a higher energy cost in thoracic paraplegics than able-bodied subjects. High energy cost of crutch walking, in thoracic paraplegics, is in line with the previous studies (Clinkingbeard et al., 1964; Cerny et al., 1980; Merkel et al., 1984&1985).

Therefore, this study shows that cardiorespiratory responses to seated arm crank ergometry in paraplegics cannot be compared to the cardiorespiratory responses in crutch walking (c.f., Nene & Jenning, 1992).

One of the differences between seated arm crank ergometry and crutch walking is the different posture. This study has also shown that despite the insignificant difference of oxygen consumption between seated and upright arm crank ergometry at the same power output, increases in heart rate were significantly higher in paraplegics than in able-bodied subjects. In addition, unstable heart rate was also shown by the T_8 paraplegic. This finding firstly confirms the previous studies of Davis et al. (1990) and Hopman et al. (1992a) which reported impaired venous return mechanisms in paraplegics; and secondly, casts further doubt upon the use of heart rate based methods for assessment of energy expenditure in paraplegics (Bowker et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993).

In addition, high heart rate values and unstable heart rate in thoracic paraplegics compared with able-bodied subjects were found during the crutch walking part of this study. This observation together with the one just mentioned, contraindicates the use of a heart-rate based method for assessment of energy expenditure in thoracic paraplegics (Bowker et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993).

4.6.3 SUMMARY

1. Thoracic paraplegics show higher energy costs of crutch walking when compared to able-bodied subjects. On the other hand, there was no significant difference in oxygen consumption, heart rate and perceived exertion during seated arm crank ergometry between paraplegic and able-bodied subjects. These results can partly explain paraplegics' reluctance to use orthoses (cf., the use of wheelchair); and also explain why data on physiological responses to seated arm crank ergometry cannot be used in upright arm activities in paraplegics (c.f., Nene & Jenning, 1992).

2. The greater increase in heart rate during upright arm crank ergometry, in paraplegics compared to able-bodied subjects, can be explained by the impaired venous return mechanisms in paraplegics (Hopman et al., 1992b). This is in line with previous studies (Davis et al., 1990; Hopman et al., 1992a).

3. Physiological cost index cannot be a reliable indicator for assessment of energy expenditure during crutch walking in thoracic paraplegics due to unstable heart rate or the long duration exercise needed to establish stability (c.f., Bowker et al., 1992; Isakov et al., 1992; Nene & Jenning, 1992; Winchester et al., 1993).

4.7 FUTURE WORK

Many modified orthoses which have been made for paraplegic locomotion have never been widely used (Nene & Patrick, 1990). Reducing the energy cost of paraplegic ambulation therefore is important for rehabilitation therapists and engineers. However, a walking aid can be used by paraplegics if the amount of energy required for crutch walking, at suitable speed, is reasonably below their exercise capacities. Therefore, not only is the assessment of energy expenditure of crutch walking with the particular orthosis important, but also knowledge of the ratio of the energy requirement to the upright maximal aerobic capacity of paraplegics during upright arm activity can be valuable.

Several researchers (e.g., Eriksson et al., 1988) measured the $\dot{V}_{O_2}$max in paraplegics during seated arm crank ergometry or wheelchair propulsion, and found values markedly lower than seated normals. As we have recurrently noted, such values can partly be attributed to pooling of blood in the leg veins, which reduces the availability of blood to the active upper body musculature and limits the individual's exercise capacity (Glaser, 1985). Recently, therefore, Glaser has argued that even sitting is an inappropriate position for paraplegics.

"Since spinal cord injured individuals typically perform arm exercise in a upright sitting position, the hydrostatic pressure favours blood pooling in the leg veins. It seems plausible that the arm exercise capacity would be enhanced by placing the individual in a supine position. This would minimise pooling, facilitate venous return, elevate cardiac output and increase arm muscle blood flow" (Glaser, 1992).

These comments add strength to the contention (this thesis) that, $\dot{V}_{O_2}$max in seated and upright arm activities in paraplegics would not be the same, due to circulatory problems in the upright position.

By contrast, Hooker et al., (1993a) have recently studied the effect of posture on the exercise capacity in thoracic paraplegics (T_1 - T_5). They also compared seated and supine arm crank ergometry, and their results showed that $\dot{V}_{O_2}$max in paraplegics did not significantly increase, when arm crank ergometry was performed in the supine posture compared with seated arm crank ergometry. Nevertheless, Figoni et al (1991) had shown that the change of posture from sitting to supine arm crank ergometry can improve arm crank ergometry performance in persons with quadriplegia. They reported that both maximal power output and $\dot{V}_{O_2}$max were greater when arm crank ergometry was performed supine.

So, it is thought that in the upright position, lower stroke volume and higher heart rate could limit maximal cardiac output and $\dot{V}_{O_2}$max. This project did not however investigate the $\dot{V}_{O_2}$max in paraplegics during seated and upright arm crank ergometry, because it was not a part of the objectives; but this can be a new area of work in the future.

On the other hand, it can be suggested that electrical stimulation of the leg muscles - not to provide propulsive movement, but simply to enhance venous return - might be able to improve physiological responses to upright upper body exercise in paraplegics. Less complex orthoses, less sophisticated stimulation patterns and less muscle training would be necessary than in the functional electrical stimulation, and fatigue could be minimised by stimulating different muscles in succession. In this condition, it would also be interesting to study anew the use of heart rate based methods of assessing energy expenditure, such as physiological cost index.

In addition, Figoni (1993) has considered the following questions as directing future research in this area, i.e.,

i. can the sympathetic nervous system be stimulated in spinal cord injuries to support high intensity aerobic activity?

ii. can we develop techniques to increase peripheral resistance in inactive areas of the body during high intensity exercise in spinal cord injured people?

iii. how would lower body electrical stimulation and/or external compression, and/or selective (pharmacological or neural) sympathetic stimulation affect exercise capacity in spinal cord injury?

In particular, the possibility that there might be a degree of hypersensitivity to noradrenaline in the lower limb vasculature seems to be worth considering. Denervation sensitizes tissues up to 10^3-fold. Are the postganglionic sympathetic neurones normally interrupted (in which case the blood vessels are indeed denervated) or just isolated from higher control? If there is increased sensitivity, it will help a great deal, because noradrenaline will be too weak to affect undamaged regions by vasoconstriction of the legs.

The last word, I would like to acknowledge the people who already worked in this area and to mention that these researchers have faced problems in recruiting the paraplegic subjects. For a strong statistical analysis, as is normal in a study with able-bodied subjects, a large number of paraplegics with the same level of lesion, completeness and cause of injury would be needed. This requirement, however, is not readily achieved by the people who are working in this field.

Authors	Number of subjects	walking aid	Speeds ($m.min^{-1}$)	VO2 ($ml.kg^{-1}.min^{-1}$)	Energy cost ($ml.kg^{-1}.m^{-1}$)	HR ($beats.min^{-1}$)	RPE
Fisher & Paterson 1981	8	Under arm C.†	30.7±1	10.7±1.2	0.35±0.05	115±9	
			50.9±1.2	14.5±0.8	0.286±0.02	137.8±14	
			70.7±1.3	18.7±1.5	0.264±0.02	161.5±17	
		Fore arm C.	30.6±0.7	10.6±0.6	0.345±0.02	111.1±5.3	
			50.2±0.7	13.5±1.9	0.268±0.04	132.8±11.1	
			70±0.62	18.9±1.6	0.27±0.02	158.8±15.7	
Patterson & Fisher 1981	8		30	11	0.367	106	
Hinton & Cullen 1982	13	ACW	42.6	17	0.39	129	
Kathrins & Osullivan 1984	10	ACW	48.4±8	16.65±3.7	0.344	140±21	14±1
Waters et al. 1987	24	ACW	50±11	15.7±2.9	0.32±0.06	153±17	
Bhambani & Clarkson 1989	12	ACW‡	38.7±5.7	17±2.9	0.439	169±14	14.8±1.7
Annesley et al. 1990	10	ACW	43	2.1		140	

† Crutches
‡ Axillary Crutch Walking

TABLE 4.1
Summarised table of data presented in literature about crutch walking in able-bodied subjects

Authors	no.	Lesion	Speed ($\overline{X}\pm SD$) ($m.min^{-1}$)	Speed range ($m.min^{-1}$)	Orthosis	Energy cost ($ml.kg^{-1}.m^{-1}$)	HR ($beats.min^{-1}$)	PCI ($beats.m^{-1}$)
Clinkingbeard (1964)	19	T_4-Cauda	18.53±22	2.5-69		5.14		
Cerny (1980)	11	T_{12} - L_2	32.4±21		KAFO*	0.99±0.69	139±17	
Markel (1984)	8	C_7 - T_{10}	8.8±5.9	3.4-22.1	SC^ walker	2.92		
"	"	"	6.3±2.5	3.4-11	SSv walker	3.02		
"	"	"	17.5±6.5	5-28.8	SC Crutch	3.31		
"	"	"	15.3±3.8	9.7-19.8	SS Crutch	4.40		
Chantraine (1984)	7(UAP†)	T_9 - L_1	8.75	7-10	LLB#	1.46	131	
"	7(AP‡)	"	15	15-16	LLB	0.61	141	
"	1(UAP)	T_{12}	4	4	LLB	3.25	115	
"	3(AP)	T_1 - T_{11}	22.6	21-24	LLB	0.73	156	
Merkel (1985)	Low Thoracic	T_{10} - T_{12}	98		SC walker	3.31		
"	Mid Thoracic	T_6 - T_9	10.3			2.73		
"	High Thoracic	C_8 - T_3	7.7			7.44		
"	Low Thoracic	T_{10} - T_{12}	17.5		SC crutch	3.37		

(continued)

Authors	no.	Lesion	Speed ($\overline{X}$±SD) (m.min^{-1})	Speed range (m.min^{-1})	Orthosis	Energy cost (ml.kg^{-1}.m^{-1})	HR (beats . min^{-1})	PCI (beats . m^{-1})
Nene&Patrick (1989)	10	T_4 - T_9	12.85	7.84-21	parawalker	0.79		
Nene&Patrick (1990)	5(ES')	T_4 - T_7	14.4	10.2-16.3	parawalker	1.97		
"	(WES')	"	13.98	10.68-16		1.78		
Edwards & Marsolais (1990)	1	T_8		25.8	FES	0.93		
"	"	"		16.2	FES & RGO	1.00		
"	"	"		16.8	RGO	0.76		
"	"	"		11.4	LLB	1.12		
Bowker et al. (1992)	28		14.36±9.31	4.31-44	RGO'		142±17	5.0±3.12
"	8		15.6±8.1	6.71-30.7	HGO'		138±11	4.1±1.5
Nene&Jenning (1992)	16	T_3- T_{12}	16.94	10.06-26.58			134±20	3.11 (1.47-4.76
Isakov (1992)	1	T_4	23.9		RGO			2.55
Isakov (1992)	"	"	25.2		RGO & FES"			1.54
Winchester et al. (1993)	4	T_5 - T_{10}	12.7±1.9		RGO	1.1±0.3	145±23	3.6±0.7
"	"		13.5±2.1		IRGO"	1.0±0.1	144±20	2.6±0.5

(continued)

★ Knee Ankle Foot Orthosis
^ Scott-Craig KAFO
v Single-stopped long leg KAFO

Long Leg Brace
† Paraplegic unaccustomed to long-leg brace use
‡ Paraplegic accustomed to long-leg brace use
' Electrical Stimulattion
" Without Electrical Stimulattion
- Reciprocating Gait Orthosis
' Hip Guidance Orthosis
" Isocentric Reciprocating Gait Orthosis
" Combine reciprocating gait orthosis (RGO) with functional electrical stimulation (FES)

TABLE 4.2
Summarised table of data presented in literature about crutch walking in paraplegics

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CONFERENCE PRESENTATIONS

1. "The effect of posture on the cardio-respiratory responses in sports-trained paraplegics"

- Poster in the Conference of British Association Sports Science and Physical Medicine, Manchester, U.K., 1993.

2. "Cardiorespiratory responses in sports-trained paraplegics to low and moderate intensities of seated arm crank ergometry"

- Presented in the Physiological Society, Kings College, London, U.K., 1993.

3. "Energy cost of walking and arm cranking in paraplegics"

- Presentation in the Conference of British Association Sports Science, Aberdeen, U.K., 1994.

4. " Walking energetics in sports-trained paraplegics"

-World Congress on Medical Physics and Biomedical Engineering , Rio De Janeiro, Brazil, 1994.

5. "Energy cost of upper body activities in sports-trained paraplegics"

- Poster in the first festival of sports medicine and sports science, Glasgow, U.K., 1994.

PUBLICATIONS

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