VARIATIONS IN PEDALLING TECHNIQUE OF COMPETITIVE CYCLISTS: THE EFFECT ON BIOLOGICAL EFFICIENCY

ALFRED ZOMMERS

Dip.Tch.Tchg. (Hawthorn Institute of Ed.) 1983

Grad.Dip.Sports Sci. (Victoria College) 1988

B.Ed. (Hawthorn Institute of Ed.) 1990

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DECLARATION

I hereby certify that I am responsible for the work submitted in this thesis, that the original work is my own except as specified in the acknowledgements on page iii, and that neither the thesis or the original work contained therein has been submitted to this or any other institution for a higher degree.

[redacted]

Alfred Zommers

Victoria University of Technology

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ABSTRACT

The efficiency of pedalling technique has been a source of controversy amongst exercise scientists, coaches and cyclists for over 100 years. Biomechanically, it has been shown that orienting the pedal tangentially to the direction of motion of the crank provides the greatest force effectiveness. This is commonly referred to as "ankling". Ankling may enable greater mechanical efficiency but this, in itself, does not guarantee improved performance, because the greater mechanical efficiency may be at a cost of reduced biological efficiency. The effect of ankling on biological efficiency has not previously been investigated and is the primary focus of this thesis. Forty-one competitive cyclists were tested on their own bicycles using an electronically-braked ergometry system designed and developed for this thesis. The system was calibrated, and when compared to existing air- and mechanically-braked ergometers, was shown to be more accurate and reliable. The study prospectively randomised athletes to three months of ankling practice or to continue with their normal technique. Biological efficiency, for both ankling and normal technique, was assessed in all volunteers at the commencement of the study and at monthly intervals during the three-month training intervention. These tests were conducted at workloads approximating endurance racing performance (at an average 90% of the anaerobic threshold). Biological efficiency was calculated as power output ÷ power input the latter estimated from the energy equivalent to O^2 consumption. During the efficiency tests, kinematic data of one lower limb was recorded. Analysis of the data showed that all of the athletes were able to increase the range of motion of the ankle joint and pedal using the ankling technique. Biological net efficiency for ankling was 22.2%±2.3%. This is in accord with existing data for normal technique. Instantaneous cadence varied sinusoidally within one crank revolution. This finding was only possible because the ergometry equipment could measure mechanical power output and cadence every degree of crank rotation. The crank angle where maximum power was measured differed by just one degree between techniques but this trivial difference was statistically significant. In conclusion, a) ankling is anatomically possible, and b) biological efficiency for ankling was not significantly different to normal pedalling after accounting for differences in the set-up of individual bicycles.

PUBLICATIONS

AND

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The following publications are related to this thesis:

Zommers, A., Gibbs, M., & Selig, S. (1996). A Computerised Electrically-Braked Bicycle Ergometry System using the Cyclists' Bicycles. *Medicine and Science in Sports and Exercise Supplement to Vol 28*:S181.

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Zommers, A., Gibbs, M., Best, R., & Selig, S. (1994). Computerised measurement of the power output and cadence of competitive cyclists. *8th Australasian Human Development Conference,* Melbourne, Australia.

Zommers, A., Gibbs, M., Best, R., & Selig, S. (1994). Measurement of the power output and cadence of cyclists at each degree of crank revolution using the cyclist's own bicycle. *International Conference of Science and Medicine in Sport,* Brisbane, Australia.

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Zommers, A., Gibbs, M., Best, R., & Selig, S. (1994). Computerised bicycle ergometry using the cyclist's own bicycle. *Victoria University Faculty Conference,* Victoria University, Australia.

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Zommers, A., Gibbs, M., & Selig, S. (1995). Calibration and comparison of bicycle ergometers used in physiological testing. *Australian Conference of Science and Medicine in Sport,* Hobart, Australia.

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Zommers, A., Gibbs, M., & Selig, S. (1996) A computerised electrically-braked bicycle ergometry system using the cyclists' bicycles. *Medicine and Science in Sports and Exercise Supplement to Vol 28*:S181.

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GLOSSARY OF TERMS

Terms	Description
AC voltage	Voltage where the polarity changes sinusoidally.
Aliasing errors	Errors caused by a low data-sampling rate. Data peaks are often missed.
Alternator efficiency	Electrical power output for a mechanical power input normally expressed as a percentage.
Analogue data	Continuous values of data
Anatomically neutral	Term used to describe a joint that is in neither flexion nor extension.
Angular displacement	Displacement of an object that is moving in a circular path measured in radians. Can also be measured in degrees.
Ankling	Cycling technique prescribed as dorsiflexion of the foot at top dead centre (TDC), and plantarflexion at bottom dead centre (BDC), of the crank cycle
Armature (winding)	The coil of wire inside a generator in which a voltage is induced by a magnetic field.
Angular velocity	Velocity of continuous movement in a circular path measured in radians per second. Can also be measured in revolutions per minute (RPM) or degrees per second.

Terms	Description
Binary (language)	Mathematical language to the base 2 as opposed to decimal (base 10).
Biofeedback	Technique of using feedback of a normally automatic bodily response to a stimulus, in order to acquire voluntary control of the response.
Biological efficiency	Power output for a given energy input. Normally expressed as a percentage.
Bottom dead centre (BDC)	Pedal crank arm vertical and below the centre of the crank
Brushes (generator or alternator)	Carbon compound connections between the electrical wiring on the rotating shaft and the outside of the housing.
Cadence	Term used to describe angular velocity of the bicycle crank, measure in revolutions per minute (RPM)
Chainrings	Circular rings attached to the crank axle. Evenly spaced teeth on the outer circumference allow the power generated by the cyclist to drive the rear bicycle wheel.
Collinearity	A condition in a regression analysis where two or more independent variables are closely related. This results in the parameter estimate not being very solid.

Terms	Description
Commutator	Copper strips, insulated from each other, mounted on the generator shaft. As the shaft rotates, the alternating contact by the brushes with the commutator segments produces a constant-polarity voltage output.
Crank arm	Metal arm connected between the crank axle and the pedal axle.
Crank axle	Short axle supported by bearings. The axle is attached to the crank arm and chainrings.
Current (electric)	The flow of electrons in a conductor caused by the application of a potential difference (a voltage)
Cyclograph	Pressure measurement device
Delta (Δ or δ)	Greek letter used as a mathematical abbreviation for a change in a numerical value.
Digital	Discrete data that is represented by a binary number (0 or 1) that can be "read" by a computer.
Dorsiflexed	The ankle joint is flexed so that the angle between the foot and the lower leg is less than 90 degrees.
Dorsiflexion	The act of dorsiflexing.
DOS	Computer abbreviation for Disc Operating System. This is software that allows the user to communicate with the computer hardware.

Terms	Description
Echelon	Term used in cycling to describe cyclists riding one behind the other.
Electro-magnet	A non-permanent magnet that requires electricity to provide a magnetic field.
Electromyography (EMG)	Study of the action of muscles by measuring the electrical signals produced by muscular action.
Elliptical	Oval in shape (i.e. not perfectly circular).
Extension	Increasing the angle between two bones.
EPROM	Computer abbreviation for Erasable Programmable Read Only Memory. The stored contents of this memory can not be readily changed manipulated by the equipment end user. It is normally programmed in the factory during manufacture.
Flexion	Decreasing the angle between two bones.
Force	A push or pull that alters, or tends to alter, the state of motion of a body.
Force effectiveness	Instantaneous proportion of the force applied to the pedal that contributes to propulsion.
Freewheeling	A wheel continues to rotate with no further external input of energy.

Terms	Description
Generator (electrical)	A device that converts mechanical energy into electrical energy.
Homogeneity (of variance)	The groups should come from populations with equal variances.
Index of effectiveness	Synonym for “force effectiveness”
Inertia	Property of matter by which it continues in its existing state of rest or uniform motion in a straight line, unless that state is changed by an external force.
Kinematics	The science of mechanics that describes the nature of motion. For example, distance, velocity, and acceleration.
LED (light emitting diode)	Light emitting device, used in various electronic equipment.
Load cell	A device that measures the forces exerted on an object. Also called a strain gauge.
Magnetic flux (alternator)	Lines of force in a magnetic field.
Mechanical efficiency	Mechanical energy output ÷ mechanical energy input. Normally expressed as a percentage.
Metabolic	Pertaining to metabolism.
Metabolism	Nutritive substances being converted into energy.
Moment of Inertia	The quantity that characterises a body's resistance to changes in its angular motion.

Terms	Description
Momentum	The quantity of motion of a body and is equal to the product of the body's mass and velocity.
Normality	The scores within each variable should be normally distributed.
Ohm(s)	Unit of measure of electrical resistance to the flow of electrons.
Pedal	Platform that is attached to the cyclist's foot when riding a bicycle.
Pedalling	Action of rotating the crank arm about its axis in order to deliver power to the rear wheel of the bicycle.
Phase shift	In a sinusoidal wave the difference in time of two parts of that wave are referred to as a phase difference. This phase difference is measured in degrees with 360 degrees being one complete cycle of the wave.
Plantarflexed	The ankle joint is flexed so that the angle between the foot and the lower leg is greater than 90 degrees.
Plantarflexion	The act of plantarflexing.
Plumb-line	When a horizontal distance is required to be measured between two objects that are not on the same vertical plane, a weighted-string line is suspended from the higher object.
Power	Power is the measurement of the rate of doing work and is measured in watts.

Terms	Description
Radial force	A force in the direction of the radius of angular motion.
Range of Motion (ROM)	Maximum angular displacement occurring in the movement of a joint in the body.
Relay	Electro-magnetic operated switch.
Resistance (electrical)	Opposition to movement of electrons.
Resistive	see Resistance
RAM	Computer abbreviation for Random Access memory. This memory stores data that can later be read. The memory is volatile in that data stored is lost when power is removed from the memory.
Rolling resistance	Frictional resistance when one object tends to rotate or roll along another object.
ROM	Computer abbreviation for Read Only Memory. This memory has a program permanently stored within it and cannot be altered.
Rotor (alternator)	An electromagnet consisting of a coil of wire mounted on the shaft of the alternator.
Serial Port	A data connection in a computer that allows the data to be transferred between devices sequentially. Data transfer via this type of connection is relatively slow.

Terms	Description
Sinusoidal	Oscillatory wavelike motion within a variable. The magnitude of the variable varies cyclically above and below the mean value over time.
Spatial model	Terminology used to define the relative positions of the various parts of the human body. This is necessary in the study of human movement.
Sphericity	The variance of the population difference scores for any two conditions should be the same as the variance of the population difference scores for any two other conditions.
Strain gauge	A device that measures the forces exerted on an object. Also called a load cell.
Synchronise	To occur at the same time.
Tacho	Abbreviation for tachometer.
Tachometer	A device which can measure angular velocity (revolutions per minute)
Tangent	A line drawn at the circumference of a circle at a right angle to the radius.
Toe clips	Strap used on a bicycle pedal to ensure constant contact with the pedal.
Top dead centre (TDC)	Pedal crank arm vertical and above the centre of the crank

Terms	Description
Torque	Torque is the turning effect about a point and is equal to the product of the force and the perpendicular distance from the point to its line of action.
Tour de France	Cycling race for world class cyclists.
Universal joint	A joint on a rotating shaft that allows a certain degree of misalignment.
Ventilometer	A device that measures airflow against a fixed resistance. Also called a pneumotachograph, pmeumotachometer or anemometer.
Vernier calliper	Precision measurement tool.
Voltage	Electrical term to describe the potential difference between two points when work is done in moving a charge of electricity between them.
Wide band (radio frequency noise)	Noise generated over a wide range of frequencies.
WINDOWS	Computer operating system that interfaces between the user and most applications software.
Zero crossing detector	Electronic circuit which is capable of detecting when an alternating voltage changes polarity.

CHAPTER 1

INTRODUCTION

Although cycling is one of the most effective means of transportation (Wilson, 1973), humans are forever trying to improve on its efficiency. Much development has gone into making bicycles lighter, stronger, and more aerodynamic, with less rolling resistance and with a more efficient drive system.

As well as looking at improving upon the technology of the bicycle, research has continued to find ways to improve the effectiveness of the human body as a means of bicycle propulsion. Although human cardiovascular fitness, muscular strength and muscular endurance are among the most important aspects of the bicycle "engine", effective technique is a common linking factor. The technique directly involved in the propulsion of the bicycle is pedalling. Pedalling technique has been manipulated and investigated regularly over the past century but remains a controversial topic as researchers, coaches and athletes have been unable to agree as to what technique(s) is/are ideal.

The pedalling technique that was investigated in this thesis is commonly known as the "ankling" technique. The objective of this technique is to orient the pedal so that it is at right angles to the direction of motion of the pedal. When the crank approaches top dead centre (TDC) the cyclist dorsi-flexes the ankle in order to push through this region and then progressively plantar-flexes the foot, maintaining the pedal orientation at right angles to the direction of travel of the pedal until the pedal passes bottom dead centre (BDC). The cyclist then progressively dorsi-flexes the ankle until the crank passes TDC.

1.1 PURPOSE OF THE STUDY

This thesis investigated the biological efficiency of volunteer competitive cyclists, comparing their normal pedalling technique with the ankling technique. Oxygen consumption and mechanical power output were measured while the athletes were exercising at or close to the anaerobic threshold. In order to be able to accurately measure the mechanical power output of a cyclist, suitable equipment was required. An ergometer was designed and constructed by the author, and an electronics engineer, that would be more accurate than the commonly used air- and mechanically-braked ergometers. In order to provide the participants with a familiar environment, the ergometry system was also designed to accommodate the athlete's own bicycle. Consideration in this design was given to a system that could accommodate a wide range of bicycles. In order to be able to investigate the power and cadence within a crank cycle, which was not possible with traditional air- and mechanically-braked ergometers, this feature was included in the design to enable testing of the supporting hypotheses.

Consequently, four studies were conducted:

1. Design, construction and calibration of an electronically-braked ergometry system that would be accurate, reliable and accommodate the athletes own bicycles.

2. Calibration of air- and mechanically-braked ergometers to determine if the electronically-braked system was superior in accuracy and reliability.

3. Pilot study of adolescent triathletes to test the operation of the electronically-braked ergometry system and examine the biological efficiency of the adolescents.

4. Investigation of 41 competitive cyclists to determine if :
    a) ankling was anatomically feasible,
    b) pedalling using the ankling technique resulted in the cyclists being biologically more efficient,
    c) the crank angle at which peak power occurred during ankling was different to that for normal pedalling, and

d) cadence varied within one crank revolution.

This study required several phases:

1. Maximum oxygen uptake test ($VO_{2\ max}$) - the results from this test were used to provide a starting workload for the anaerobic threshold test (50% $VO_{2\ max}$).
2. Anaerobic threshold test - the results from this test were used to determine the workload for the subsequent tests (95% of the anaerobic threshold).
3. Four efficiency tests conducted approximately one month apart with the experimental group practising the ankling technique throughout the periods intervening the four efficiency tests. The control group underwent a sham intervention. During this phase, video analysis allowed biomechanical variables to be measured and analysed.

1.2 HYPOTHESES AND RATIONALE

1.2.1 Central Hypotheses

1. Ankling is an anatomically feasible technique.

Rationale

Although it has been reported that ankling is possible, it was important to prove this and establish that ankling did occur in the intervention by measuring the change in the range of motion of the ankle joint. This hypothesis leads into the major hypothesis that was that the use of ankling would lead to an increased metabolic efficiency.

2. Ankling technique is significantly more efficient than normal pedalling while pedalling at a workload that approximates endurance racing intensity.

Rationale

Although several studies have shown that there is a relationship between pedal orientation and effective force production, there have been no studies that have attempted to discover whether pedal orientation was related to metabolic efficiency. This is the major hypothesis in this thesis in an attempt to resolve a controversy that has existed for over 100 years.

1.2.2 Supporting Hypotheses

1. The crank angle at which peak power occurs during ankling is different to that for normal pedalling.

Rationale

If the angle is different then ankling does have an effect and this may affect the average power for the "power" cycle and therefore metabolic efficiency.

2. Cadence varies within one crank revolution.

Rationale

Bicycle ergometers employ a flywheel that stores kinetic energy and therefore any variation in angular velocity may be minimal. They also do not have a freewheel capability. This would lead to incorrect assumptions about angular velocity of the bicycle cranks while cycling on the road. The inertia of the rider and the bicycle only affects the peaks of the cadence variations and not the troughs as the cyclist has the capacity to ease up on the pedalling as desired due to the freewheel capability on a road-racing bicycle.

3. The bicycle ergometer that will be developed for this thesis is more accurate than air- and mechanically-braked ergometers.

Rationale

Although there is much evidence to show that various types of air- and mechanically-braked ergometers were not accurate measurement tools, the use of an electronically-braked ergometer would not be valid if it was shown to be less accurate, despite its advantage of being able to use the cyclists own bicycles.

1.3 ORGANISATION OF THE THESIS

The current chapter (Chapter 1) provides the rationale for the current research and a statement of the problem. Chapter 2 provides a background to the current problem. The relevant literature is reviewed in four main areas. First, the concept of mechanical efficiency, with respect to force applied to the pedals, is reviewed followed by a review of mechanical force application. This second section also includes a review of studies on biofeedback used to modify the pedalling technique. The third area includes a review of the shape of the chainrings in relation to their influence on the pedalling efficiency. Following this is a review of the various definitions of biological efficiency.

Chapter 3 describes the design, development and calibration of the electronically-braked ergometry system and the calibration of air- and mechanically-braked ergometers in order to compare the accuracy and reliability of existing ergometry systems with the present one. This chapter also includes the general methods used in the biomechanical and physiological data collection. Chapters 4, 5, 6 and 7 describe the results of the four major studies.

- Chapter 4 - Development and calibration of the electronically-braked ergometry system.
- Chapter 5 - Calibration and comparison of existing air- and mechanically-braked ergometers with the ergometry system developed for this thesis.
- Chapter 6 - Pilot study to test the operation of the electronically-braked ergometry system.
- Chapter 7 - Comparison of the biological efficiency of ankling and normal pedalling. This chapter also includes the biomechanical analysis of the angle where peak force is applied, for both techniques, and the cadence variations within one crank cycle.

Chapter 8 provides the summary of the results and the conclusions and Chapter 9 outlines the contributions of this research including suggestions for further work.

CHAPTER 2

LITERATURE REVIEW

2.1 HUMAN POWER AND BICYCLE ERGOMETRY

Muscles transform chemical energy into mechanical energy (Wilkie 1960). The chemical energy comes ultimately from the oxidation of food. Long duration energy production relies on an adequate supply of oxygen. Cycling is one form of exercise where the muscle mass of the legs is more than adequate to utilise all of the oxygen than can be absorbed.
As the human being is a limited power source, small changes can make a significant difference (Kyle & Edelman, 1974). If human power is used to propel a bicycle, not only can changes in the design of the bicycle improve the performance of the human powered vehicle but also changes in the technique that the cyclists employ. Many studies have utilised an ergometer of one form or another and those that examined their accuracy found many ergometers, especially the Monark, to be an inaccurate measurement tool (Maxwell et al, 1998; Harman, 1989; Telford et al, 1980; Wilmore et al, 1982 and Zommers et al, 1995a). There are only a few studies that have compared different ergometers. Three studies compared the Kingcycle (Palmer et al, 1996), the Velodyne (Attaway et al, 1992) and the Schwinn Air-Dyne (Milesis et al, 1991) with the

Monark ergometer. All studies found a high degree of correlation in physiological response between their ergometer and the Monark. Considering the many studies that have found the Monark not to be an accurate ergometer, especially the recent study by Maxwell et al (1998), this casts doubt on the accuracy of the Kingcycle, Velodyne and Schwinn Air-Dyne ergometers.

2.2 PEDALLING TECHNIQUE

Pedalling technique has been studied from a biomechanical and to a lesser extent from a physiological perspective for about 100 years. Efficient force application has been one of the most controversial topics with researchers, coaches and elite cyclists being unable to agree as to what pedalling technique should be adopted. This section focuses on pedalling technique from a mechanical efficiency perspective. The next section addresses the critical question of biological efficiency (power output for a given energy input) in relation to technique. Biological efficiency, rather than mechanical efficiency, is the focus of this thesis and in particular, the thesis addresses the question of whether biological efficiency can be improved using the "ankling" technique. As will be seen from the research cited below, this technique has some support from a biomechanical viewpoint. For the remainder of this section only, the term efficiency is taken to mean mechanical efficiency except where indicated otherwise. It is recognised that physiological attributes such as endurance and power are critical to cycling performance while mechanical and biological efficiency may potentially facilitate performance.

2.2.1 Mechanical Efficiency

As early as 1896, Sharp cited research, by a Mr R.P. Scott, that purported to investigate the pressure applied to the pedals of a bicycle by means of a "cyclograph". Despite being a crude form of pressure measurement by today's standards, it determined that the pressure applied was a sinusoidal function repeated for each crank revolution. Sharp acknowledged that these results gave no indication of the tangential effort on the crank. By resolving the force applied to the pedal into two components, radial and tangential (Figure 2.1), the tangential forces during the crank cycle could be calculated.

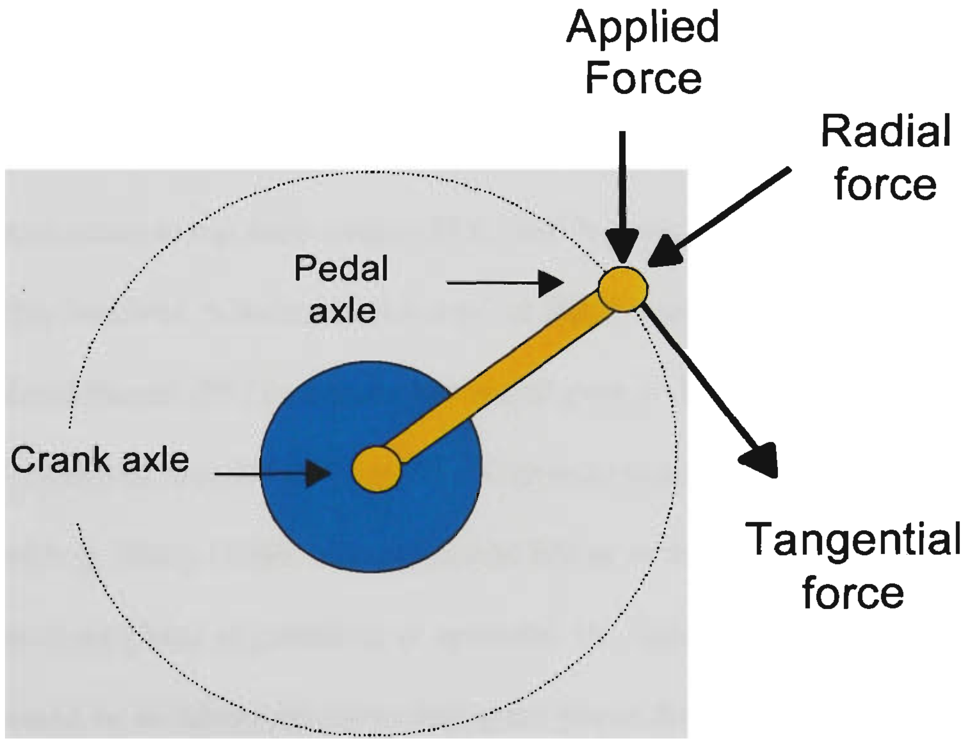


Figure 2.1 Crank effort as defined by Sharp (1896). Applied force may not necessarily be vertical, depending on the angle between pedal and crank.

By measuring only the force applied in a perpendicular direction to the pedals (Figure 2.2), and not measuring the forces in the other two axes, the true applied force could not be calculated.

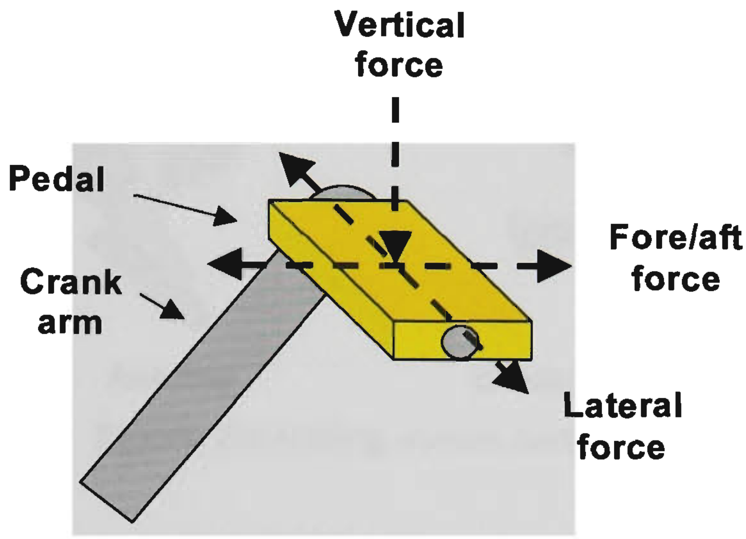
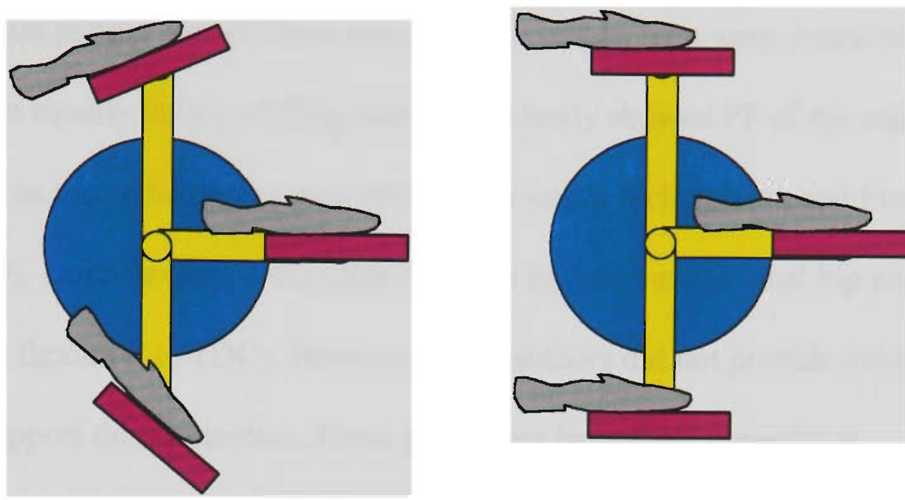


Figure 2.2 Actual forces applied to the pedals

He went on to advocate a pedalling technique to overcome the low torque production at top dead centre (TDC) and bottom dead centre (BDC). This involved "clawing backwards" at BDC, which required a plantarflexed (PF) foot and a horizontal push at TDC that required a dorsiflexed foot (DF; Figure 2.3). This technique today is referred to as ankling. Sharp (1986) also advocated lifting of the leg during the recovery phase of pedalling or upstroke. He suggested that toe clips would be an advantage by reducing the forces that opposed propulsion (Figure 2.4).



Ankling **Normal pedalling**

Figure 2.3 Ankling versus normal pedalling

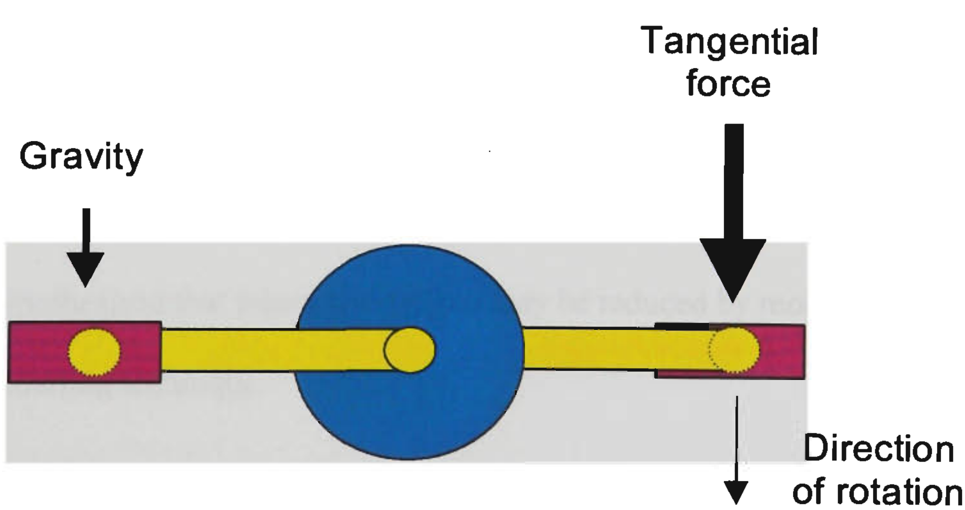


Figure 2.4 Gravitational force on the upstroke opposes propulsion on the downstroke.

2.2.1.1 Mechanical Force Application

Motion picture data of three experienced cyclists, who were instructed not to modify their pedalling technique, clearly showed PF of the ankle joint as the pedal approached BDC of the crank cycle (Houtz and Fischer, 1959). Dorsiflexion of the ankle was also reported at maximal hip and knee flexion (i.e. TDC). However, these authors did not provide evidence to support this contention. Force pedal data from five competitive cyclists, with a mean $VO_{2\,max}$ of 64 $ml.kg^{-1}.min^{-1}$, showed that they were in DF at an angle approximating TDC and were in PF at an angle approximating BDC (Gregor, 1976). Hull and Davis (1981) hypothesised that optimal tangential force could only be attained by orienting the pedals at or near perpendicular to the tangential axis. Their calculations showed that efficiency could be increased by applying PF between 90° and BDC. Toe clips on the pedals can theoretically increase efficiency by permitting greater DF movement. Hull and Davis (1981) also hypothesised that injury and fatigue may be reduced by modifying the pedalling technique.

The "index of effectiveness", often called "force effectiveness", has been used frequently to define effective force production relative to the applied force (Ericson and Nisell, 1988; Lafortune and Cavanagh, 1980 & 1983; Lafortune et al., 1983; and Lafortune, 1986). In a study investigating biomechanical factors associated with elite performance, 15 competitive cyclists were divided into two groups based on their performance in a 40 km time trial (Group 1 <56 minutes, Group 2 >56 minutes) (Coyle et al., 1991). Group 2 cyclists were more effective in their force production (effective force relative to applied force) than the elite cyclists (Group 1), although the latter were simply able to generate more force throughout the crank cycle. From this study, it could be implied that effective force production was not practised as much by elite cyclists, compared to other trained cyclists, and does not appear to be the essential component in determining performance.

Increased electromyographic (EMG) amplitude in the tibialis anterior muscle between 300 and 360 degrees suggests that this muscle was actively engaged in DF towards TDC during recovery (Hull and Jorge, 1985; and Jorge and Hull, 1984). This DF reoriented the foot and therefore the pedal so that the force could be applied more effectively during the early part of the down stroke. They presented a graph of pedal load versus crank angle; showing that the tangential force was increased between 300 and 360 degrees and therefore the cyclist was pulling the pedal through this range. This provides evidence that this DF was active in force production, rather than a passive reorientation following PF during the down stroke and the first half of the up stroke.

Analysis of EMG data in eight leg muscles, of six experienced cyclists of whom two were racers and two former racers, showed that gastrocnemius was active between 30 and 270 degrees (Jorge and Hull, 1986). In the region between 180 and 270 degrees, it is not clear if gastrocnemius acted synergistically to the hamstrings in producing knee flexion. The increased amplitude of EMG in the range of 180° to 270° does suggest that gastrocnemius was active in flexion of the knee although active PF cannot be ruled out.

Francis (1986) also hypothesised that force had to be directed tangentially to the crank in order to produce more efficient propulsion. He also proposed that extensive practice would enable the cyclist to learn more efficient pedalling patterns. Film analysis of the leg movement pattern of one Olympic champion sprinter showed DF occurring in the TDC region (300 to 360 degrees) and PF progressively between TDC and BDC. The EMG data indicates that gastrocnemius activity may have been largely responsible for the PF from 90° to 180°. The above description may be incomplete due to the lack of EMG data of other leg muscles (e.g. soleus) that also contribute to PF.

Despite there being some research supporting the concept of ankling, or documenting its usage by athletes, there is a body of evidence to show that many elite cyclists do not use ankling (Cavanagh and Nordeen, 1976;

Faria and Cavanagh, 1978; Lafortune and McLean, 1989). DF, rather than PF, was used during the downstroke by some athletes (Cavanagh and Nordeen, 1976; and Soden and Adeyefa, 1979). When power on a cycle ergometer was increased, the extent of DF at BDC increased (Black et al., 1993; and Kautz et al., 1991). Similarly, some racing cyclists used PF at TDC, in contrast to the ankling concept (Kolin and de la Rosa, 1979). Kolin and de la Rosa (1979) hypothesised that PF of the foot at BDC, as in ankling, was not a conscious effort but occurred as a passive consequence of centrifugal force which did not allow the heel to drop, especially at high RPM. However against this proposal is the EMG data showing that gastrocnemius activity increased between 90 and 180 (Jorge and Hull, 1986). Kolin and de la Rosa (1979) also hypothesised that ankling was an exaggeration of the motions used in walking, that it was passive rather than active, and thus was not applied to optimise force application.

Despite the research that showed that an ankling technique where DF takes place at TDC and PF at BDC was not going to lead to a more effective force production (Black et al., 1993; Cavanagh and Nordeen, 1976; Cavanagh and Sanderson, 1986; Faria and Cavanagh, 1978; Kautz et al., 1991; Kolin and De la Rosa, 1979; Lafortune and McLean, 1989; and Soden and Adeyefa, 1979) the Australian Cycling Federation coaching manual (Watters, 1987) advocates that a form of ankling should be adopted. The Australian Cycling Federation (Watters, 1987) recommend that ankling be used for high RPM pedalling but avoided in hill climbing.

To further confound the issue, Bernard Hinault, a multiple winner of the Tour de France had a pedalling style that was very different to other athletes at the time by dorsiflexing the ankle joint at TDC (Lieb, 1980). Anquetil, another world-class cyclist exhibited a lot of PF as the crank approached BDC (Hinault and Genzling, 1988a). Thus, both these champion cyclists exhibited some components of ankling in their techniques. However the variation in technique that has been described by many researchers (Davis and Hull, 1981; Hinault and Genzling, 1987a,b & 1988a; Jorge and Hull, 1984; Kautz et al., 1991; and Lafortune and McLean, 1989) provides empirical evidence against the universal adoption of a technique such as ankling.

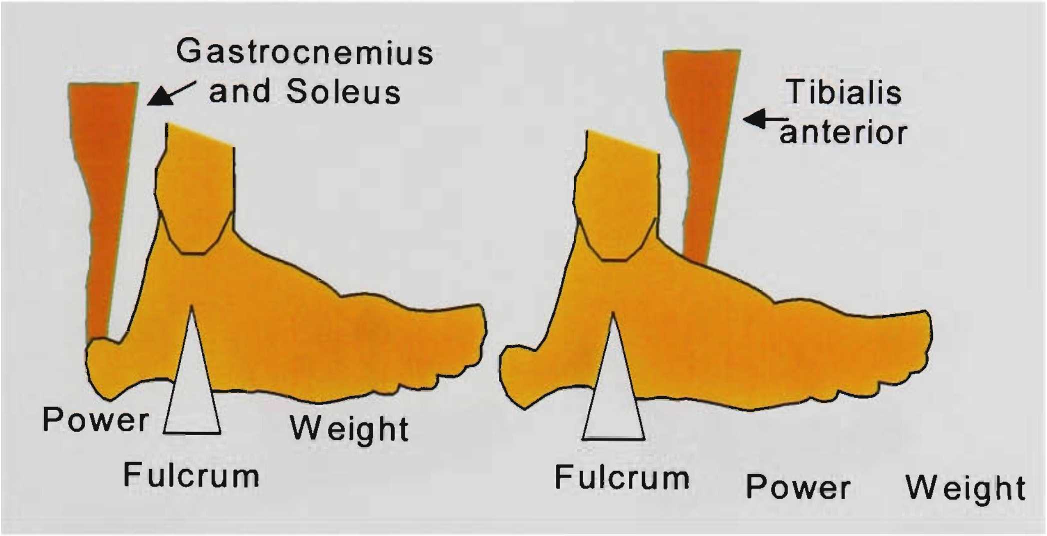
From film analysis of seven elite 4000 metre pursuit cyclists Cavanagh and Sanderson (1986) hypothesised that ankling was anatomically impossible while the rider remained seated. However when the seat was raised, the ROM of the ankle joint increased (Ericson et al., 1988). This raises the possibility that more PF may have occurred in earlier studies, had the seat been raised.

DF of the ankle joint varied between 13^{o} (Boone and Azen, 1979) and 40^{o} (Glanville and Kreezer, 1937) from the anatomically neutral position, while PF varied from 23^{o} (Sammarco et al., 1973) to 56^{o} (Boone and Azen, 1979) from neutral. This data indicates that it is theoretically possible to move the ankle through the ROM required to perform ankling. The ROM of a recreational cyclist, who was instructed to display the ankling technique, was 20^{o} DF to 45^{o} PF (Zommers, unpublished, 1987). This indicates that it is also possible to pedal using the ankling technique but whether athletes can improve their effective force output by learning the technique, such that it becomes natural, has not been satisfactorily established.

In a study to investigate the pedal loading in cycling (Davis and Hull, 1981), one volunteer was instructed to use ankling with a special emphasis on pulling up during the upstroke (i.e. BDC to TDC). The results showed that the athlete was able to increase the upward pull and shear forces on the pedals during the upstroke, however the use of ankling also resulted in a decrease in the applied forces to the pedals during the downstroke from TDC to BDC. However this study was anecdotal (n=1). Although variations in pedalling technique have been observed by many, the kinematic data alone does not reflect the changes in force applied by the limbs (Davis and Hull, 1981).

Several factors have either been overlooked or ignored in this pedalling technique controversy. As the foot moves in a circular pattern, there will be a large inertial component at the pedal, which is not measured and therefore may alter the requirements for optimising the torque production (Broker and Gregor, 1996; Kautz and Hull, 1993; and Papadopoulos, 1987). The second factor, gravity, will assist during the downstroke but from BDC to TDC will oppose the work done by the opposite leg (Broker and Gregor, 1996). Ignoring the gravity and inertial components could provide a misleading measure of the true pedalling efficiency. Removal of these components (gravity and inertia) may provide a better representation of effective and ineffective pedalling (Broker and Gregor, 1996).

The physical laws of levers are another important factor (Figure 2.5). With first and third class levers such as gastrocnemius/soleus and tibialis anterior, respectively, force production is relatively low (Thompson, 1985). This indicates that a fully PF ankle would not be able to provide much assistance to the quadriceps at BDC (Gravel et al., 1990). Against this, gastrocnemius nevertheless produces some force, which is useful. Both of these types of levers are also able to produce speed of movement and large range of motion (Thompson, 1985). Large range of motion of the ankle joint is necessary for the ankling technique and in DF the high speed of movement allows the athlete to quickly reorient the foot to a DF position prior to TDC to optimise the effective force production (Gravel et al., 1990).



First class lever **Third class lever**

Figure 2.5 Levers. First class lever (gastrocnemius and soleus) enables plantarflexion while third class lever (tibialis anterior) enables dorsiflexion.

Another factor, but possibly the most important, is the angle of pull of the muscles associated with DF and PF. The most advantageous angle of pull is when the limb segment (in this case the foot) moves in parallel to the contraction of the active muscle fibres. Usually this occurs near the mid point of the ROM of the joint (Figure 2.6 C & D). When the joint is in plantarflexion the angle of pull generates less applied force (Figure 2.6 B & F). The angle of pull for the dorsiflexors is similar to that described for the plantarflexors above (Figure 2.6 A & E)

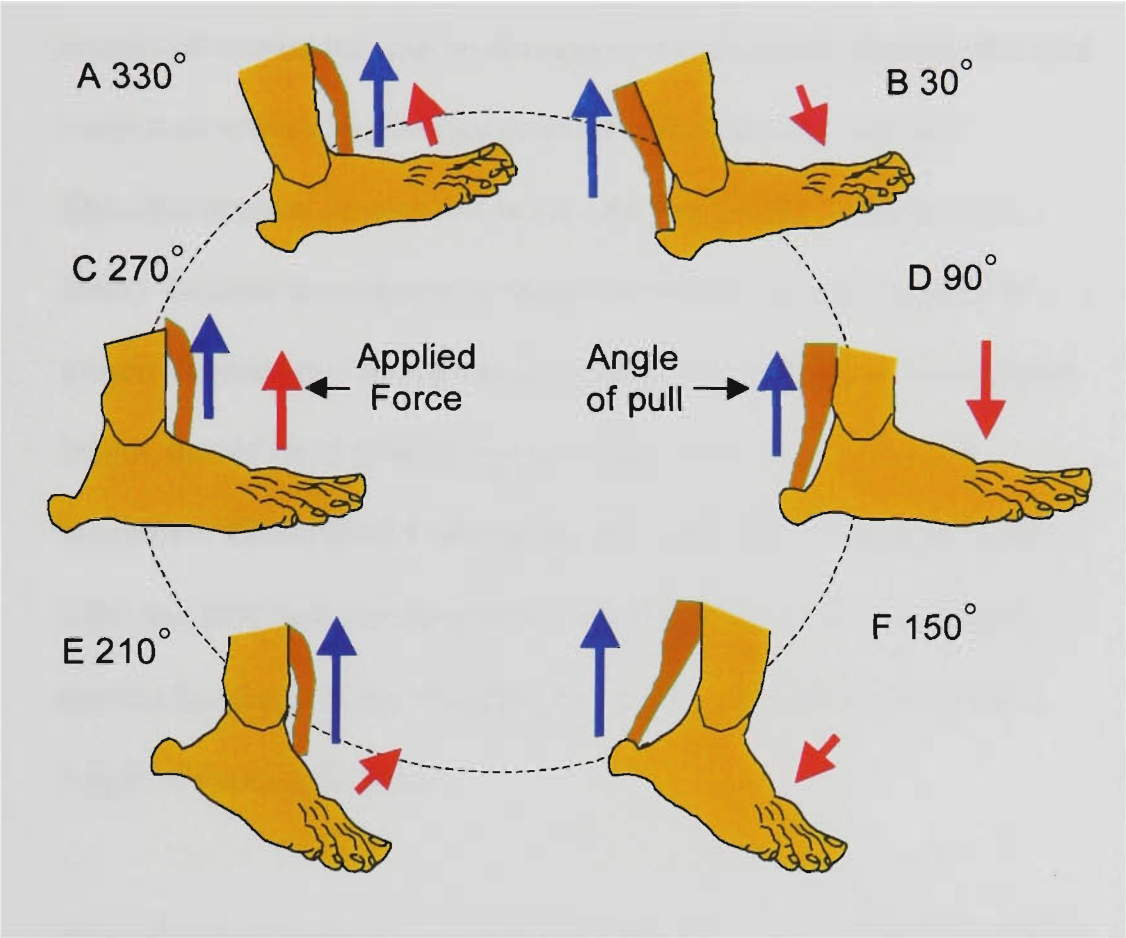


Figure 2.6 Angle of pull and resultant forces during 360 degrees of the crank cycle.

Finally, the relaxed length of a muscle that is attached to a bone is at or near its optimal length. At this length, the muscle will develop its maximum tension. As the muscle lengthens the amount of overlap between the thick and thin filaments decreases thereby decreasing the amount of tension that can be developed. As the muscle shortens the total number of active cross bridges decrease and eventually the thick filaments become compressed against the two Z lines. (Vander et al., 1985) Tension development is therefore reduced. As the length of a muscle attached to a bone rarely exceeds a 30% change from its optimal length, the ability to develop tension rarely falls below 50% of its maximum. This applies to plantarflexion during the downstroke between TDC and BDC and dorsiflexion between 300° and TDC. In each case, tension developed is not thought to be severely compromised by the length of the muscle fibres.

Muscular work depends on the length – tension, force – velocity – power relationships and the effectiveness of force production. This is affected by joint angle, muscle length and muscle movement arm length (de Groot et al, 1994). These variables are affected by pedalling rate, position and orientation of the body and changes in seat to pedal distance. When the angle of the seat tube was changed, resulting in a different horizontal seat to pedal distance, there was a change in the crank angle where maximum force was obtained. Examination of graphs of muscular force versus velocity indicate that there may be a small range of velocity where muscular force remains at a maximum (Gulch, 1994). This was also evident in graphs of power versus velocity (Chapman, 1985). Data obtained from a study conducted by Ericson et al (1986) show maximum power of the ankle joint occurred at an angle of 115 degrees (i.e. plantar flexion), which would be likely to occur after the crank had passed 90 degrees. In this same study Ericson et al (1986) found that the ankle plantar flexors made a major contribution to the total force exerted on the pedals (20%). Price and Donne (1997) found that the highest metabolic efficiency was obtained with steeper seat tube angles. They also found a decrease in the effective force from 0 to 90 degrees of crank angle but an increase between 90 and 180 degrees. One of their hypotheses to this was the alteration in the ankling pattern caused by the steeper seat tube angle.

Of the studies investigating the pedalling technique of cyclists, one important finding is that biofeedback via a computer (PC) screen can be used to modify the pedalling technique (Broker et al, 1993; McLean and Lafortune, 1988; Sanderson, 1987; and Sanderson and Cavanagh, 1990). In these studies participants were instructed to perform active recovery from BDC to TDC, thereby reducing the gravitational resistance of crank recovery. Using this, higher average torque production, over the crank cycle, was observed. However these investigators did not measure the extra metabolic cost of active recovery and hence it is not possible to make recommendations about the benefit of this technique. Is the increased efficiency outweighed by a greater metabolic cost, thus leading to reduced biological efficiency with this technique? In the present thesis, ankling rather than active recovery, is the focus of an investigation of biological efficiency.

We don't know whether the athletes who have used ankling performed the technique in response to learning or adopted the technique naturally Can cyclists who are performing another technique change their technique to ankling, with instruction, such that the movement becomes natural? In this thesis it was necessary to modify the pedalling technique to determine if biological efficiency was greater with ankling after a suitable learning period. There have been no studies published (to the author's knowledge) to date of any attempt at modifying the pedalling action between TDC and BDC.

There appears to be a range of conflicting views on the subject, based on the results of the research, some of which could be explained by the small size and the disparate nature of the experimental groups (e.g. Black et al., 1993, n=5; Cavanagh and Sanderson, 1986, n=7; Davis and Hull, 1981, n=6; Ericson and Nisell, 1988, n=6; Ericson et al., 1988, n=6; Francis, 1986, n=1; Gregor, 1976, n=5; Houtz and Fischer, 1959, n=3; Hull and

Jorge, 1985, n=3; Jorge and Hull, 1984, n=3; Jorge and Hull, 1986, n=6; Lafortune et al., 1983, n=6; Soden and Adeyefa, 1979, n=2).

Many studies did not use elite cyclists (e.g. Houtz and Fischer, 1959, experienced cyclists; Hull and Jorge, 1985, experienced cyclists; Jorge and Hull, 1986, varying from touring to elite cyclists; Lafortune and Cavanagh, 1983, college students; Sanderson, 1987, recreational cyclists; and Sanderson and Cavanagh, 1990, recreational cyclists) that could provide very different results to that obtained from elite cyclists.

Questionable statistical approaches used by some investigators have undermined the strength of their conclusions. The use of multiple comparisons (e.g. T tests) by Black et al., (1993), Coyle et al., (1991), Kautz et al., (1991), Lafortune and McLean (1989), and McLean and Lafortune (1988), gives rise to an increased Type I familywise error rate ($\alpha_{FW} = 1-(1-\alpha)^{C}$ where C represents the number of comparisons). It would have been more appropriate to use a MANOVA if it was suspected that the dependent variables were related followed by appropriate post hoc comparisons. However, in studies that used analysis of variance as a statistical tool, the small sample size would have introduced a type II error if the Type I familywise error rate was maintained at 0.05 (e.g. Ericson and Nisell, 1988 n=6; Ericson et al., 1988, n=6: Keppel,1991). Although Ericson et al. (1988) used an ANOVA for their statistical analysis, a MANOVA would have been more appropriate. They analysed the hip, knee and ankle joint motions relative to workload, pedalling rate, seat height and foot pedal position. As the range of motion of the three joints is related whilst riding a bicycle this precludes the use of the ANOVA.

Other investigators also reported significant results but they neglected to describe in full the statistical methods that they employed (Cavanagh and Nordeen, 1976; and Lafortune and Cavanagh, 1983)

The question raised from this is whether improved mechanical force application translates in to improved biological efficiency. In other words, does the athlete achieve greater power output with the same input when mechanical efficiency is improved? This thesis will not address the issue of whether increases in biological efficiency leads to increased performance. Biological efficiency is only one of many factors that contribute to increased performance.

2.2.2 Elliptical Versus Round Chainrings

Cyclists have been attempting to find the ideal pedalling technique with the belief that pedal orientation played a major part. Bicycle component manufacturers have also been looking for an improvement in mechanical efficiency by modifying the design of the bicycle system. One such modification has been the shape of the chainring. By making the chainring elliptical a lower gearing could be obtained at TDC and BDC where the force effectiveness was significantly less than otherwise ideal. Due to the lower gearing at TDC and BDC the crank could now be accelerated through these points. Although this concept was different to ankling it likewise attempted to improve performance by increasing torque at specific points of the crank revolution. This concept is supported by data from a study of 10 active males (20-30 years old), using elliptical chainrings. The researchers found a significant, albeit small, increase in biological efficiency at submaximal power but no increase in maximal power (Henderson et al., 1977). Harrison (1970) also reported that during maximal intensity short duration work, there was no increase in power output using elliptical chainrings. Nevertheless every participant subjectively preferred the use of the elliptical chainrings under high gear, low cadence conditions. The volunteers were not elite cyclists and none had participated in sport for about two years prior to being recruited for this study. A mathematical approach to the problem theorised no increase in maximal power using an elliptical chainring but an increase in maximal power of 13% could be obtained by building a split crank spindle with two chainwheels and two chains (Miller and Ross, 1980). However these authors acknowledged the limitations of their mathematical approach and recommended that studies be conducted on cycling performance using their design.

Although the elliptical chainrings may be more efficient and athletes may prefer to use them, the concept has one potential flaw. Due to the decreased gear ratio at TDC and BDC, compared to the rest of the crank revolution, the angular velocity at these points would be expected to increase. However, the product of moment of inertia and angular velocity must remain constant (conservation of momentum; Hay, 1982). The moment of inertia is increased significantly while one leg is extended at BDC. Therefore the angular velocity must be decreased to conserve the momentum of that leg. An analogy of this is the rotational speed of divers in open (e.g. pike) and closed (e.g. tuck) positions. Therefore any advantage of the leg at TDC may be counteracted by the disadvantage at BDC of the opposite leg.

Shimano Corporation created the Biopace chainring, which differed radically from the design of other elliptical chainrings. The major axis is at TDC and BDC (i.e.) about 90° to the major axis of most other elliptical chainrings (Okajima, 1983). Okajima argues that the rider matches the source to the load by utilising the gearing on the bicycle. This concept was extended to within one revolution of the crank cycle by varying the gearing by the use of an elliptical chainring. This chainring design supports the principle of conservation of momentum although this was not acknowledged by Okajima. There is no research that supports the Biopace concept apart from the data supplied by Shimano. In one study of elliptical chainrings, using only the Biopace manufactured by Shimano, it was found that efficiency did not increase in seven elite cyclists using this design (Cullen et al., 1992). Okajima tested the Biopace concept at a cadence of 56 rpm that may be common amongst recreational cyclists. This is a lower cadence than used by elite cyclists (Cavanagh and Sanderson, 1986; Lafortune et al., 1983; Marsh and Martin, 1993). The Biopace chainrings have gained popularity with the recreational cycling public, partly due to saturation marketing, but have not been accepted by many elite cyclists (anecdotal evidence). Shimano has recently acknowledged this by modifying the design to allow for higher cadences but there is no data collected on elite cyclists to advocate the use of this modified crank. Telephone communication with Shimano, Sydney Australia has confirmed that the marketing of these chainrings ceased in Australia in 1993. This may be an acknowledgement that these chainrings were of little benefit for competitive or even recreational cyclists.

2.3 BIOLOGICAL EFFICIENCY

Four definitions have been used in defining the biological efficiency of athletes (Gaesser and Brooks, 1975). These are:

1. Gross efficiency. No base line correction of oxygen consumption.
2. Net efficiency. Oxygen consumption whilst sitting on the bicycle is subtracted from the oxygen consumption during exercise.
3. Work (or unloaded) efficiency. Oxygen consumption whilst pedalling unloaded, that is zero power output, is subtracted from the oxygen consumption during exercise.
4. Delta efficiency. Changes in oxygen consumption relative to changes in work output.

The different methods of calculating efficiency make comparisons between studies difficult. Summary data from several studies, reporting cycling efficiency, is presented in Table 2.1. Faria et al. (1982) measured gross efficiency of four members of the Danish National Championship road racing cycling team. At a power output of just 140 watts, efficiency varied between 14% at a cadence of 130 and 18% at a cadence of 64. At more functional workloads, ranging between 280 and 300 watts, efficiency remained relatively constant at 22% for pedalling cadences ranging between 70 and 130. These results were consistent with those obtained in another study of gross efficiency of an Olympic Gold Medal oarsman (Banister and Jackson, 1967). In the only previously published study of endurance-trained adolescent triathletes, Zommers et al. (1995c) reported a net efficiency of 21% at a mean power output of 180 watts. Similar estimations of net efficiency were reported for a variety of trained and untrained volunteers (Table 2.1). However net efficiency tended to be lower for lower power outputs, although this observation is potentially confounded by small sample sizes and different experimental approaches.

Several investigators have commented on the relative merits of the different methods of calculating efficiency (Dickinson, 1929; Gaesser and Brooks, 1975; and Whipp and Wasserman, 1969). The assumption underlying the calculation of "work" efficiency is that no "useful" work is being performed when freewheeling. However, biological work is being performed and is proportional to the torque generated by the legs and the cadence at which freewheeling is performed. Banister and Jackson (1967) have provided data which extends on the dependent relationship between biological work and cadence showing that there is a higher oxygen concentration for higher cadence at any given power output (Figure 2.7 graphed data calculated from regression formulae). It is clear that the effect of cadence on biological work is greatest for freewheeling and diminishes, but is still evident, at higher workloads (Banister and Jackson, 1967).

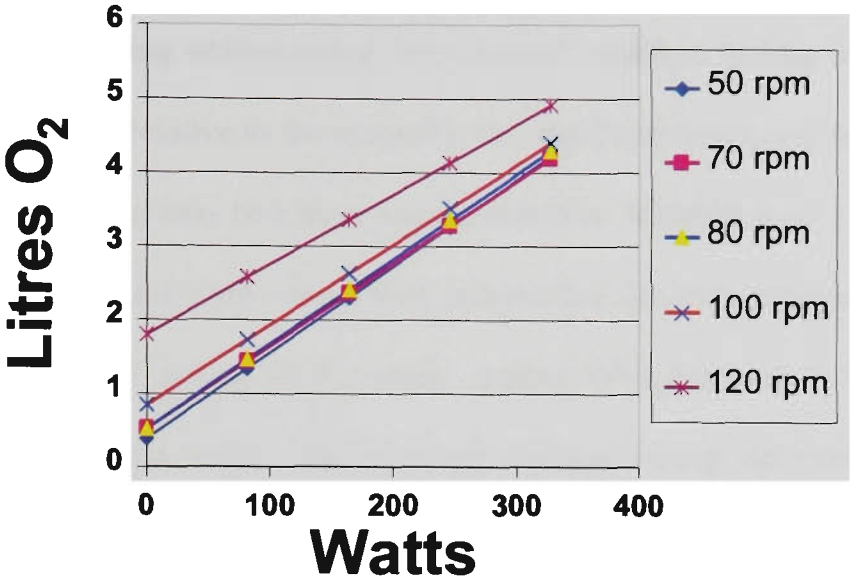


Figure 2.7 Effect of cadence on biological work is significant while freewheeling (0 watts) but diminishes at higher workloads (from Banister and Jackson, 1967).

This dependency of biological work, and therefore efficiency, on cadence, undermines the validity of both the work and delta methods of calculating efficiency and thus these methods were not adopted in this thesis. Furthermore although the use of work efficiency may be appropriate where the athletes maintain a constant cadence, the dependency on cadence introduces another variable, thereby making comparison between different experimental approaches difficult. Despite a claim that calculations of work and delta efficiency produce similar results (Garry and Wishart, 1931), this was not borne out by their own data or that of Gaesser and Brooks (1975). This leaves one possible conclusion; that there are several factors, including cadence and power, which need to be taken into consideration if work or delta efficiency is used. Testing athletes using their "natural" cadence, setting an individual workload relative to the anaerobic threshold and then using net efficiency calculations may be a more valid alternative. Wendell et al., (1980) proposed that from a theoretical perspective, the reference measurements used in calculating net (i.e. work - resting VO_2), delta (i.e. $work_1$ – $work_2$) and work (i.e. work – freewheeling) efficiencies may not remain constant from one testing condition to another. In other words, Wendell et al. (1980) argued that the validity of biological efficiency measurements is less than perfect. As all participants, in this study, worked at or just below their anaerobic threshold then it could be argued that the conditions within this study were similar. Net efficiency was adopted in this thesis.

Author(s)	n	Subject type	Notes	Gross efficiency		Net efficiency		Work efficiency		Delta efficiency	
				% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)
Dickinson (1929)	1	Unknown				11.4	178 watts (222 rpm)				
						22	280 watts (67 rpm)				
Garry and Wishart (1931)	2	Experimenters	Mean data calculated from selected samples [1,2]	12.75	89 watts (96.5 rpm)	14.95	89 watts (96.5 rpm)	33.35	89 watts (96.5 rpm)		
				17.15	111 watts (52 rpm)	20.55	111 watts (52 rpm)	25.25	111 watts (52 rpm)		
Banister and Jackson (1967)	1	Olympic Gold Medal oarsman	Mean data calculated [1]	18.2	195 watts (85 rpm)						
Whipp and Wasserman (1969)	8	Healthy male volunteers				20.2	57 watts (60 rpm)	29.8	57 watts (60 rpm)		

Table 2.1 Studies on biological efficiency of cycling (see text for further details)

Notes:

1. The author of this thesis calculated mean efficiency from the data of Garry and Wishart (1931) to simplify the table presentation.

2. Samples were selected from the published data to show relative efficiency for the same cadence. The data chosen does not necessarily show the highest and lowest efficiency for each method of calculation.

Author(s)	n	Subject type	Notes	Gross efficiency		Net efficiency		Work efficiency		Delta efficiency	
				% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)
Asmussen and Bonde-Petersen (1974)	3	1. average fitness skin diver 2. experienced middle/long distance runner 3. a fit cyclist	Study used a bicycle on a motorised treadmill					25.1	8.36 km/h (65 rpm)		
Gaesser and Brooks (1975)	12	Well conditioned males	Selected samples shown [2]	16.6	65.4 watts (60 rpm)	22.3	65.4 watts (60 rpm)	26.8	65.4 watts (60 rpm)	29	65.4 watts (60 rpm)
				20.4	130.8 watts (60 rpm)	24.1	130.8 watts (60 rpm)	26.6	130.8 watts (60 rpm)	26.3	130.8 watts (60 rpm)
Seabury et al. (1977)	3	Two trained distance runners and one active recreational cyclist		15.4	40.8 watts (42 rpm)						
				23.6	326.8 watts (64 rpm)						
Faria et al. (1982)	4	Elite cyclists		14	142 watts (130 rpm)						
			Mean data calculated [1]	22	291 watts (100 rpm)						
Zommers et al. (1995c)	15	Endurance trained adolescent triathletes	Mean data calculated [1]			21	184 watts (100 rpm)				

Table 2.1 Studies on biological efficiency of cycling (cont.)

CHAPTER 3
METHODS

3.1 RATIONALE

By measuring the mechanical power output and the oxygen consumption, of cyclists, biological efficiency for two pedalling techniques, ankling and normal could be compared. The first objective was to design and develop an accurate system to measure the power output of athletes using their own bicycle. The second objective was to compare the accuracy of existing air- and mechanically-braked ergometers with the present system. The third objective was to compare the biological efficiency of ankling with normal pedalling in competitive cyclists to determine whether ankling is a more efficient technique.

3.2 ERGOMETER DESIGN AND DEVELOPMENT

3.2.1 Evolution

3.2.1.1 Hardware

The original concept started by mounting the front of a Monark ergometer (Monark, Vanberg, Sweden) on to a force platform to measure the rotational forces of the ergometer caused by rotation of the flywheel. Preliminary testing showed sinusoidal force patterns that were smaller peak-to-peak whilst using the ankling pedalling technique. However Telford, et al., (1980) and Wilmore, et al., (1982) have shown that the Monark ergometer was not an accurate measurement tool therefore this idea was discarded in favour of developing an accurate ergometry system.

It was hypothesised that by measuring electrical power output from a generator or alternator, mechanical power could be calculated. The electrical power output could be calibrated against a known mechanical power input. Using a regression equation mechanical power could then be predicted from electrical power.

A bicycle wind-trainer (Figure 3.1) was used as a development tool.

Figure 3.1 Windtrainer used during the early development of ergometry system

The wind turbines were removed and the drive shaft connected, via a rubber tube coupling and a couple of tubing clamps, to the drive shaft of a direct current (DC) motor (24 volts, 24 watts; La 36 mini computer printer motor; A. Zommers). Being a DC motor it could also be used as a generator to produce electrical power. The generator could be attached to a resistive load and would produce energy dissipated in the form of heat in a load resistance. The energy dissipated could be measured by measuring the voltage across the load resistance and the current flowing through the resistance using a voltmeter and ammeter (Voltage * Current = Power; Weir, 1987).

This method of measuring power output was limited in that only the instantaneous electrical power at specific time intervals was obtained and this needed to be done manually. This task was automated by the substitution of an electronic load resistance interfaced to a PC using an analogue to digital (A-D) conversion card (M. Gibbs, and A. Zommers) that was controlled via a computer program (M. Gibbs). Voltage and current were converted to digital data using the same A-D card and fed to the PC for presentation on the computer screen as well as storage in a data file.

At this stage average electrical power output could be obtained for the duration of a test by sampling at fixed time intervals. However, detailed analysis between crank cycles was not possible. In order to compare the power output of individual crank cycles, an optical sensor was designed and manufactured (M. Gibbs, A. Zommers) to provide a positional reference for each crank cycle. The sensor would provide an electrical pulse output when the right crank passed top dead centre (TDC). The sensor was mounted on the wind trainer frame. Although the sensor was capable of identifying the position of one crank once per crank revolution, all other positions could only be interpolated and this interpolation was based on the assumption that angular velocity was constant throughout the revolution (Hoes, et al., 1968; Hull and Gonzalez, 1988; Hull, et al., 1991; and Jorge and Hull, 1984).

A study by the author had found that angular velocity was sinusoidal within each crank revolution in cycling (unpublished data; Zommers, 1987), with the peak angular velocity occurring when the feet were approximately horizontal with respect to each other. In this thesis it was hypothesised that power output was also sinusoidal as power is a function of force and velocity (power = Work ÷ time, Work = Force × distance, distance ÷ time = velocity therefore Power = force × velocity; Eastop & McConkey, 1969; Hope, 1975).

Because of the sinusoidal variance of angular velocity, the next major objective was to develop technology capable of measuring the time elapsed for each degree of crank revolution. In this way angular velocity and position of the crank could be calculated rather than interpolated. One degree was chosen as the increment (rather than say 10 degrees) because firstly it is easy to relate to and secondly the high frequency of data acquisition would minimise the effect of aliasing errors (Figure. 3.2).

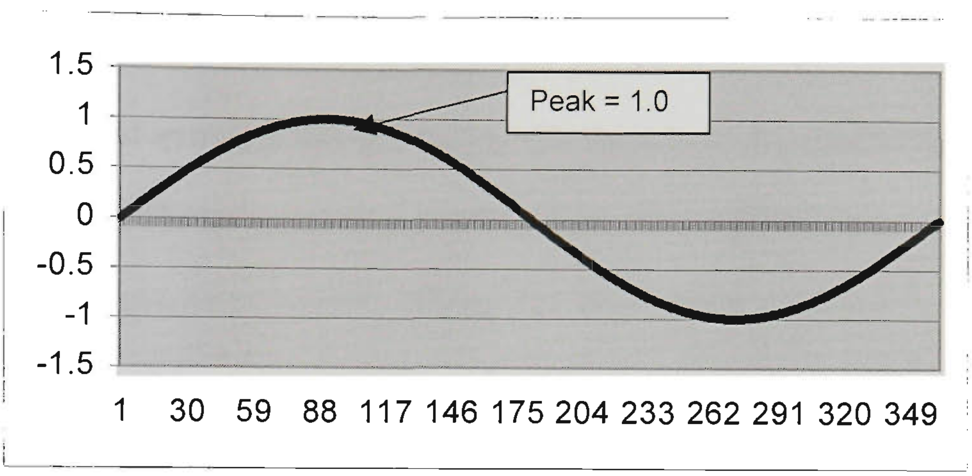


Figure 3.2(a) Aliasing effects – 360 samples

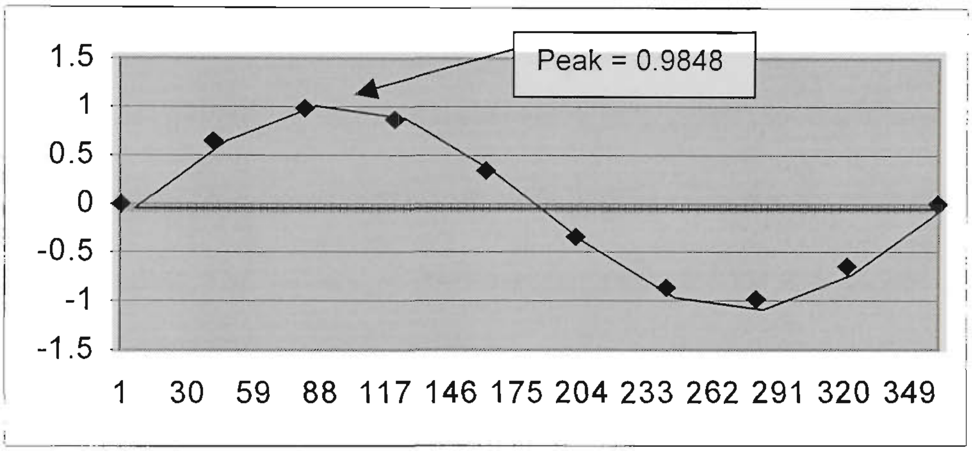


Figure 3.2(b) Aliasing effects – 10 samples

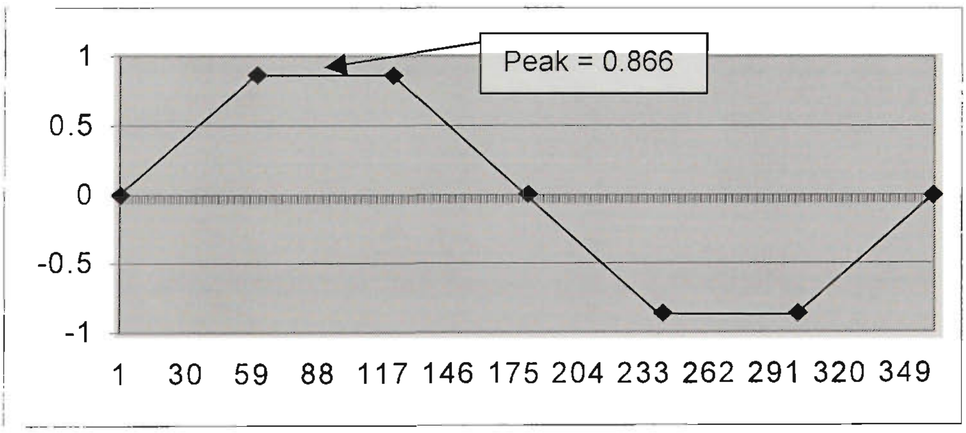


Figure 3.2(c) Aliasing effects – 7 samples

Note: Aliasing errors result in peak values not being recorded.

The generator had a tachometer (tacho) mounted on the drive shaft that could provide 500 pulses per revolution. However, the speed of rotation resulted in an output that was too small to be able to differentiate from the noise. The tacho circuit was redeveloped using optical fibre technology components (M. Gibbs). The tacho output was then multiplied by 2 to obtain 1000 pulses per tacho revolution which equated to between 28,000 and 55,000 pulses per crank revolution, depending on the gearing ratio selected (52/42 front sprocket and 12-19 rear cluster). The tacho pulse count was divided by 360 (in the software) to obtain a pulse count per degree of crank revolution. A countdown timer was then used to trigger the data sampling at each degree of crank revolution. The data collected was voltage, current and elapsed time for that degree. After preliminary testing it was found that as the bicycle tyre warmed up, the tacho pulse count per crank revolution changed. This resulted in errors in the angular position of data sampling. This was overcome by resetting the microcomputer after a warm-up (i.e. immediately before testing commenced).

The speed of the tacho was still limited due to the number of pulses that could be counted by the 16-bit computer circuitry, (65,536 pulses). This speed limitation determined the minimum diameter of the tacho drive roller.

Problems developed with interrupts in the PC. This meant that when the data was supposed to be collected, the PC may have been performing other housekeeping functions such as keyboard scan and screen refresh. Rather than try to overcome these problems, it was decided to start again and develop a microcomputer that could not be interfered with by a PC. A data acquisition circuit board, with an on board microprocessor (micro), was obtained from Nissan Motor Sport (Figure 3.3).

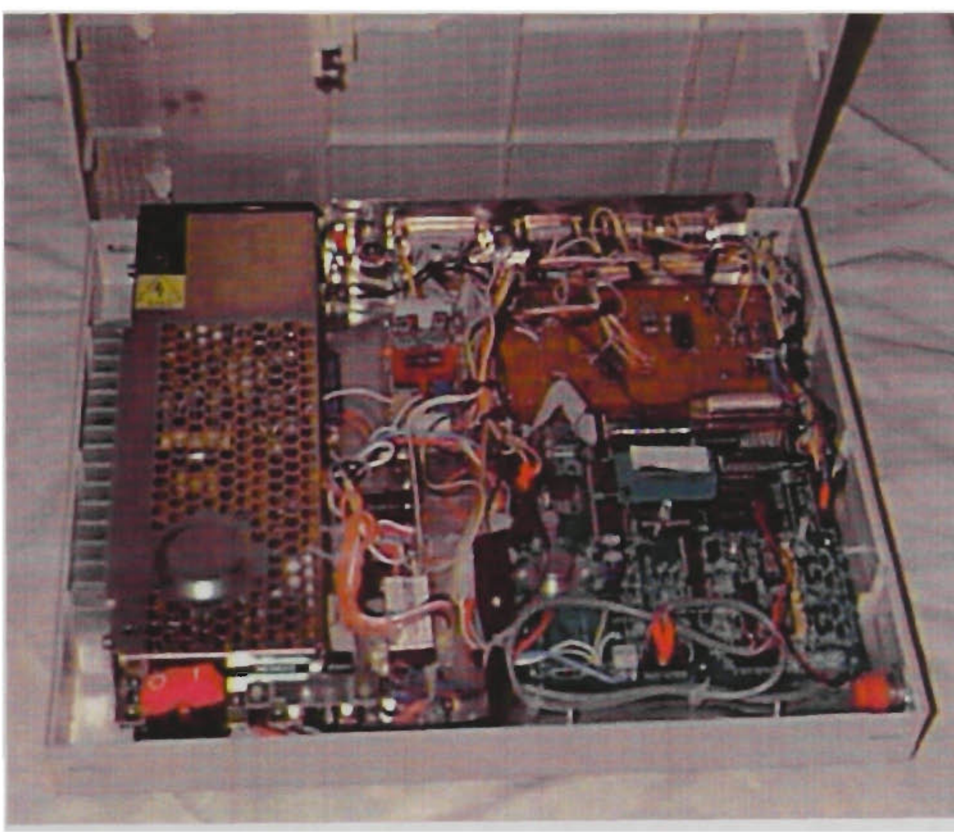


Figure 3.3 Microcomputer and associated circuitry.

The micro has an inbuilt memory (RAM) that could be used to store the program and to store data. The program to collect the data was stored in the ROM. The data was stored in RAM until requested by the PC however this meant that true real time data acquisition was not possible.

The transfer of data to the PC from the A-D board via the serial port was not fast enough for the quantity of data involved. A parallel printer interface card, which could transfer data at a faster rate, was modified (M. Gibbs) to allow two-way communications (for data one way and instructions the other way).

A rigid frame (Figure 3.4) was designed (A. Zommers) and constructed (R. Sloane) to support the bicycle.

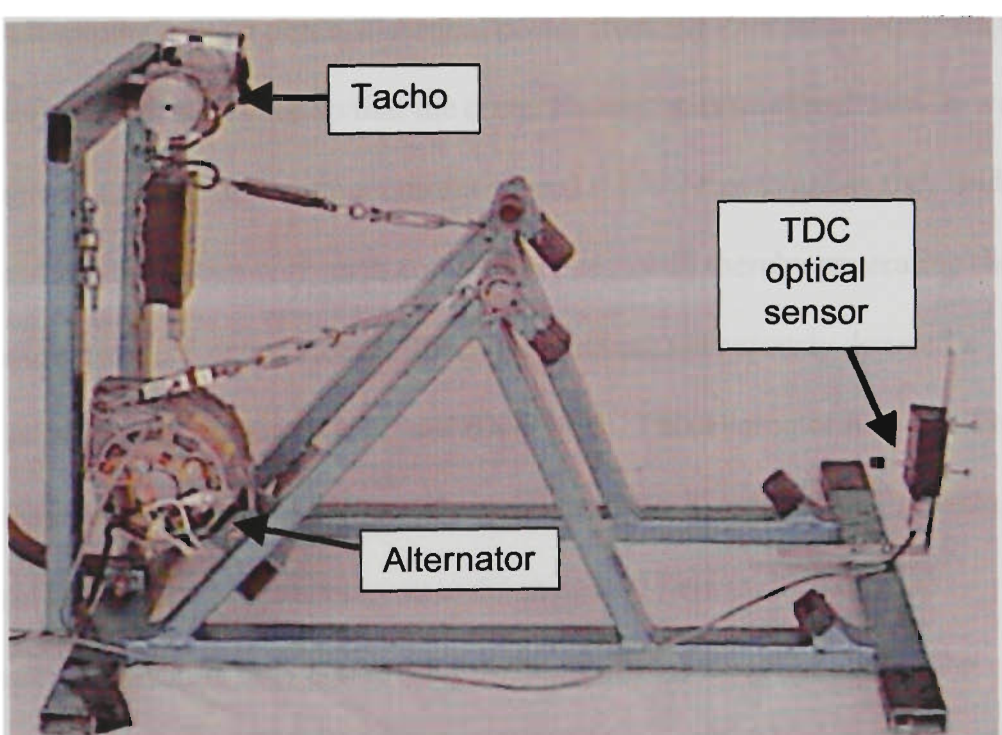


Figure 3.4 Frame constructed to support the bicycle, alternator, tacho and TDC optical sensor.

The DC generator and TDC optical sensor were attached (A. Zommers, M. Gibbs) and reliability testing was commenced. As soon as the power output was increased arcing of the generator brushes caused wide band radio frequency (RF) noise that had serious detrimental effects on data acquisition. Although many attempts at shielding this noise from the microcomputer were tried the problem could not be eliminated. The DC generator consisted of a coil of wire wrapped around an armature. Immediately the armature was rotated, the windings of the coil of wire cut the weak residual magnetic field, held by the soft iron core, generating an output voltage. This output voltage was fed back to the electro-magnets strengthening the magnetic field. This action is termed self-excitation. To obtain electrical power from the generator a split ring commutator was used so that the current output was unidirectional. It was shown that the high output current caused the brushes to arc as they broke and made contact with each commutator segment, thereby generating the wide band RF noise. Robert Bosch (Australasia) generously donated a 24-volt truck alternator (part no: BXU2455). This alternator had a coil of wire wrapped around the rotor (which was rotated) and excited by voltage from an external source to create the magnetic field (self-excitation normally occurs only above 2000 RPM in this type of alternator). The output was taken from another coil of wire surrounding the armature (the stator). The excitation of the rotating field was supplied via two slip rings and brushes. The current required to supply the magnetic field was not large enough to cause arcing of the brushes.

The pulley of the alternator was removed and replaced with an aluminium roller (A. Zommers). This would allow the bicycle rear tyre to drive the alternator. The drive roller surface was bonded with 10mm thick rubber. This would result in minimal losses of energy (especially heat) caused by sliding friction. This small loss of energy did not cause a validity problem because the ergometer was calibrated. The roller needed to be as large a diameter as possible to provide a good contact surface and therefore minimise losses due to slippage. The maximum diameter of the roller was limited due to the minimum speed requirement of the alternator (1000 revolutions per minute (RPM)). With a roller diameter of 75mm this alternator RPM was achieved at a cadence of 60 and a gearing ratio of 42/23 (the lowest gear ratio).

To provide sufficient electrical power at low RPM, the alternator required an external voltage to supply the magnetic field. A battery connected to the output of the alternator normally supplied this. As the output was connected to a load resistance this was not possible. With a suitable contact force, where sliding friction was minimal, the minimum power required to turn the crank was higher than the minimum required for physiological testing (75 Watts) at a cadence of 90 using the lowest gearing on the bicycle. The resistance to rotation was primarily caused by the magnetic flux in the rotor while self-excitation in the alternator was commencing. For self-excitation, the rotor had a residual magnetic field that is typical if the alternator had been producing an output in recent history (Garas, 1998, e-mail communication). Full self-excitation of the rotor, that is enough to produce an electrical output, would occur when the mechanical input was greater than 120 watts. This meant that a power input of less than 120 watts could not be measured. The alternator was modified (A. Zommers, M. Gibbs) so that a separate voltage supply could be used to produce the magnetic field. A fixed supply of 24 volts meant that major modifications would be required to the load resistance. A variable voltage supply was manufactured (M. Gibbs) and the control of this voltage was linked to the load resistance control so that as the load was increased so too would the field voltage increase. This meant that the alternator was able to produce a larger range of power output for a given gearing ratio and cadence.

The maximum possible speed of the alternator was well within its specifications (6000-RPM; Appendix A). This would occur with a 52/12 gear ratio (the highest gear ratio) at a cadence of 155. The alternator was fixed into position by the use of a bracket similar to that used in motor vehicles to tension the fan belts (Figure 3.5).

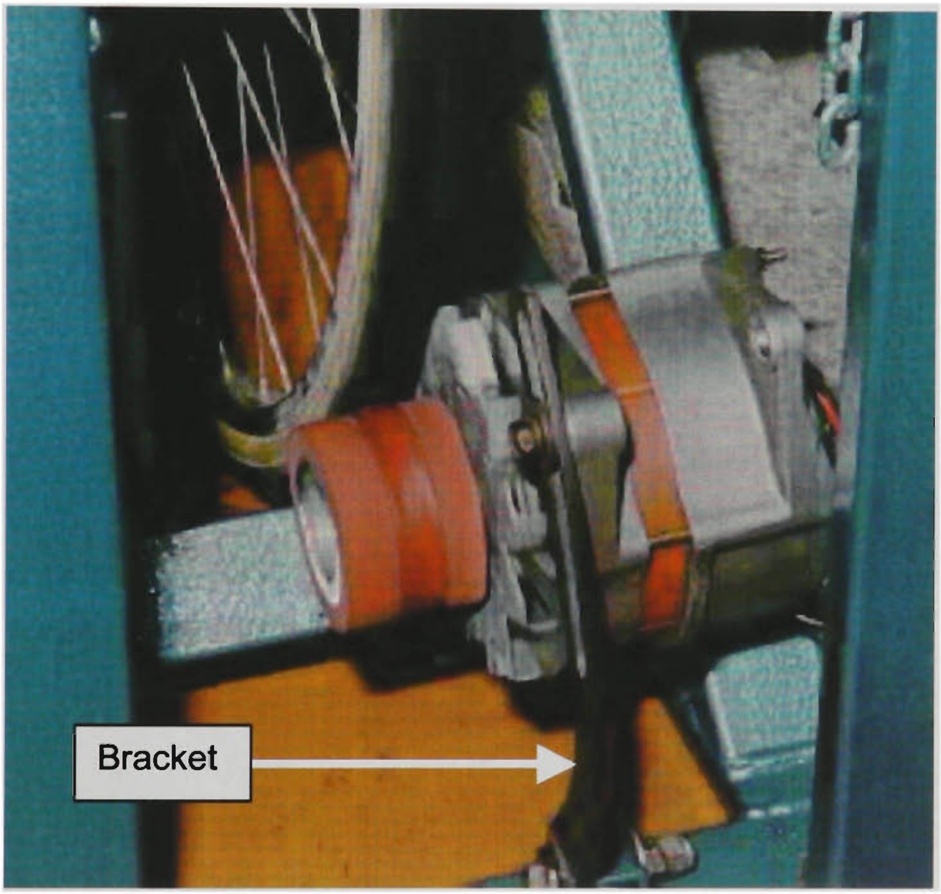


Figure 3.5 Bracket used to fix the position of the alternator.

However, this meant that the contact force between the bicycle rear wheel and the alternator drive roller could not be kept constant. A system was necessary to ensure that the contact force either remained constant or could be quantified by the microcomputer at the same time that power and cadence data were recorded. Ultimately the latter approach was adopted. A loadcell (XTRAN, Applied Measurement, Melbourne Australia specifications in Appendix B), connected via an amplifier (Strain Gauge amplifier 435-692 RS Components Ltd, Corby, Northants, UK. specifications in Appendix C), was used to measure the contact force with a turnbuckle to adjust the force (Figure 3.6). The contact force was a variable that was used in the prediction equation.

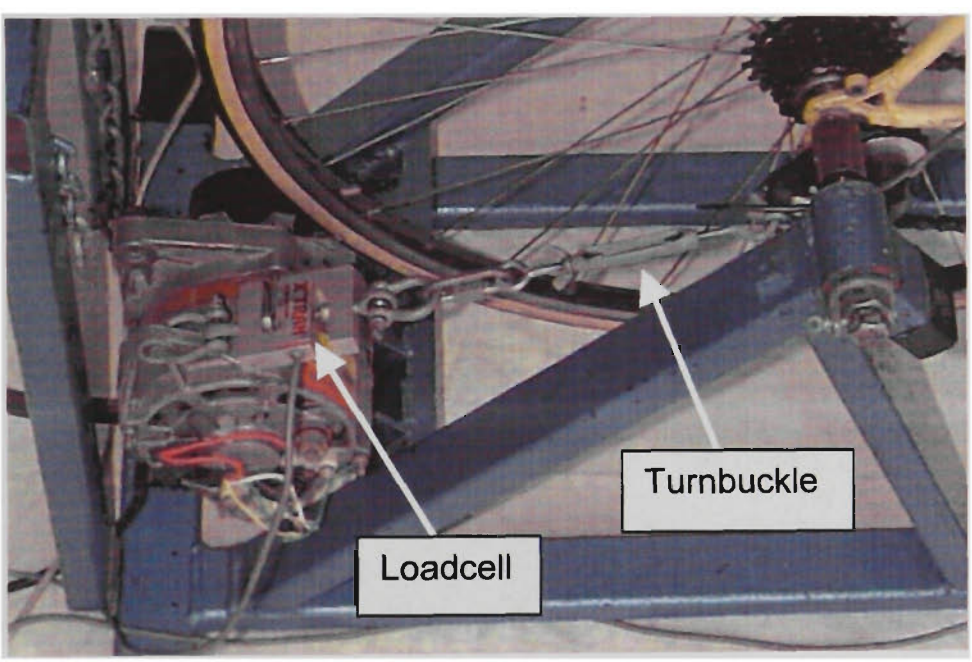


Figure 3.6 Load Cell measuring the contact force between the alternator and the rear wheel.

Testing of the ergometer revealed that a variation of 1 kg in contact force between the tyre and the tacho drive roller yielded a variation in mechanical power output of less than 5 watts. Because of the low error due to contact force, a spring balance was used to control this force (Figure 3.7). Precise measurement of this force with a load cell was not justified.

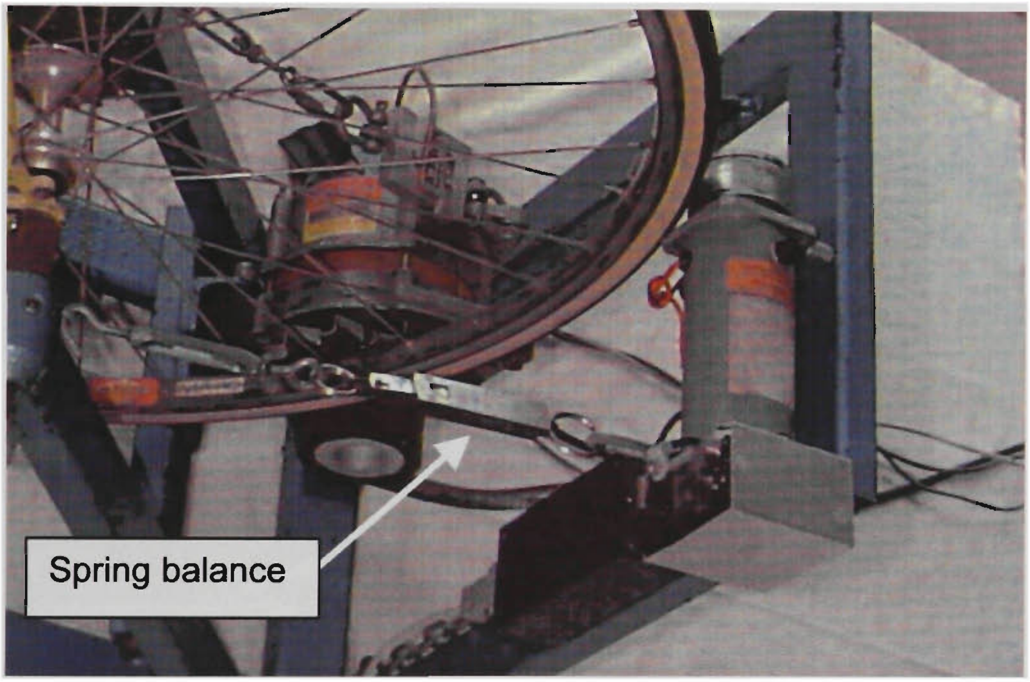


Figure 3.7 Spring balance used to set a constant contact force between the tacho and rear wheel.

Although the alternator did not generate any wide band RF noise the higher power output available caused the load resistance to fail. The load capacity was increased by 450%; cooling fans (240V AC), temperature sensing and a warning system were installed in case of overheating (M. Gibbs, A. Zommers; Figure 3.8). However when the cooling fans started, via a relay switch, wide band RF energy produced by the relay contacts closing and opening would cause the microcomputer to react unpredictably. This problem was overcome with the use of a zero crossing detector (M. Gibbs) that would allow the fans to turn on and off when the AC voltage was close to zero.

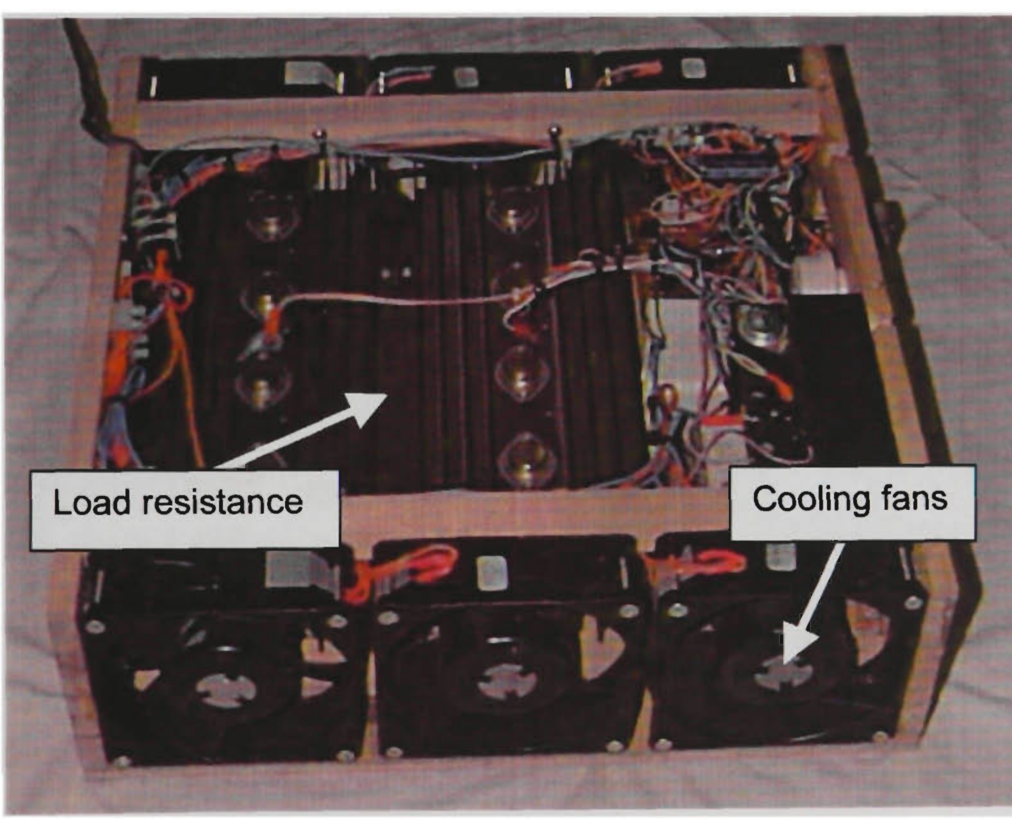


Figure 3.8 Load resistance, cooling fans and associated circuitry.

At about this time, the department's technical staff had modified a calibration-device that had been donated by REPCO (an Australian cycle manufacturer) for the project. Once calibration was commenced it was obvious that the rubber chosen for the alternator drive roller was too soft (Figure 3.9).

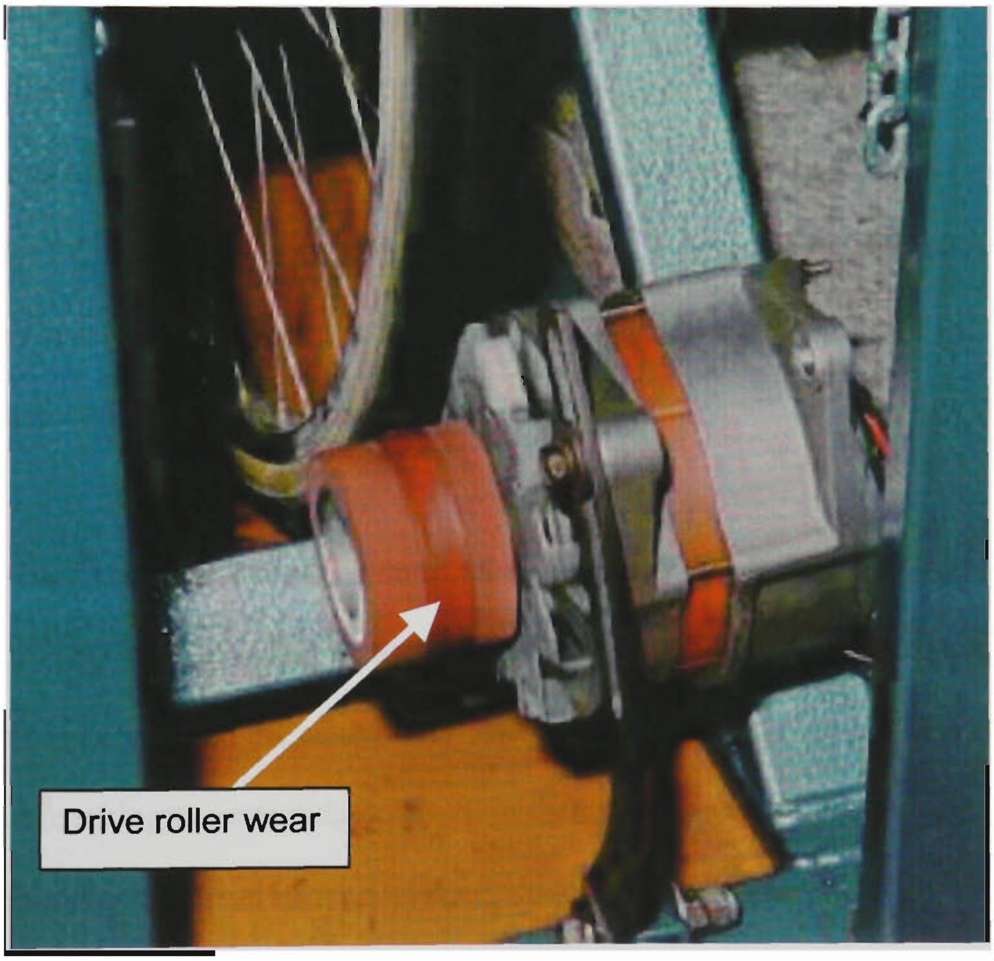


Figure 3.9 Alternator drive roller wear.

The drive roller was rebonded with a harder rubber. Even this rubber wore and therefore caused errors during later calibration. The wear of the roller reduced the diameter and caused the speed of the alternator to increase and therefore the alternator power output increased. This changed the predicted mechanical power. Although the speed of the alternator could be measured by recording its phase output, we had already used up all of the available A-D and counter/timer circuits on the microcomputer circuit board. Redesign of the circuit board at this late stage was not feasible therefore the roller diameter was measured prior to each test session with a digital vernier calliper. This data was entered into the PC and modified the regression formula used to predict mechanical power output. This was validated, using several rollers of different diameters, with the calibration-device.

Several problems surfaced during the calibration phase. First, the load cell (XTRAN, Applied Measurement, Melbourne, Australia), used to measure force on the calibration-device, was connected in line with a large spring to protect the load cell in the case of an excessive force. However when the spring stretched, the frame holding the motor "bottomed out", and the load cell would only read 8.5 kg although the force was greater. The spring was replaced by a solid coupling link. The next problem was the design of the voltage supply within the microcomputer. The two load cells, together with their respective amplifiers (RS Components, Corby, Northants, UK.), drew a current of 100 mA. As the load cells and amplifiers had been added after the voltage supply had been manufactured they caused an excessive load on the voltage supply therefore resulting in data that varied by 87 grams (0.7%) during testing. This data was rejected and the voltage supply in the microcomputer modified (M. Gibbs) to cope with the increased current demand. After modifying the voltage supply, the loadcells were checked and it was found that there was a 0.2% error in one load cell system. After many hours of fault finding, a simple typographical error was detected in the software. This was corrected and the system was now ready for the final test and calibration phase.

Prior to the equipment being able to be used in the study, it was necessary to develop a method of synchronising the ergometer data with the video recordings that were used to record the descriptive kinematic data. A strip of 10 lights was manufactured (A. Zommers, M. Gibbs) that would be visible in the field of view of the video camera and would indicate which revolution was being recorded. This was appropriate at the time as the microcomputer equipment could only record approximately 10 crank revolutions of data due to memory limitations. Later developments in the data recording system enabled 2-minutes of data to be recorded 4-times in each testing session and therefore a better system was needed to synchronise the data recording with the video. Auto digitising was to be used with the video equipment but this feature required a start and a stop signal to indicate the series of frames to use for digitising. It was decided to use a light emitting diode (LED) that would be visible within the field of view of the camera that the digitising system would recognise as a start/stop signal.

3.2.1.2 Software

Firmware was developed for the EPROM to collect data, and software for the PC to take the data from the micro-computer and not only store it in the hard disk drive but also present it on the computer screen in a suitable format. The first software package developed was DOS-based however it was limited in presentation. The bar graphics were chunky and distracting. A DOS-based graphing program was also developed at this time and used for analysis of the data during development of the ergometry system. The main program (named WINLOG) was rewritten using the WINDOWS presentation (M. Gibbs). WINDOWS software having a "mind of it's own" meant that it was necessary to prevent WINDOWS taking control for housekeeping functions unpredictably. Software was included to allow the data-logging program to have 0.5 seconds of time each second, the remainder available for WINDOWS. Within the program, it was possible to display the graph of a single crank revolution on the screen or the accumulated data. However the display of a single revolution, which was as close as possible to real time data, was not possible until a faster PC was obtained.

The system featured three principal modes of operation for testing athletes. The first, the "Normal" mode, measured power output and cadence at each degree of crank revolution. Using this mode, the researcher (and ultimately the coach) could determine the angle of crank revolution where the cyclists exhibited maximum power and identify any pedalling irregularities within the crank cycle. The second, the "Physiology" mode, featured an incremental step protocol for $VO_{2\,max}$ and anaerobic threshold testing. Starting power, step size and step time intervals were controlled by the PC. Step size could be set as low as 5 watts. This mode of testing provided the mean power output and cadence of each crank revolution. The third mode was developed to meet the requirements of the pilot study to validate the hardware and software. A predetermined cycling distance was entered into the PC and mean data over 15 second intervals was provided for the duration of the test. Collected data was displayed on a computer screen for the athlete, and researcher as well as being stored in a data file for analysis. Data was also available in graphical form for real-time analysis, if required. Apart from the above testing modes, a calibration mode was added to record data and control the load resistance during the calibration of the system. A phase shift mode was also added to record the phase shift between power input and power output. This necessitated a separate EPROM chip being programmed to change the load resistance from minimum to maximum automatically.

The changes in power input and output were recorded during this test and compared on the same time scale (Figure 3.10).

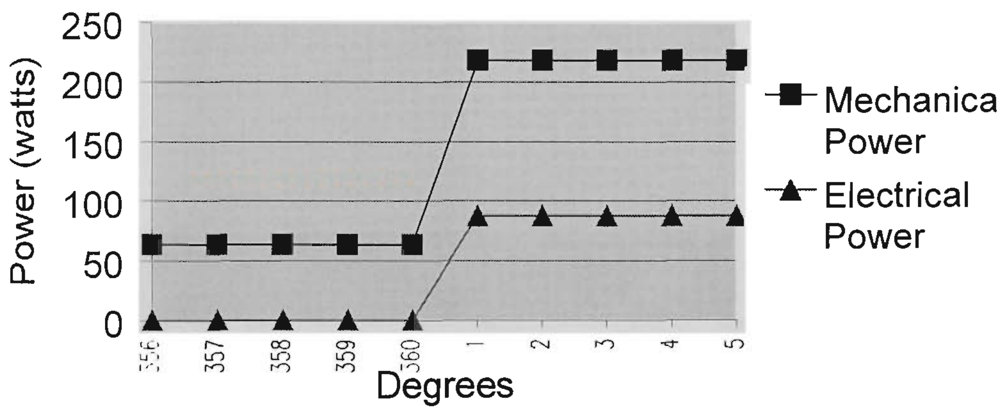


Figure 3.10 Phase shift between electrical and mechanical power is less than 1 degree.

3.2.2 Final Prototype

At this point of development, no further modifications were needed or undertaken. The final prototype was used in the validity, calibration and main studies in this thesis.

The back wheel of the bicycle drove the alternator and tacho via drive rollers (Figure 3.11). The alternator produced electrical power that could be mathematically converted to mechanical power that was produced by the cyclist.

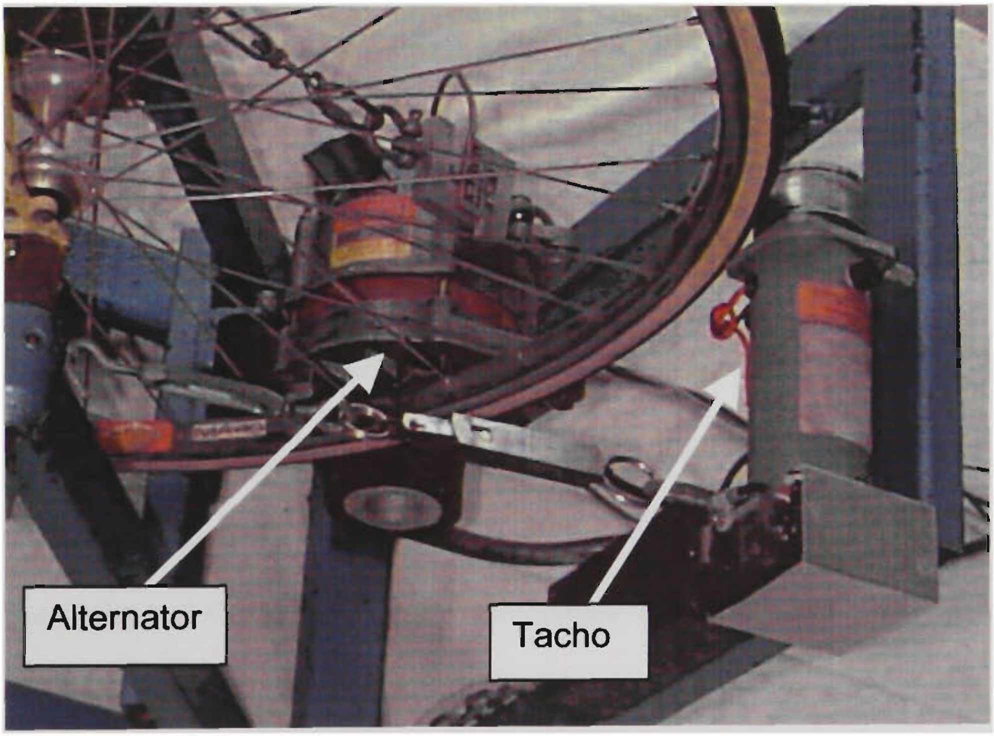


Figure 3.11 Alternator and tacho driven by the rear wheel.

The tacho produced approximately 38,000 pulses per crank revolution. The time of one crank revolution was obtained using an optical sensor to detect TDC. The tacho pulse count was divided by 360 to obtain a pulse count per degree of crank revolution (i.e. approximately 105 pulses). The microcomputer then sampled the power output from the alternator every 105th tacho pulse. As the angular displacement was one degree then this provided the angular velocity that was then converted mathematically to cadence (RPM). For more technical information refer to Appendix D

3.2.3 Development of Data Management Software

Due to the large size of a 2-minute data file (320-420Kb in binary language), available statistical software was not able to analyse the data. Software (named WINSTAT) was developed (M. Gibbs, A. Zommers) and a small sample file was used to compare the results with those produced by Microsoft Excel (Microsoft Corporation, USA) and SPSS (SPSS Inc, Chicago, USA). WINSTAT also provided data smoothing using Savitzky and Golay's least squares method with an interval of nine (Savitzky and Golay, 1964). This smoothing minimised any phase shift of the data but removed the "noise". During development smoothed data was overlaid with unsmoothed data to ensure that no perceptible phase shift occurred and no important data was lost.

3.2.4 Ergometer Validation

Although the ergometer was later calibrated (Section 3.3), the electrical output of the ergometry system was measured to ensure stability, linearity and accuracy of the voltage and current output. A known measured voltage was used to simulate the validation. The voltage was measured using an 8 digit Hewlett Packard voltmeter (Hewlett-Packard Company, Palo Alto, C.A. USA) and current was calculated by measuring the voltage across a precision resistance of known measured value. The internal clock circuitry of the micro was compared with the time measurement of one crank cycle. This was measured using an 8-bit frequency counter (manufactured by Mark Gibbs). The frequency counter had been calibrated using a Fluke frequency meter (Fluke Corporation, Everett, Washington, USA).

3.3 ERGOMETER CALIBRATION

Four ergometers were calibrated using the calibration-device donated by REPCO. (Hooper, 1977; Zommers, et al., 1995a,b) These ergometers were:

- the prototype developed for this thesis (electronically-braked system; A. Zommers, M. Gibbs),
- a mechanically-braked ergometer (Monark),
- an air-braked ergometer manufactured locally for the Victorian Institute of Sport, and
- a commercially available air-braked ergometer (REPCO) modified to include derailleur gearing.

The objective of the calibrations was to compare the accuracy of the ergometers (Section 5.3.2). Calibration on each of the four ergometers was repeated at least once and intra variability was tested using conventional statistical methods.

3.3.1 Calibration-Device

The calibration-device (Figure 3.12; Hooper, 1977) is a dynamometer which consists of a variable speed 1.25 horse power DC electric motor mounted on a frame with gearing to effect a speed reduction of 22.4:1.

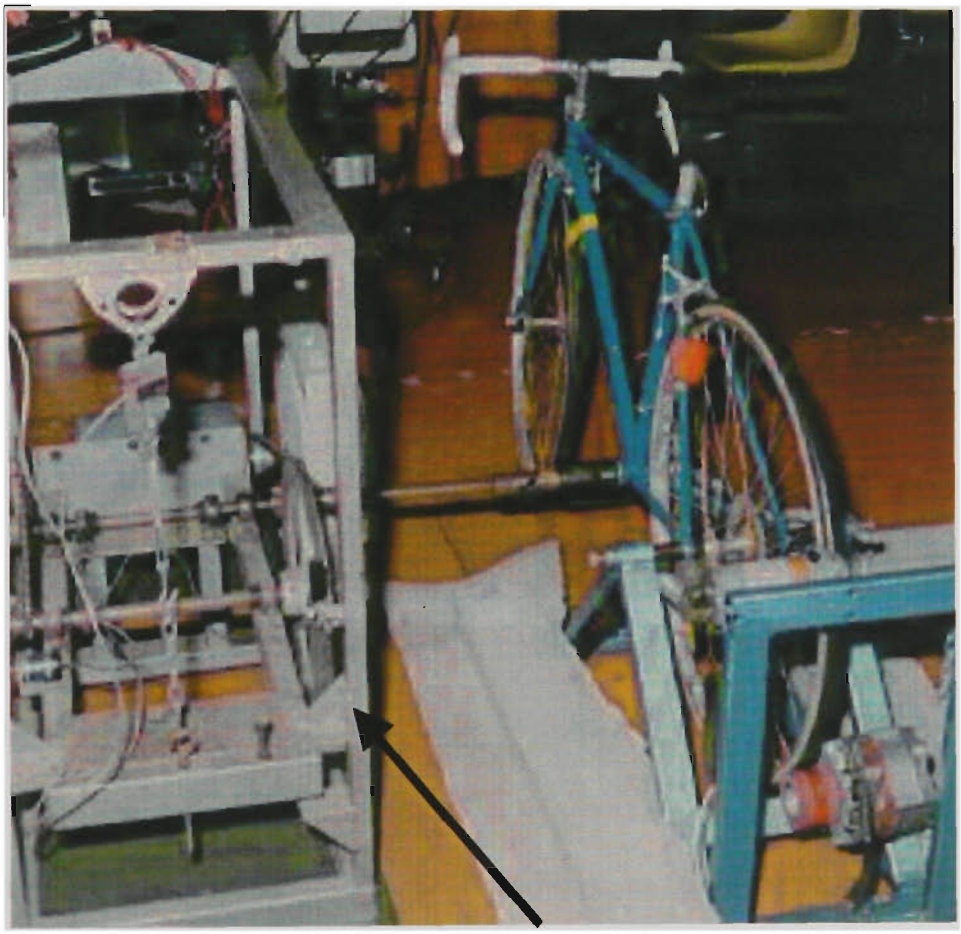


Figure 3.12 Repco calibration-device attached to the ergometry system

The frame that is suspended via bearings on the shaft connects to the crank of the ergometer under test. A strain gauge was connected 63.66 cm from the output shaft, which equates to a circumference of 4 metres. The variables of distance and force together with velocity, measured independently using a calibrated frequency counter, were used to calculate the mechanical power input to the ergometer under test. To connect the dynamometer to the ergometer, a double universal joint coupling shaft was used in an attempt to remove unwanted variability in the strain gauge output.

3.3.2 Calibration of the Prototype (Electronically-Braked Ergometer)

The ergometer was calibrated, using the calibration-device, for several gear ratios from 52/13 to 42/23. The calibration was carried out using the independent variables of electrical power output from the alternator, alternator rpm to account for the non linear output of the alternator (Appendix A), alternator contact force to account for the power lost in friction between the alternator and the bicycle wheel and load resistance. These variables were used to predict the mechanical power produced by the cyclist. Electrical power = V*I and V ÷R = I (Weir, 1987). The close relationship between load resistance and power could violate one of the assumptions of a multiple regression (collinearly). Control of the load resistance settings were directly linked to control of the alternator field voltage, therefore justifying the inclusion of this variable. The calibration was carried out over the expected operating range of each variable. Angular velocity measurement was carried out using an accurate frequency counter and force measurement was achieved with a load cell attached to a load cell amplifier and the data fed to the microcomputer. The calibration-device used the primary measurements of force, angular distance and angular velocity to measure power input to the ergometer under test.

Data was collected while the test bicycle gearing was varied. Eight gearing combinations were used. The cadence was varied from 60 rpm to 120 rpm in 10-rpm increments. Alternator contact force was varied from 5 Kg to 7 Kg in 1-Kg increments. The load resistance was varied in 20 steps from 2 to 544 ohms.

Data was collected and fitted to a multiple regression. A cubic polynomial gave the best fit although some violations of assumptions were observed to be marginal. On advice from the University statistician, a quadratic polynomial was used in preference. The variables of electrical power, load resistance, cadence and load cell contact force were the independent variables used in the final regression equation to predict mechanical power.

The wear of the alternator drive roller (see Section 3.2.1.1 above) caused small errors that were not detected until the equipment was recalibrated to ensure that measurement would remain constant over time. Instead of measuring the velocity of the alternator directly from the “phase” output the drive roller diameter was measured prior to each testing session with an electronic vernier calliper. This measurement was entered into the software via Winlog.ini (a form of lookup table; Appendix E), a file that modified the regression prediction formula during data collection.

3.3.3 Calibration of Mechanically-Braked Ergometer

A Monark mechanically-braked bicycle ergometer (Model 864; Monark, Varberg, Sweden) was calibrated using the calibration-device (Hooper, 1977). The Monark ergometer had been modified to include an electrical strain gauge in line with the friction cord. This method theoretically took into account any losses as a result of changes in the frictional force caused by heat. Two sets of data were collected from this ergometer; one set of data using the traditional method of calibration, as developed by Monark using a weight cradle, and the second set of data from the modified force measurement system.

3.3.4 Calibration of Air-Braked Ergometers

A REPCO air-braked ergometer, modified to include derailleur gearing and an air-braked ergometer used by the Victorian Institute of Sport (VIS), similar in design and function to the modified Repco; were calibrated using the calibration-device (Hooper, 1977). Theoretically cooling fans, walls, other objects and even people can affect the airflow of the ergometer turbine. This will alter the measured power output. Although not the subject of this thesis, some qualitative measures of the influences of these objects were made. Subsequently the calibrations on these air-braked ergometers were performed with the ergometers being positioned well clear of any objects or walls.

3.4 PARTICIPANTS FOR THE MAIN STUDY

Forty-five competitive cyclists were recruited after they were informed of the details of the study, the possible risks involved and had signed a letter of consent (Appendix F). Volunteers over the age of 35 were required to obtain approval to participate in the study from their medical practitioner. The Human Research Ethics Committee of Victoria University approved all experimental protocols. Physical characteristics of the athletes (age, body mass, height, $VO_{2\ max}$; mean ± standard deviation (SD), max, and min) are presented in Table 3.1 below. Cycling ability ranged from club level to elite (Olympic standard) with the cycling disciplines of road, track, triathlon and mountain bike being represented.

		Age	Mass kg	Height cm	$VO_{2\,max}$ ml.kg^{-1}.min^{-1}
Males n=31	Mean	32.5	72.6	177.9	62.7
	±SD	11.5	7.5	6.7	6.9
	Max	52	87.5	191.5	80.0
	Min	16	58.4	164.5	50.0
Females n=14	Mean	28.5	60.8	166.6	53.6
	±SD	7.8	6.0	5.3	7.2
	Max	49	73.4	179.0	70.7
	Min	16	50.9	159.0	44.3

Table 3.1 Physical characteristics of the athletes

3.5 EXPERIMENTAL DESIGN

On the first visit to the laboratory cyclists underwent a test to determine resting VO_2. During this procedure, participants sat quietly on their own bicycles and the lowest VO_2 during a 10-minute continuous recording was used. Following this, each athlete underwent a $VO_{2\,max}$ test. On the next visit, anaerobic threshold was determined by estimating the second breakpoint in plasma lactate during a graded exercise test starting at 50% $VO_{2\,max}$ with 10 $watt.min^{-1}$ increments until volitional exhaustion. In the case that blood samples were not obtained anaerobic threshold was determined as the ventilatory breakpoint in a separate test. The anaerobic threshold was used to set the training intensity for all volunteers. Immediately following the anaerobic threshold test the athletes were randomised to the experimental or control groups (Table 3.2).

GROUP	Pre Training				Ankling Practice	Post$_1$				Ankling Practice	Post$_2$				Ankling Practice	Post$_3$			
	N	A	N	A		N	A	N	A		N	A	N	A		N	A	N	A
Experimental n = 21	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓
Control n = 20	✓	✓	✓	✓		✓	✓	✓	✓		✓	✓	✓	✓		✓	✓	✓	✓

Table 3.2 Research Design

Notes:

1. In each group half of the participants were randomised to the efficiency test using Normal – Ankling – Normal – Ankling (N-A-N-A), while the remaining half had the opposite order (A-N-A-N; not shown above).

2. The pre training test and post training tests (Post$_1$, Post$_2$ and Post$_3$) were done one month apart, with training in between. Cyclists who were randomised to the experimental group were required to practise ankling during each training period whereas the control cyclists continued with their normal training (see text for further details).

A stratified randomisation was used to maintain equal numbers of males and females in each group and equal numbers of volunteers under and over 35 years of age. The cyclists were then tested on four occasions one month apart. Data collected during these tests were oxygen consumption, mechanical power output and cadence. The cyclists were also video taped for subsequent analysis of their pedalling technique. Cyclists were required to pedal using normal or ankling technique for the first five minutes and then change to the other technique. In total athletes completed four by five minutes, alternating between the two techniques, on each of the four laboratory tests. Prior to each data collection participants from both groups were instructed on the requirements of the ankling technique. In the case of the experimental group these instructions served to reinforce the technique of ankling that they were required to practise in training. The first three minutes of each five-minute phase enabled VO_2 to plateau and allowed the pedalling technique to be refined (Casaburi, et al., 1987; Hagberg, et al., 1978). Mechanical power output and cadence data was collected during the final two minutes of each phase. Measurement of VO_2 and video filming continued throughout each 20-minute test.

Athletes in the experimental group were required to practise ankling throughout the training phase of the study and if possible during competition. Prior to each efficiency test they were asked what percentage of their training time they had used the ankling technique. Based on the information supplied they were then given instructions on what percentage of time should be spent using ankling prior to the next efficiency test. Athletes were instructed that they should be using ankling as the preferred pedalling technique 100% prior to the fourth efficiency test.

Athletes in the control group were given a sham intervention. They were required to keep a diary of their training and racing and to record any periods of illness or other disruptions to training. This data has not been analysed. Apart from these interventions all volunteers were required to maintain their normal training and racing regimes. Cyclists were also required to abstain from training for the 24-hr period prior to each test.

Athletes were requested to comply with the following dietary guidelines for 48 hrs prior to any testing procedure: 60% carbohydrate, 25-30% fat and 10-15% protein. Efficiency tests were done on the same day of the week and the same time of the day for each athlete.

3.6 $VO_{2\,MAX}$

Cyclists were required to warm up on the bicycle for 5 minutes at 75-100 watts. The $VO_{2\,max}$ test then commenced with the athletes cycling at 100 watts with increments of 25 watts.min^{-1} to volitional fatigue. They were allowed to select their own cadence but required to maintain this once the test commenced. Expired air was directed through a low-resistance valve (Hans Rudolph 2-way non-rebreathing valve (2700) MO, USA.) then to a 3L mixing chamber, (Collins, Australia). A sample of mixing chamber gas was analysed for oxygen (Applied Electrochemistry S-3AII, AMETEK, PA. USA.) and carbon dioxide (Applied Electrochemistry CD-3A, AMETEK, PA. USA) and the volume of this gas was measured with a turbine ventilometer (KL Engineering, CA, USA.) and corrected in the software (Vista-Turbofit V3.2, VACUUMED, California, USA). The gas analysers were calibrated before and checked after the test using high precision gases that were similar to lung gas concentrations. In the event that there was a difference between the expected and measured concentrations of O_2 and CO_2 at the end of a test in excess of 0.03% then the test data was interpolated to correct for this shift in calibration. In this study this problem did not arise and therefore there was no need to correct the data. Oxygen consumption (VO_2), carbon dioxide output (VCO_2) and ventilation (VE) were determined every 15 seconds using standard algorithms installed in an IBM compatible computer (Vista-Turbofit V3.2, VACUUMED, California, USA). Heart rate was monitored throughout the test by electrocardiography (X–SCRIBE ™, Mortara, Milwaukee, USA); six lead recordings were obtained for those under 50 and 12 lead recordings for those over 50 years. Prior to each test a cooling fan was placed in front of the cyclist. The position of the fan was recorded to ensure the correct position each time. The environment of the laboratory was controlled with air-conditioning to 22 degrees.

Criteria that was used to terminate the $VO_{2\,max}$ test included:

(i) Cyclist wished to stop

(ii) Cyclist experienced chest pain (typical of angina), severe shortness of breath or any other pain related to, or caused by the exercise.

(iii) Cyclist wished to continue but there were abnormal changes to the ECG (especially rhythm or rate) or other signs of metabolic, cardiorespiratory or thermoregulatory distress were evident (e.g. facial pallor)

(iv) Cyclist's sweating responses were inappropriate to the environmental conditions in the laboratory.

(v) Cyclist had reached $VO_{2\,max}$ (i.e. the subject was able to exercise safely up to $VO_{2\,max}$)

3.7 VENOUS CATHETERISATION

One hand was pre-warmed (Figure 3.13) and then a 20-gauge teflon catheter (Jelco, Johnson & Johnson, USA.) was inserted into a dorsal hand vein. The catheter was connected to a 25-cm minimum volume extension tube (Tuta, Melbourne Australia) and a three-way stopcock for ease of sampling. The deadspace of the tubing and stopcock did not exceed 0.3 ml. Patency was maintained between samples by filling the tubing with heparinised saline (10 $IU.ml^{-1}$ heparin).

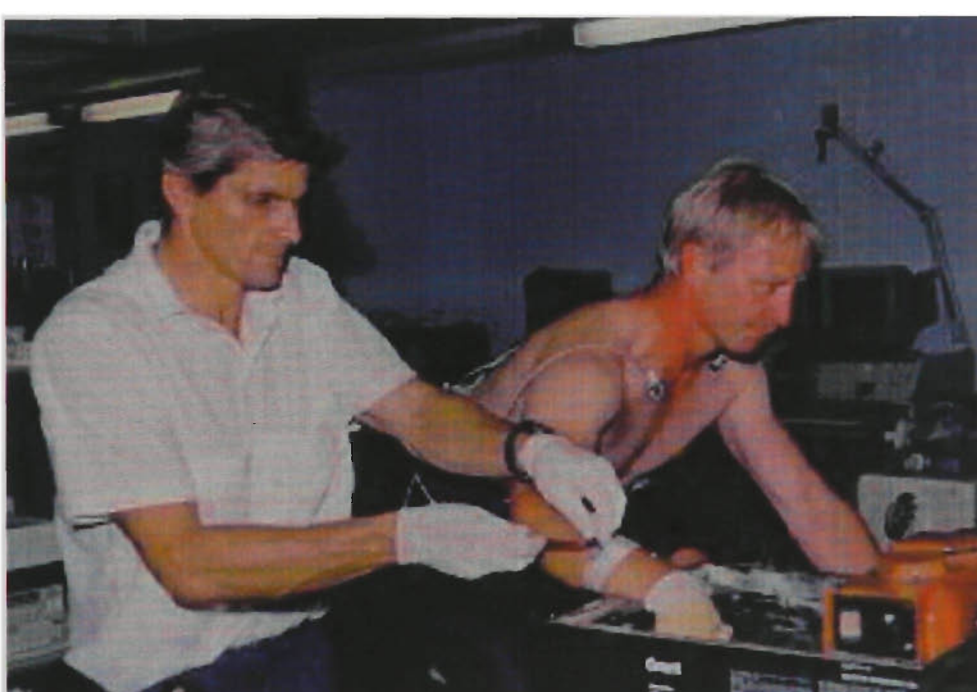


Figure 3.13 Arterialised blood sampling.

The dorsal hand vein was arterialised according to the method of McLoughlin et al. (1992), by placing the gloved hand in a warm water bath throughout the test (44-45 °C).

3.8 ANAEROBIC THRESHOLD TESTING

3.8.1 Lactate Threshold (Lactate turnpoint as defined by Davis et al., 1983)

After the volunteers had been cannulated for blood sampling (see Section 3.7 above), they were required to warm up on the bicycle for 5 minutes at 75-100 watts and then progressively increase the workload until 50% of their previously determined $VO_{2\ max}$ workload was reached. The lactate threshold (LT) test commenced at 50% $VO_{2\ max}$ with increments of 10 watts.min^{-1} to volitional fatigue. Blood samples were taken at two-minute intervals (i.e. at 20-watt increments). They were allowed to select their own cadence but required to maintain this once the test commenced. The test-retest reliability of this LT protocol was assessed in five consecutive volunteers (Appendix G). The criteria used to terminate the LT test were the same as for the $VO_{2\ max}$ test (Section 3.6 above). The LT was determined as the second breakpoint in lactate versus power. Each LT was detected by two experienced observers, with the mean of these being used subsequently. In the case of a disparity of more than 20 watts between the two observers, a third observer was utilised and the closest two were averaged.

3.8.2 Ventilatory Threshold (second breakpoint as defined by Skinner and McLellan, 1980; and McLellan, 1987)

Eight participants either did not consent to blood sampling or could not be cannulated. These underwent a ventilatory threshold (VT) test using the same warm-up and exercise protocol as for the LT test. The criteria used to terminate the VT test were the same as for the $VO_{2\,max}$ test (Section 3.6 above). The VT was determined as the second breakpoint in VCO_2 versus VO_2, VE versus VO_2 and VE/VO_2 versus VO_2 (sample graph in Appendix H). The procedure used was the same as that used in the LT test in Section 3.8.1.

3.9 PEDALLING EFFICIENCY TESTING

Prior to the efficiency test, the athlete's bicycle was attached to the ergometry system and prepared for filming with a video camera. The moving parts of the bicycle were masked with black tape to minimise unwanted reflections affecting the auto-digitising of the video film. The inseam leg length of each athlete was measured while wearing cycling shoes to ascertain the ratio between leg length and seat height. Once seated on the bicycle, with the crank at 90° (horizontal; Figure 3.14), a plumbline was dropped from the front of the kneecap.

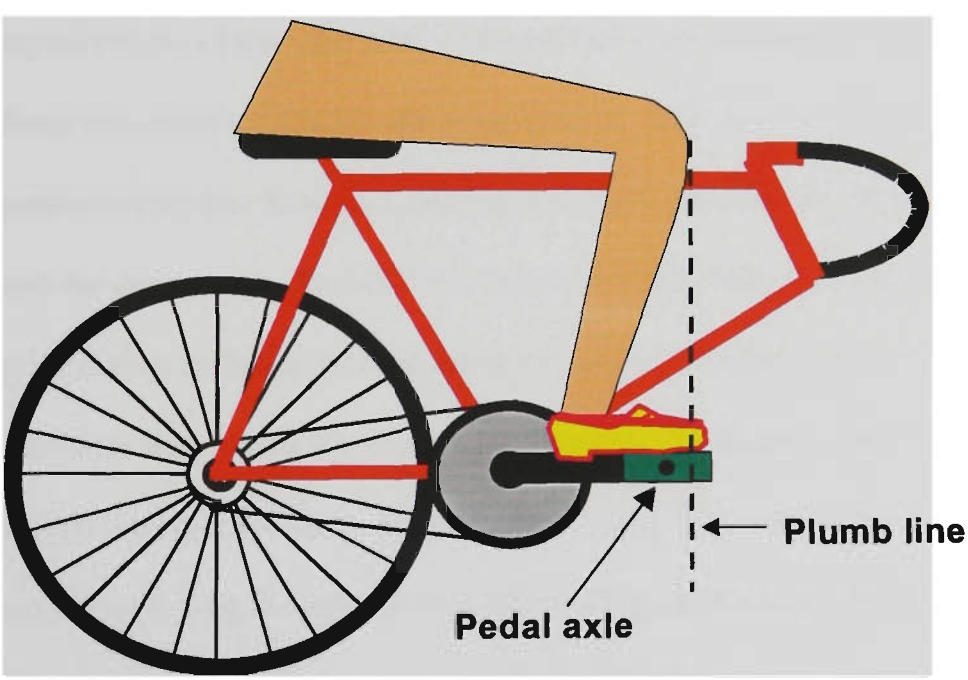


Figure 3.14 Measurement of horizontal distance between kneecap and pedal axle

The horizontal distance from this plumbline to the pedal axle was measured. This distance and the seat height/leg length ratio were measured in order to determine if they were covariables in the calculation of biological efficiency. The bicycle-seat position and handlebars were set-up identically for each test within each participant. This guaranteed that the settings were constant within cyclists.

3.9.1 Ergometer Set-up and Calibration

Once the bicycle had been set-up and attached to the ergometry system, the tyre pressure was adjusted to 100 pounds per square inch (psi) and the ergometer was then calibrated. The load cell used to measure contact force was connected to the alternator and the alternator drive roller was positioned so that there was minimal clearance between the bicycle tyre and the drive roller. As data input was required by the microcomputer prior during calibration of the contact force, a tacho and cadence simulator was attached to the microcomputer. The contact force was zeroed and then the force was adjusted to 6 kg. The tacho contact force was set to 4.5 kg. If the tacho started bouncing on the tyres, due to tyres that were not perfectly circular, the contact force was increased by up to 1 kg. The tacho/cadence simulator was removed and the top dead centre (TDC) sensor position was adjusted so that an indicator light turned on when the crank was at TDC (measured with a plumb bob). The sensor was positioned so that flexing of the bicycle frame (common with aluminium bicycles) would not cause the frame to connect physically with the sensor. Sensitivity of the sensor was adjusted to ensure reliable operation. When the volunteer was sitting on the bicycle, the system was tested with the cyclist pedalling with no load resistance for one minute. The volunteer was then fitted with a valve to measure metabolic data and instructed to commence pedalling (Figure 3.15).

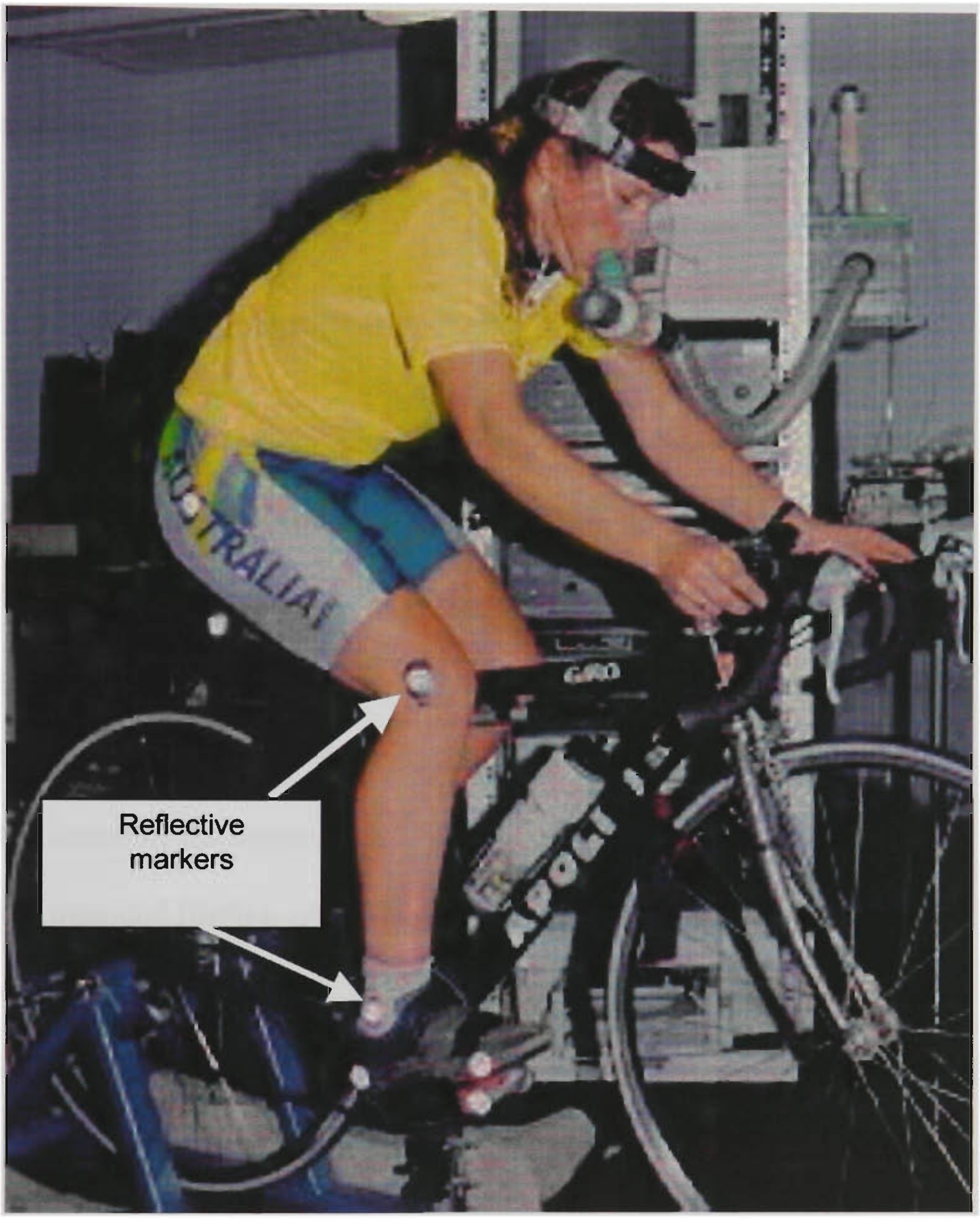


Figure 3.15 Metabolic, power and video data collection.

Once the volunteer was pedalling at the desired workload, the measurement system was reset. This forced the micro-computer to recount the tacho pulses for one crank revolution and reset the micro counter-timer circuitry ensuring the correct pulse count per degree of crank revolution (see Section 3.2.1.1). The resistive load was adjusted prior to commencement of the test using a combination of gearing and load resistance settings to achieve the desired power.

3.9.2 Video Set-up

The video camera (Panasonic NV-MS4 SVHS, National, Japan) was located on a rigid tripod 5 metres away from the right hand side of the bicycle. The remote location of the camera from the bicycle reduced the perspective error due to target objects being at different distances from the camera lens (Bartlett, 1989). The height of the camera was set so that the centre of the camera lens was in a horizontal line with the centre of the intended filming area. The camera was aligned in the horizontal plane with a spirit level placed on top of the camera. A 60-watt light was positioned immediately below the camera to illuminate the cyclist while pedalling. Reflective markers were placed on the hip, knee, ankle, toe and heel, using standard anatomical methods, as well as the crank centre and pedal centre (Figure 3.15). The markers were constructed from foam balls of 20mm diameter cut in half and covered with reflective tape. The markers were attached to the athletes' skin with adhesive foam pads and to the crank centre and pedal centre with double-sided tape. The movement pattern to be analysed was circular therefore, it was not necessary to film a distance scale. The camera lens was focused manually and shutter speed set to 1/500th of a second. A light emitting diode (LED) was positioned in the field of view of the camera lens and was illuminated at the start and finish of each two-minute data collection. The zoom was adjusted so that the subject image was as large as possible. Videotapes used for the study were TDK Extra High Grade E180.

3.9.3 VO_2 Measurement

Athletes were required to warm up on the ergometer for 5 minutes at 75-100 watts and then the load was incremented until the power corresponded to 95% of anaerobic threshold, previously determined. Volunteers selected their own cadence but were required to maintain this once the test commenced. Metabolic analysis was identical to $VO_{2\ max}$ testing (see Section 3.6). In the event that cyclists felt that they could not maintain the required work rate for the full 20 minutes, the power was reduced during the warm-up phase. Similarly, where the RER was below 0.90 (in three of 41 tests), the load was increased to a level that the cyclists could maintain for twenty minutes. All volunteers were then able to complete four efficiency tests at approximately the same power for each test (four volunteers had variations where the coefficient of variation% (CV% = SD/mean expressed as a %) was >10%) and with the mean respiratory exchange ratio (RER) being maintained at less than 1.0 The variables collected were RER and VO_2 ($L.min^{-1}$).

3.9.4 Power output and Cadence Data Analyses

Mechanical power and cadence data were smoothed using Savitzky and Golay's least squares method (Savitzky and Golay, 1964). Data was overlaid graphically with the raw data to ensure that only noise was removed.

3.10 BIOCHEMICAL ANALYSES

For each sample, 5 ml of blood was drawn into a plain syringe, transferred immediately to a lithium heparin tube (10 IU heparin), mixed and placed into an iced slush. A sample of whole blood was transferred to three hematocrits that were plugged and then spun for 5 minutes. The remainder of the blood was then transferred to three 1.5 ml Eppendorf tubes and centrifuged (10,000 rpm) for 3 minutes. An aliquot (300 μl) of plasma was taken off and was transferred to 600 μl of cold 0.6 molar Perchloric acid (PCA; Stat-Pak™ Rapid Lactate test, Behring Diagnostics Inc.), mixed vigorously and centrifuged (10,000 rpm) for 2 minutes. The resultant supernatant was frozen in liquid nitrogen and stored at -80 °C until analysis in duplicate for plasma lactate concentration. The remaining plasma was frozen in liquid nitrogen and stored at -80°C.

3.10.1 Development of the Lactate Analysis Procedure

Prior to data collection and analysis, the lactate analysis protocol was developed starting with a procedure that had been developed previously for use with the Shimadzu spectrophotometer. The concentrations of the ingredients of the reagent produced less than satisfactory results when tested with the Cobas Bio spectrophotometer (Cobas™ Bio 8326, F. Hoffman – La Roche & Co. Limited Company Basle, Switzerland). Concentrations of the ingredients were varied progressively until a linear regression was obtained.

The final regression of the La⁻ samples versus the absorbance data supplied by the Cobas Bio spectrophotometer produced a linear regression R^2 of 0.999 (Figure 3.16).

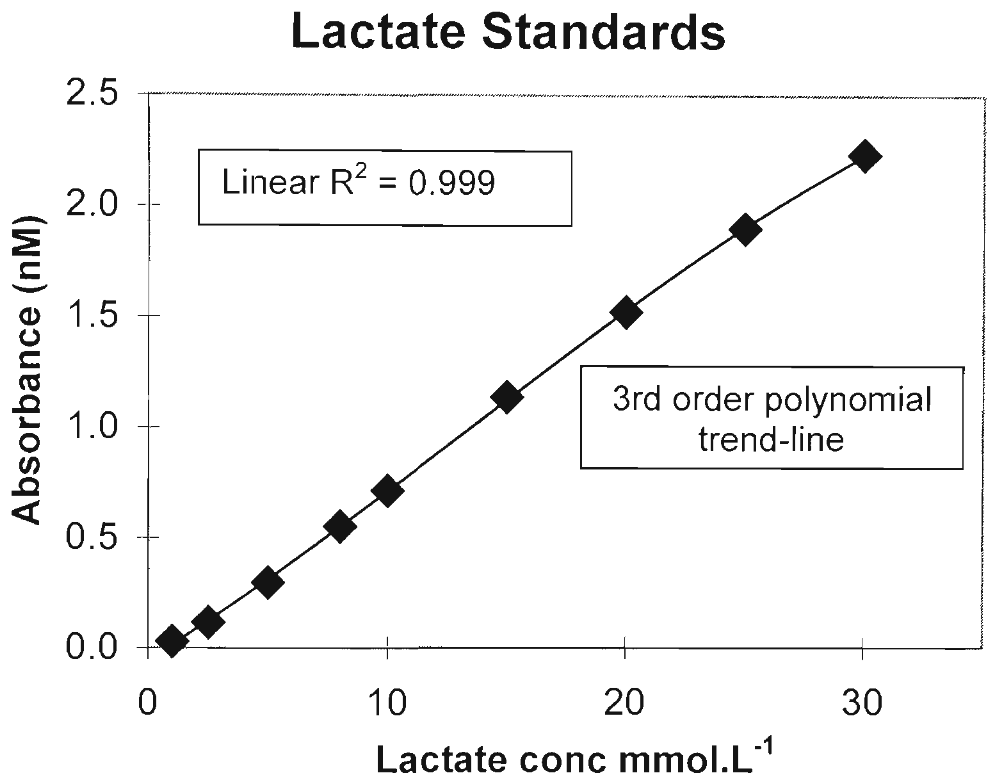


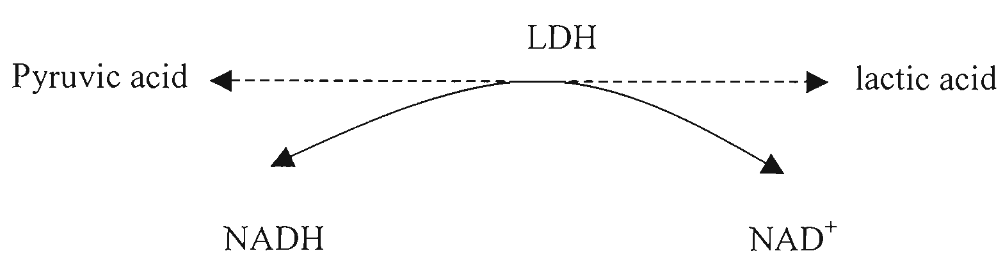
Figure 3.16 Analysis of lactate samples using the Cobas Bio spectrophotometer.

Curvature was observed in the data that was confirmed by overlaying the plotted data with a third order polynomial trendline. Further regression analysis indicated that a cubic polynomial was the best fit with this set of La⁻ samples. A statistical analysis procedure was subsequently developed using Microsoft Excel spreadsheet software.

Statistical analysis was carried using the data analysis procedures in Microsoft Excel 5. The analysis procedure included a test of significance to determine if the higher order polynomials produced a significantly better R^2 result than could be obtained with a linear regression (Appendix I).

3.10.2 Lactate Concentration

Lactate was analysed based on the method of Lowry and Passonneau (1972).



The change in absorbance at 340 nano meters, due to the increase in NADH, is proportional to the plasma lactate concentration (Behring Diagnostics Inc, 1988). Lactate standards were prepared from 0.05 mol lactic acid. 0.6 mol perchloric acid was added in the ratio of 2:1. Reagent was prepared by adding 5 mls milli-Q H_2O, 7 mls glycine-hydrazine buffer and 0.4 mls lactate dehydrogenase to 0.02 gms NAD (this formula was developed by the author to obtain the best R^2 from a regression analysis of measured La^- versus expected La^-). Analysis was carried out using a semi-automated system (Cobas^{TM} Bio 8326, F. Hoffman – La Roche & Co. Limited Company Basle, Switzerland; analysis parameters are listed in Appendix J). Standard graph for the lactate assay is given in Appendix K.

3.11 DATA ANALYSIS

3.11.1 Video Analysis and Macro Development for Spreadsheets

The primary objective of analysis of the videotapes was to confirm that the technique used during the ankling phase produced a more effective application of force than for normal technique (i.e. orientation of the foot perpendicular to the tangential movement of the crank). Prior to digitising any of the videotapes the aspect ratio of the video camera, used in this thesis, was determined by filming a square object. The corners of the square were digitised using the Peak 5 Motion Measurement System (Peak Performance Technologies Inc, Co, USA). The Peak software was used to calculate the aspect ratio, which was stored in a data file. The videotapes were encoded with Peak time code, as a time code generator was not available during filming. A spatial model was set-up to define the number of points to be digitised. The points set-up were hip, knee, ankle, heel, toe, crank centre and pedal centre. Angles defined were the knee, ankle, foot relative to crank arm and crank arm relative to the vertical (Y) axis. The Y-axis remained constant throughout as the camera was levelled prior to each testing session (see Section 3.9.2 above). Angular data only was collected. Five seconds (250 frames) were auto-digitised within each 2-minute film segment (digitising the whole 2-minute segment was not feasible due to the high demand on the use of the equipment). The five-second window commenced one minute after the start of each film segment. This ensured that athletes had used the technique (normal or ankling) for four minutes continuously, including three minutes when data was not being collected. The data files were conditioned using a Fast Fourier Transform (Press, 1988) as the data was cyclical in nature. Optimal cut-off frequencies were chosen by the Jackson Knee method (Jackson, 1979). A batch file was created in the Peak software to automate the conditioning process. Data was saved in a format that was suitable for importing into a spreadsheet. Macro programs were embedded in the spreadsheet to obtain the following data for each test:

- The mean angle between the foot and the crank for 0-180° - provided a quantitative measurement of the pedalling technique Figure 3.17).
- Maximum angles of plantar and dorsi-flexion and range of motion of the ankle - provided evidence whether ankling technique was possible.

- Angle of crank where maximum power exhibited - this angle was then able to be related to the angles reported in the literature where maximum force was applied (as measured with force pedals).

Angle between pedal and crank = θ

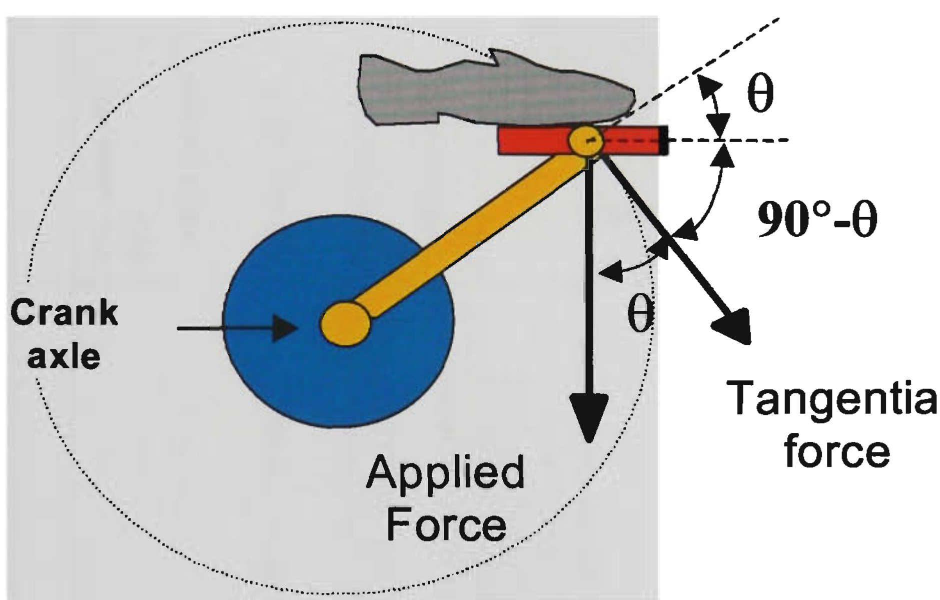


Figure 3.17 Measurement of the angle between the foot (pedal) and the bicycle crank (θ)

3.11.2 Biological Efficiency Calculations

Formulae

$$\text{Efficiency\%} = \frac{\text{Power output (watts)}}{\text{Power input (watts)}} \times 100$$

(deVries H. A., and Housh, T. J., 1994)

$$\text{Power Input} = \text{Energy expended (l.min)}^{1,2} \times 69.759 \text{ (watts)}^{3}$$

1. For exercise with RER <1.0, a look-up table (Carpenter, 1964) converts Non-protein RER to the energy equivalent (EE) ($kcal.min^{-1}$) per litre of O_2.
2. Using the look-up table (displayed graphically in figure 3.13), a regression equation was derived to calculate EE (dependent variable = EE, independent variable = RER) This regression provided a quantitative means of interpolating data that lay in between discrete values in the look-up table and graph.

   Regression equation:
   EE ($kcal.min^{-1}$) = 3.815291 + 1.231664.RER

3. Conversion of EE to power input (watts) (Flood, 1996).

$$\text{Efficiency\%} = \frac{\text{Power output (watts)}}{\text{EE }(kcal.min^{-1}) \times VO_2\ (l.min^{-1}) \times 69.759} \times 100$$

Example

VO_2 (litres.min^{-1})	= 2.42 $l.min^{-1}$
RER	= 0.98
∴ EE ($kcal.min^{-1}$)	= 3.815291 + (1.231664 x 0.98)
	= 5.02 kcalories
∴ Power input	= 5.02 x 2.42 x 69.759
	= 848 watts
Power output	= 190 watts
∴ **Efficiency%**	$= \frac{190}{848} \times 100$ = **22.4%**

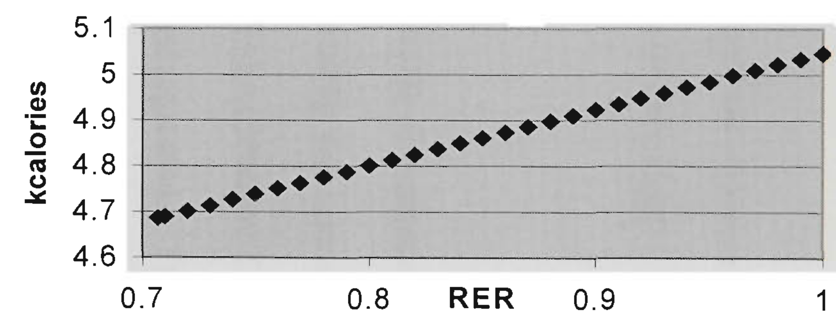


Figure 3.18 Caloric values for Nonprotein R

3.12 STATISTICAL METHOD

Analysis of the data was carried out using SPSS version 8.0 (SPSS Inc, Chicago, USA) and Microsoft Excel 5 & 97 (Microsoft Corporation, USA).

During the development and calibration of the electronically-braked ergometry system the data were analysed using descriptive statistics (means) together with graphs, Pearson Product Moment Correlation Coefficients (r), and linear and polynomial regressions. The data obtained from the calibration of the air- and mechanically-braked ergometers were similarly analysed using descriptive statistics (means and SDs) and polynomial regressions. The data from the pilot study were analysed with the aid of Student independent T-tests as well as descriptive statistics (means and SDs). The main results, described in Chapter 7, were analysed using Pearson product Moment Correlation Coefficients (r), Student T-tests (independent and repeated measures) and Analysis of Variance (ANOVA) for a test of homogeneity. The biological efficiency data was analysed with a two by two factorial ANOVA and a two by two factorial ANOVA and ANCOVA with repeated measures on the second factor. Data additional to the central hypotheses was analysed using descriptive statistics only.

CHAPTER 4

DEVELOPMENT AND CALIBRATION OF THE ELECTRONICALLY-BRAKED BICYCLE ERGOMETRY SYSTEM

4.1 DEVELOPMENT

The aim of this phase was to develop an ergometer that accurately measured mechanical power at each degree of crank revolution. During the development, the mechanical power input was provided by the calibration system. In Chapters 6&7, the cyclists provided the mechanical power input.

4.1.1 Alternator Modifications

With no electrical load resistance, the alternator still provided an inherent electrical resistance that manifested as a mechanical resistance to the calibration system. The mechanical power input needed to overcome this inherent load was dependent on cadence (Table 4.1)

Footnote: A decrease in the electrical load resistance (R), with a constant output voltage (V), results in an increases current flow (I) and therefore a higher power output (P) ($P = I^2 \times R$ or $P = V^2 /R$; Weir, 1987). Throughout this thesis, "load resistance" refers to the electrical load resistance.

Cadence	Mechanical power input
60	89 watts
90	144 watts
115	180 watts

Table 4.1 Threshold of mechanical power input to the alternator

At a cadence of 90 revolutions per minute (RPM), a cadence typically used competitive cyclists, the inherent load of 144 watts was considered to be too high for anaerobic threshold testing (Craig and Walsh, 1989). This meant that modifications to the alternator were required to lower this inherent load.

4.1.2 Regression of Mechanical Power Versus Electrical Power

The first regression analysis (electrical versus mechanical power) was conducted by measuring the mechanical power input for three discrete electrical resistances (32, 54 and 108 ohms). For each resistance, mechanical power was measured for a range of cadence, commencing at 60 RPM with five-RPM increments to 115 RPM (Figure 4.1).

In reviewing the data, the three curves (Figure 4.1) corresponded to the calibration curves generated for the three discrete resistances (Figure 4.2). From this initial data, it was clear that the electrical load resistance across the output of the alternator was an independent variable in the prediction of mechanical power.

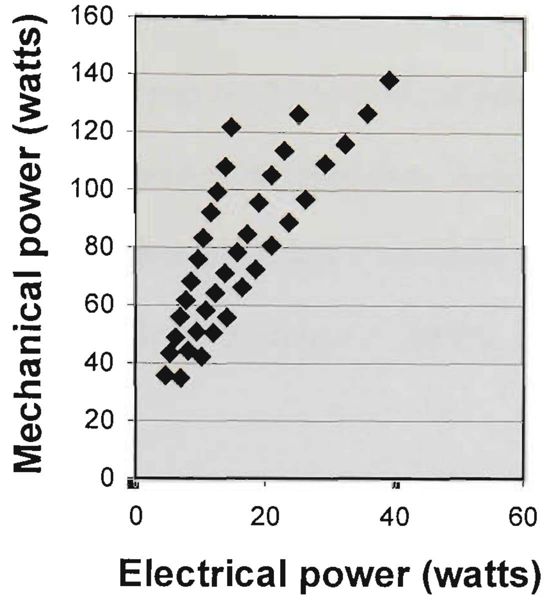


Figure 4.1 Mechanical power versus electrical power with cadence varying between 60 and 115 RPM.

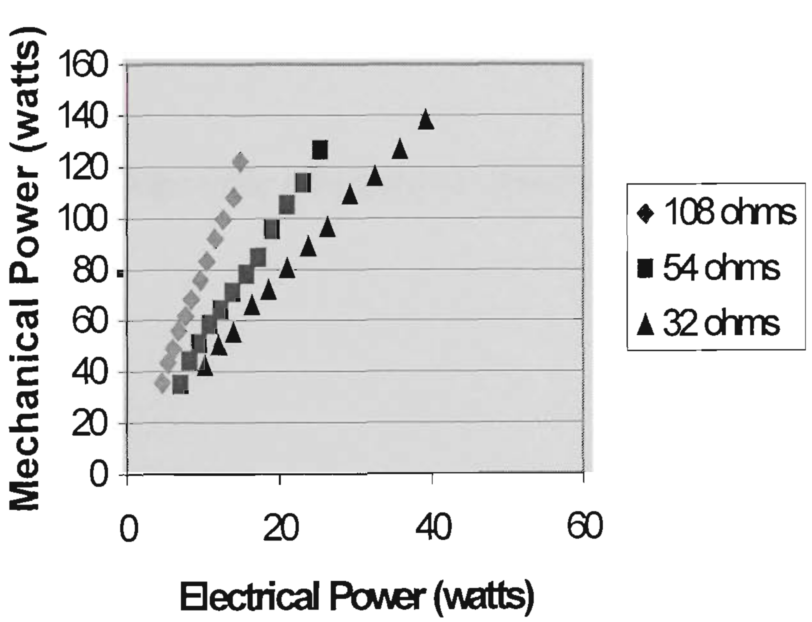


Figure 4.2 Mechanical power versus electrical power for three load resistances.

4.1.3 Alternator Efficiency

Alternator efficiency was calculated and tabulated versus mechanical power. The data supported the decision to include the load resistance as an independent variable as efficiency increased with a decrease in load resistance (Table 4.2); see Section 4.1.2 above.

Load Resistance (ohms)	**Alternator efficiency %**
108	12.5
54	19.5
32	26.2

Table 4.2 Alternator efficiency for three load resistances

4.1.4 Load Cell and Tachometer (tacho) Pulse Count Stability Tests

Thirty tests were carried out to determine the stability of the load cells (and associated circuitry) and the tacho pulse count for each crank revolution. Each load cell was tested with a zero load for over 20 minutes. The output varied by up to 87 grams (0.7%) that exceeded the resolution of the load cells (0.1%). The problem was caused by drift in the voltages supplied to the load cells and their associated amplifiers. The power supply used to supply these voltages was subsequently rebuilt.

Data collected included the number of output pulses from the tacho. Examination of this data revealed that the tacho count varied by only 0.1% at low cadence (70 RPM) but as much as 0.73% at high cadence (118 RPM). This was considered unacceptable. The cause was unknown at this stage, however it was suspected that there was some slippage between the tacho roller and the rear wheel of the bicycle.

To further investigate the variability in the tacho pulse count a series of tests were carried out from 40 RPM to 120 RPM in 10-RPM increments.

RPM	Tacho pulse count variability (%)
40	0.12
50	0.15
60	0.07
70	0.09
80	0.10
90	0.20
100	0.79
110	0.94
120	0.77

Table 4.3 Tacho pulse count varies with RPM

The tacho pulse count for each crank revolution varied by as much as 0.94% (Table 4.3) and was significantly related to cadence ($r = 0.82$, $p = 0.006$). After investigation of the possible causes, the variability in tacho pulse count at high cadence was caused by slippage between the tacho roller and the bicycle rear wheel. This was rectified by increasing the contact force between the tacho roller and the bicycle rear wheel to 4.5 kg. Variation of the pulse count was reduced to 0.02% during the final recalibration.

4.1.5 Load Cell Calibration

Load cells were calibrated prior to the second and third (final) calibration tests. During the first load cell calibration two problems were detected. First the load cells needed a warm-up time prior to use as drift was observed between the start and finish of the calibration. Repeated testing indicated that a warm-up time of 30 minutes would be more than adequate. This was subsequently added to the protocol for the main study.

The second problem was an error in the software for one of the load cells. After correction both load cells were tested with a known weight and were within the resolution of each load cell (±0.1%).

4.2 ALTERNATOR DRIVE ROLLER ERRORS

After the second calibration, an investigation of the data between the start of the calibration and the end found a significant error (5%) in the predicted mechanical power. The wear in the alternator drive roller that changed the alternator RPM and therefore the efficiency of the alternator caused this error. This resulted in a higher electrical power output for the same mechanical power input supplied by the calibration system. Although the alternator RPM could be measured, the microcomputer hardware needed major redesign to incorporate the required extra circuitry. The feasibility of measuring the diameter of the drive roller prior to each test in the main study was investigated. The alternator RPM (by measuring the frequency from the phase output) was compared with the diameter for three rollers which each had different diameters (Table 4.4). A correlation of 0.99, $p = 0.04$ was obtained between the measured diameter and the frequency output. An observation of the raw data indicated that a one tailed test of significance could be used. Although only three samples were obtained, this high R-value clearly indicated that the diameter could be used as a substitute for the measured RPM. The measured diameter was subsequently added to the software program via Winlog.ini, a file that loaded parameters into the software during set-up prior to use as a measurement device.

Measured diameter (mm)	Frequency (hertz)
74.72	311.0
74.34	311.2
72.21	317.7

Table 4.4 Alternator drive roller wear - comparison of diameter with alternator frequency output.

4.3 CALIBRATION

The measurement of mechanical power depended on five independent variables, namely electrical power, alternator contact force, alternator load resistance, cadence and tacho pulse count per crank revolution (the latter is dependent on the gearing ratio selected). However calibration was achieved by inputting known mechanical power (from the calibration system, Hooper, 1977) for a range of contact force, alternator load resistance and cadence and measuring the resultant electrical power. During the development of the ergometry system and prior to use as a measurement device the system was calibrated three times. The first calibration was preliminary to find the significant variables to predict mechanical power. The equipment was subsequently used in a pilot study to test for reliability of the equipment. The second calibration was to have been the final calibration prior to use but variability in the data was detected caused by the wear in the alternator drive roller. A new roller was fitted, the variability of the wear in the roller was accounted for in the software (see Section 4.2) and a final calibration was carried immediately prior to the first $VO_{2\ max}$ test in the main study. Prior to commencing the second and third calibration of the ergometry system, the load cells were calibrated with a test weight. The stability of the load cells was found to be within the resolution of each load cell (±0.1%).

4.3.1 First Calibration

This calibration was carried out using three gears of a racing bicycle, low (42:23), medium (42:17) and high (42:13). For each gear, cadence was varied from 40 to 120 in 10-RPM increments, alternator contact force was varied from 5 to 8.5 kg in 0.5-kg increments and load resistance was varied from 2 to 544 ohms in 12 equal steps. Therefore the complete calibration process measured mechanical power for 2,592 steps in total. This corresponded for a range of mechanical power of 25 to 662 watts. Calculation of the dependent variable, mechanical power, was achieved by the measurement of the three primary variables namely force, distance and output angular shaft velocity (Figure 4.3) and using standard physics formulae (shown below). Force was measured by a load cell (Figure 4.3), output angular shaft velocity was measured by measuring the angular crank velocity of the bicycle attached to the calibration system (the bicycle crank was connected to the output shaft) and the displacement was a known measurement of 4 metres (0.6366 metre radius between the load cell attachment point and the output shaft; Hooper, 1977).

Formulae for calculating mechanical power generated by the calibration system

$$Power = \frac{Work}{Time}$$

$$Work = Force \times Displacement$$

$$Force = Mass \times Gravity$$

$$\therefore Power = \frac{Mass \times Gravity \times Displacement}{Time}$$

$$Power(watts) = \frac{Mass(kg) \times 9.807 \times 2\pi \times 0.6366 \times Cadence(RPM)}{60}$$

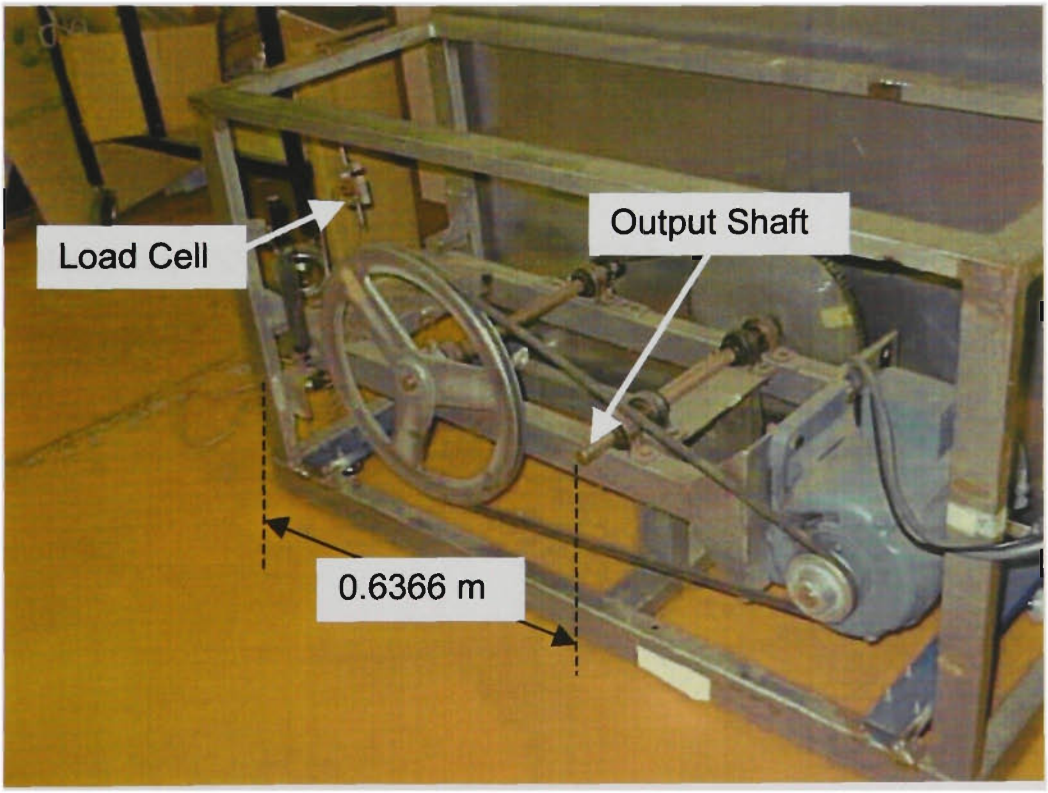


Figure 4.3 Calibration system

A stepwise regression was used to find the significant variables needed to predict mechanical power. A significant R^2 of 0.992 ($F(5) = 68091$, $p = 0.000$,) was obtained with a standard error of 9.9 watts. No violation of assumptions were detected (especially collinearity). The following significant variables were used in the prediction equation:

- Alternator contact force
- Cadence (RPM)
- Digital electrical load resistance value
- Alternator electrical power output
- Tacho pulse count per crank revolution

The graph of predicted mechanical power versus measured mechanical power output from the calibration system (Figure 4.4) shows a linear relationship. ANOVA table and a table with beta coefficients, standard error statistics and multicollinearity statistics are presented in Appendix L.

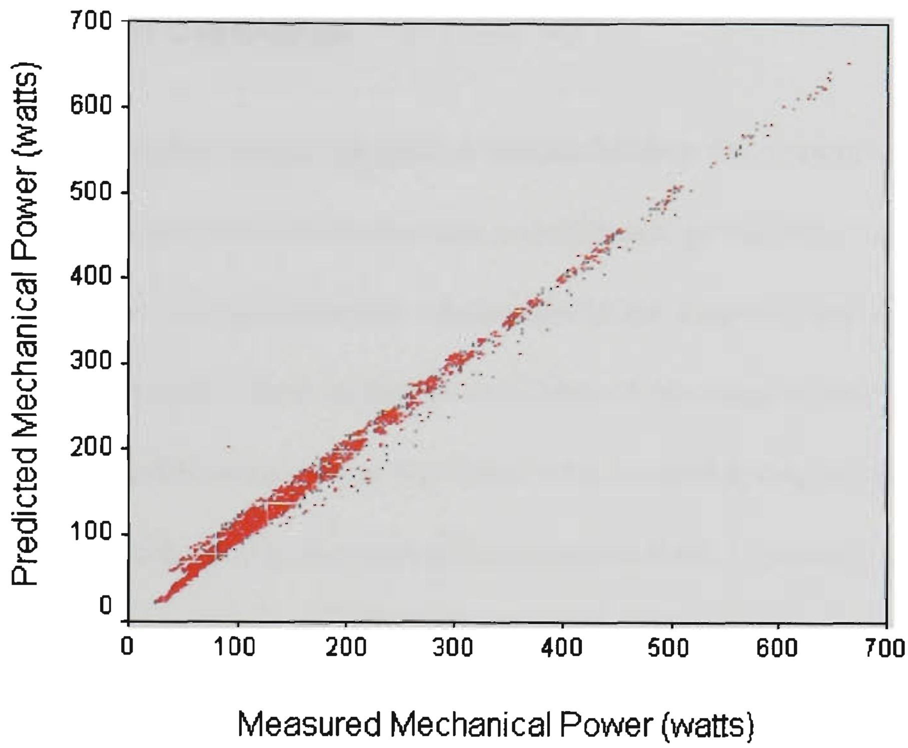


Figure 4.4 Graph of predicted value of mechanical power output versus measured mechanical power output.

With an R^2 of 0.992, 1/(1-R2) = 125. As all of the Variance Inflation Factors (VIFs) are much smaller than this (Appendix L) then it may be assumed that little or no multicollinearity existed between the independent variables. Multicollinearity could occur if too many variables had been put into the model (Freund and Littell, 1986).

The prediction equation data from this calibration were used in the pilot study.

4.3.2 Second Calibration

The purpose of this second calibration was to calibrate the system using the highest gearing on the bicycle with a mechanical power input of up to 900 watts. The rationale was that a better prediction equation would be obtained if the system was calibrated well beyond the range of any elite cyclists. The calibration system was fitted with a coiled spring in line with the load cell used to measure the mechanical force to protect the load cell from inadvertent damage. A complete calibration was carried out and after examination of the data it was realised that the coiled spring in line with the load cell limited the maximum power input due to the mechanical design of the calibration system.

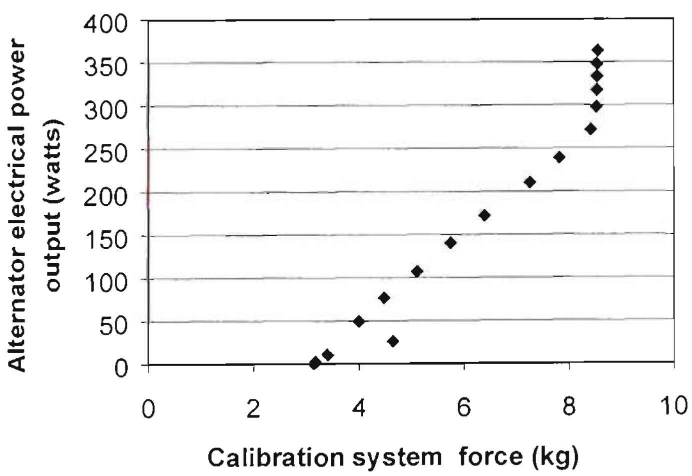


Figure 4.5 Effect of spring in line with the load cell used to measure force

The electrical power output from the alternator was increasing with no increase in the force measured by the load cell and therefore no increase in mechanical power input (Figure 4.5).

After careful consideration of the possible damage to the expensive load cell it was decided that because the load cell could cope with a force of up to 250 Newtons (25 kg) and the safe overload was 150% with a maximum overload of 500%, the risk was negligible as the force would never exceed ≈130 Newtons (13 kg). The coiled spring was replaced with a solid coupling link and the calibration was recommenced after the load cells were recalibrated. This calibration was carried out using eight gear ratios of a racing bicycle from a very low of 42:21 to the highest of 52:13. Data was removed where the software limit of the load cell, measuring the force on the calibration system, was exceed (12 kg).
A stepwise regression was used to predict mechanical power. An R^2 of 0.997 ($p = 0.000$, $F(5) = 198858$) was obtained with a standard error of 9.7 watts. ANOVA table and a table with beta coefficients, standard error statistics and multicollinearity statistics are presented in Appendix M. With an R^2 of 0.997, $1/(1-R^2) = 333$. Again as in the previous calibration all of the Variance Inflation Factors (VIFs) are much smaller than this (Appendix M) Therefore it may be assumed that little or no multicollinearity exists between the independent variables.

While investigating different regression analysis models a plot of a cubic function against the raw data showed an excellent fit (Figure 4.6).

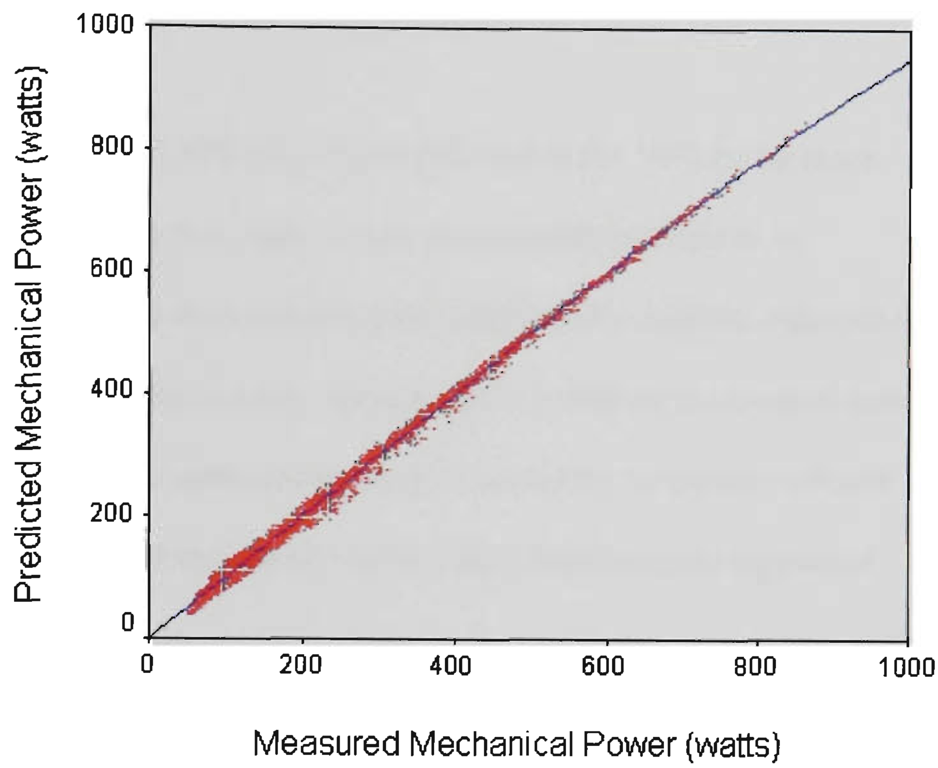


Figure 4.6 Graph of predicted value of mechanical power output versus measured mechanical power output. The line of best fit is a cubic function indicating curvilinearity.

As the alternator electrical output relative to the RPM of the alternator was also curvilinear (Appendix A), the alternator RPM was included as an independent variable. It was calculated from the tacho pulse count and the cadence. Cadence and tacho pulse count were therefore no longer included as variables as their inclusion would have violated the assumption that no strong linear relationship (multicollinearity) exists between independent variables. A stepwise regression was used to predict mechanical power. An R^2 of 0.998 ($p = 0.000$, $F(5) = 530419$) was obtained with a standard error of 6.6 watts. The ANOVA table and table showing beta coefficients, standard error statistics and multicollinearity statistics are presented in Appendix N.

With and R^2 of 0.998, $1/(1-R^2) = 500$. Due to the VIFs being much smaller than this then again it may be assumed that little or no multicollinearity exists between the independent variables Appendix N. Curvature was still evident in the graph of predicted mechanical power versus measured mechanical power. The overlay of several different curves supported the use of a cubic polynomial multiple regression (Figure 4.7).

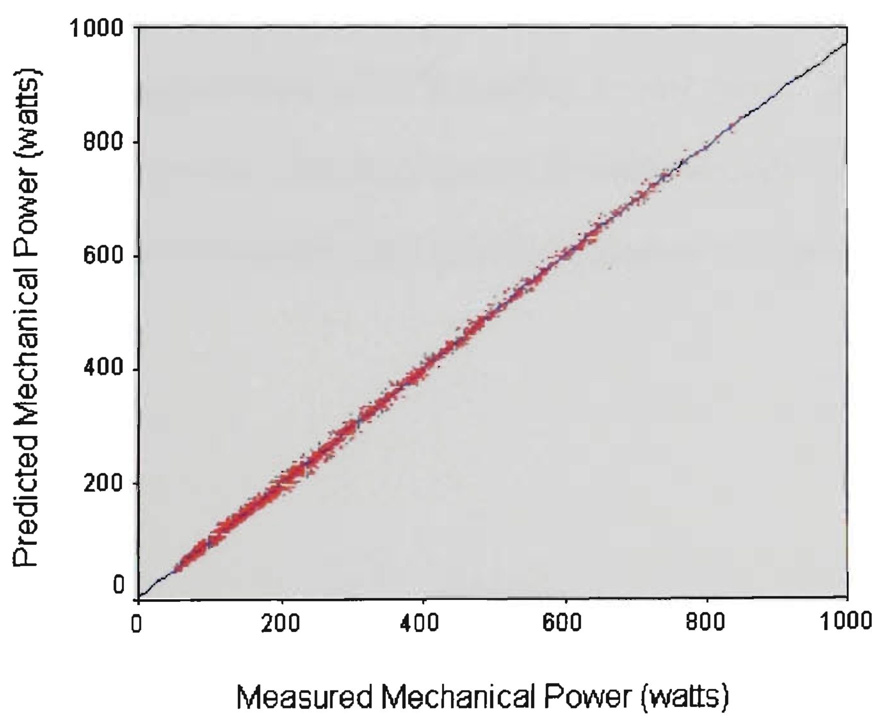


Figure 4.7 Graph of predicted value of mechanical power output versus measured mechanical power output with alternator RPM included as a variable. The line of best fit is a cubic function indicating curvilinearity.

The cubic polynomial produced an R^2 of 0.999 ($p = 0.000$, $F(17) = 177819$) with a standard error of 5.6 watts. Variables were entered in a single step (Enter method) on advice from the University statistician. The ANOVA table and a table showing beta coefficients and standard error statistics are presented in Appendix O. Multicollinearity statistics are not presented as squared and cubed functions of the independent variables would obviously be related and therefore high VIFs would be reported.

The data was then given to the University statistician to obtain assistance in finding the best model. After detailed analysis of several models the suggested preference was for a quadratic model with cross-product terms. A quadratic model produced an R^2 0f 0.999 ($p = 0.000$, $F(14) = 201292$) with a standard error of 5.8 watts. The ANOVA table and table showing beta coefficients and standard error statistics is presented in Appendix P.

4.3.3 Third and Final Calibration

Finally, after all of the known problems had been solved and all of the known errors minimised, the equipment was calibrated prior to use in the main study.

This calibration was carried out using nine gear ratios of a racing bicycle, from a low of 42:23 to a high of 52:13. For each gear, cadence was varied from 60 to 120 in 10-RPM increments and load resistance was varied from 2 to 544 ohms in 20 equal steps. Alternator contact force was varied from 5 to 8 kg in 1-kg increments during the calibration. The single contact force selected at each combination of the above variables (gearing, cadence and load resistance) was relative to the mechanical power output of the calibration system. Therefore the complete calibration process measured mechanical power for 1260 steps in total. This corresponded for a range of mechanical power of 40 to 670 watts. The regression analysis, to predict mechanical power, was carried out using the independent variables of electrical power output from the alternator, load resistance connected to the alternator, alternator contact force and alternator RPM. The alternator RPM was calculated from the tacho pulse count and cadence and corrected for the diameter of the alternator drive roller.

An R^2 of 0.999 (p = 0.000, F(14) = 114061) was obtained with a standard error of 3.8 watts. The scatterplot of predicted versus measured mechanical power has been overlaid with a linear line of best fit showing that the quadratic model was suitable (Figure 4.8).

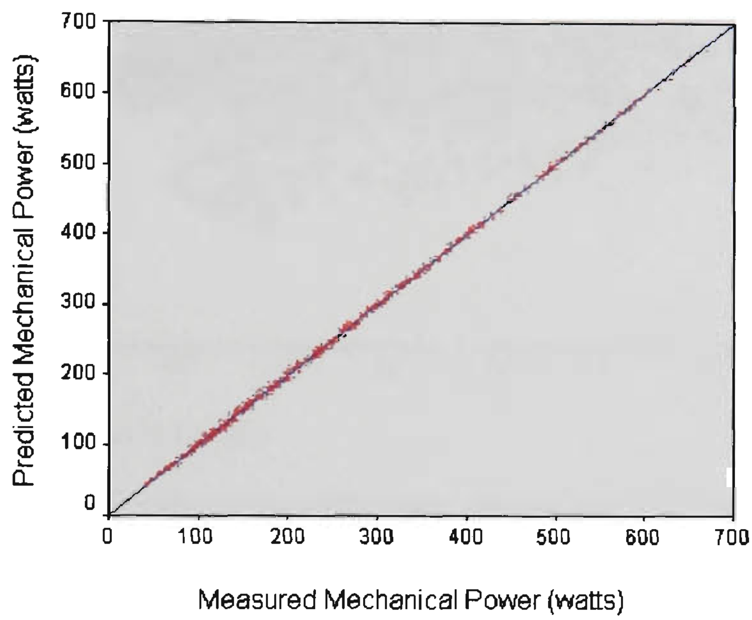


Figure 4.8 Graph of predicted value of mechanical power output versus measured mechanical power output. The line of best fit is linear indicating that the quadratic model is a good fit.

Examination of the residuals (Figure 4.9) found no requirement for transformation of the data.

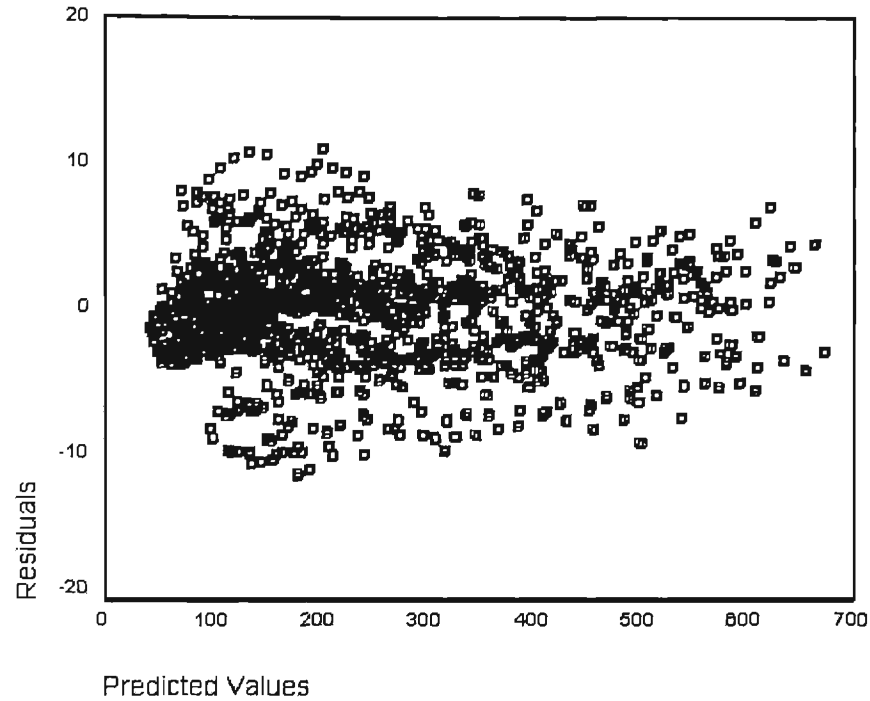


Figure 4.9 Graph of the residuals versus the predicted values of mechanical power output of the final calibration.

ANOVA table and a table with beta coefficients, standard error statistics and multicollinearity statistics are presented in Appendix Q.

As all the known variability had been accounted for either in the hardware or in the software it was not necessary to conduct another calibration to validate the third and final calibration. The equipment was then used in a pilot study (Chapter 6) prior to use in the main study (Chapter 7).

4.4 CONCLUDING REMARKS

The development of the ergometer described in Chapter 3, and the results described in Chapter 4, has centred on two main approaches.

(a) Mechanical flaws were corrected. In particular the wear of the alternator drive roller was found critical to the accurate measurement of alternator RPM.

(b) Development of the best possible statistical model to predict mechanical power. This involved identification and progressive selection of appropriate independent variables and accounting for non-linearity by selection of the appropriate polynomial function.

Because of the above, the regression equation used in this thesis to predict mechanical power has an R^2 of 0.999.

CHAPTER 5

CALIBRATION AND COMPARISON OF AIR- AND MECHANICALLY-BRAKED BICYCLE ERGOMETERS USED IN THE LABORATORY TESTING OF ATHLETES

5.1 INTRODUCTION

The objective of this part of the thesis was to compare the precision of commercially available ergometers with the electronically-braked system developed for this thesis. Three ergometers were calibrated, one mechanically-braked (Monark, Varberg, Sweden; Model 864) and two air-braked (REPCO and Victorian Institute of Sport prototype). Included in the calibration of the air-braked ergometers was a qualitative analysis of the effects of objects that may interfere with airflow near the ergometer. These comprised walls and electric fans. The REPCO calibration system (Hooper, 1977) that was used to calibrate the present ergometer (see Chapter 4) was also used to calibrate these other ergometers.

5.2 METHODS

5.2.1 Calibration of a Mechanically-Braked Ergometer

Two sets of data were collected from a Monark mechanically-braked bicycle ergometer (Model 864; Monark, Varberg, Sweden); one set of data using the traditional method of calibration, as developed by Monark using a weight cradle and the second set of data from a modified force measurement system using a strain gauge that had been fitted in sequence with the friction cord (Figure 5.1).

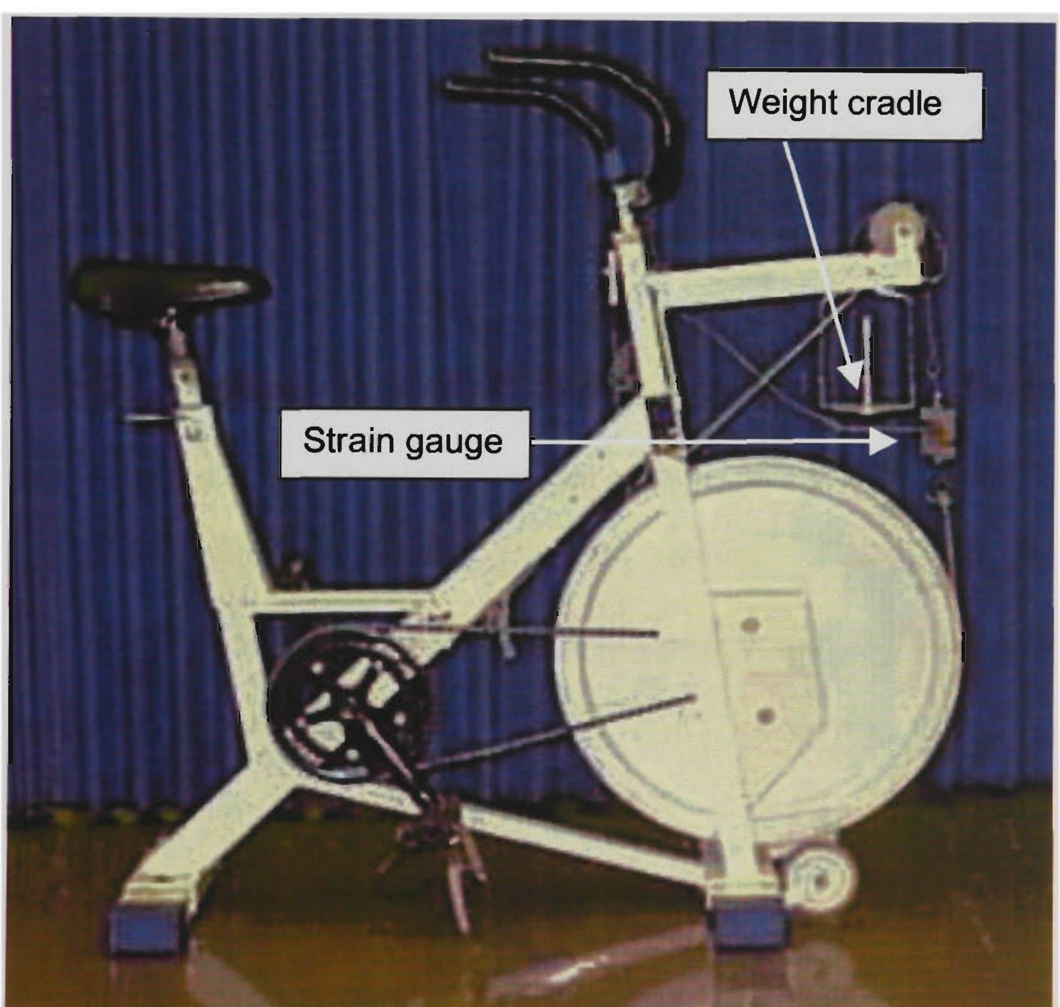


Figure 5.1 Monark ergometer (Model 864)

The ergometer was connected to the calibration system via a mechanical coupling designed by REPCO. Cadence was measured using the optical sensor from the electronically-braked ergometer. Power input to the ergometer was calculated from the variables of force, distance and output shaft revolutions per minute (RPM), as shown in the following formulae.

$$Power = \frac{Work}{Time}$$

$$Work = Force \times Displacement$$

$$Force = Mass \times Gravity$$

$$\therefore Power = \frac{Mass \times Gravity \times Displacement}{Time}$$

$$Power(watts) = \frac{Mass(kg) \times 9.807 \times 2\pi \times 0.6366 \times RPM)}{60}$$

Note: The value 0.6366 is the distance between the point of measurement of force and the output shaft. This equates to a circumference of 4 metres which is the distance travelled in one revolution.

The range of weights added to the weight cradle was 1 to 5 kg in 1-kg increments. Cadence was varied from 58.5 to 117.5 RPM in seven evenly -spaced increments. The calibration system delivered a power input to the ergometer of between 67 and 565 watts depending on the weight and cadence combination selected. The ergometer was retested to determine the test-retest reliability.

5.2.2 Calibration of Air-Braked Ergometers

The ergometers tested were a REPCO air-braked ergometer, modified to include derailleur gearing, and a locally manufactured air-braked ergometer used by the Victorian Institute of Sport (VIS; Figure 5.2).

Figure 5.2 A locally manufactured air-braked ergometer used by the Victorian Institute of Sport

The air-braked ergometers were connected to the calibration system using a coupling link designed by REPCO. Cadence was measured using the optical sensor from the electronically-braked ergometer. Power input to the ergometer was calculated from the variables of force, distance and output shaft RPM (cadence) as shown in the formulae in Section 5.2.1. Temperature was measured, barometric pressure was obtained and both were recorded prior to each calibration session. Power output data calculated from regression equations was corrected for ambient conditions to recalculate the data corresponding to 22° Celsius and 760 mm Hg (Craig & Walsh, 1989). Both ergometers were retested to determine the test-retest reliability.

5.2.2.1 REPCO Air-Braked Ergometer

The range of cadence was 59 to 119 RPM in seven evenly-spaced increments giving a wheel speed of between 360 and 1178 RPM. The ergometer was tested over the complete range of gears available, from 48:13 to 40:19. The power input delivered to this ergometer ranged between 31 and 921 watts depending on the cadence and gearing selected. This range of cadence and gearing was also used to test the ergometer when placed close to a fixed object (wall). Separately, the air-braked ergometer was also calibrated when a cooling fan was placed front-on.

5.2.2.2 Victorian Institute of Sport Prototype Ergometer

This ergometer was tested over the complete range of gears 44:13 to 35:21 and a cadence range of from 59 to 119 RPM in seven evenly-spaced increments producing a wheel speed of between 339 and 1055 RPM. Power input to this ergometer varied between 38 and 921 watts depending on the gearing ratio and cadence.

5.3 RESULTS AND DISCUSSION

5.3.1 Accuracy and Reliability of the Mechanically- and Air-Braked Bicycle Ergometers

Both mechanical and air-braked ergometers are assumed accurate and reliable, thus making them suitable for testing and re-testing elite cyclists. Researchers have questioned the accuracy and reliability and at least one manufacturer (Monark) has vigorously defended the criticisms. (Harman, 1989; Maxwell et al, 1998; Ohlsson, 1989; Telford et al, 1980; Wilmore et al, 1982 and Zommers et al, 1995a).

A statistical analysis of the present data found that the best regression equation for both air-braked ergometers was a cubic polynomial without cross-product terms. Using a linear regression in the first instance produced large maximum errors at the low and high powers. This indicated that a polynomial model may be more appropriate.

Mean differences, expressed as a percentage of variation from the calibrated power, varied considerably between the systems with the largest differences obtained in the Monark system. The use of a strain gauge to measure the frictional forces on the Monark ergometer improved the accuracy of the Monark.

After an initial calibration of the modified REPCO ergometer, a mean difference of 2.01% between the predicted and measured power was obtained. However the largest error encountered for the REPCO over the range of power of 31 to 921 watts was considerably less than for the Monark bike. Errors on the REPCO increased with the introduction of a cooling fan in close proximity (1.5 metres) to where the test subject was seated and errors also increased when the ergometer was placed front-on and in close proximity to a wall. **Note:** In normal usage of the air-braked ergometer, cooling fans, personnel and equipment such as computers are in close proximity. The present data suggests that all of these may impair the accuracy of the air-braked ergometer and nearby walls will compound this problem.

Test-retest reliability ranged from 1.13% to 2.30%.

Difference %[2]	weight cradle[1] Cal 1[3]	cal 2[4]	Retest[5]	strain gauge measurement Cal 1	cal 2	retest
Mean	5.74	4.63	2.14	3.66	2.0	2.3
Max	15.10	16.45	10.34	6.44	3.8	8.3
±SD	3.73	3.54	2.05	1.54	1.03	1.93
n	36	36	36	36	36	36

Table 5.1 Mechanically-braked ergometer calibration data

Difference %[2]	initial cal[7]	Cal 2	Retest
Mean	2.9	2.9	1.13
Max	9.13	7.7	4.71
±SD	1.8	2.13	1.0
n	74	74	74

Table 5.2 VIS air-braked ergometer calibration data

	Initial cal	Retest	Proximity[8]	cooling fan
Mean	2.01	1.68	2.6	4.71
Max	4.78	9.3	14.34	7.97
±SD	0.97	1.71	2.5	1.98
n	88	89	88	13

Table 5.3 Modified REPCO air-braked ergometer calibration data

Notes:

1. The power measurement uses the variables of force, flywheel circumference and pedalling cadence.
2. The difference between calibrated power and the power estimated by the ergometer were calculated as a percentage of variation from the calibrated power. Any negative values were converted before calculating the mean, max and SD.

3,4 The figures shown in these columns are the descriptive statistics of the percentage differences between calibrated power and measured power on the ergometer during each of the calibration tests.

5. The retest columns show the differences between two tests carried out on the same day.
6. Difference between predicted power and calibrated power expressed as a percentage of calibrated power
7. Difference between predicted power and calibrated power expressed as a percentage of calibrated power.
8. The air-braked ergometer was initially tested one metre from the wall of the laboratory. It was then moved 1.5 metres further from the wall and a retest carried out to assess the effect of objects in close proximity to the front of the ergometer.

A recent study, in our laboratory, of the effects of an exercise training intervention in post-menopausal women exposed the lack of suitability of the Monark ergometer in studies where the training effect is small, relative to the measurement error.

In conclusion, the above errors may be acceptable for relative testing, although the test-retest reliability is of some concern. If absolute rather than relative tests are to be carried out (e.g.. biological efficiency or inter-laboratory testing on a single subject), then even a 2% error may contribute significantly to the variability of the collected data.

5.3.2 Comparison of Air- and Mechanically-Braked Bicycle Ergometers With the Author's Electronically-Braked Ergometry System

The residual data obtained from the regression of the final calibration of the electronically-braked ergometry system was converted to positive values. The mean of this data was then expressed as a percentage of the mean power input from the calibration system. The percentage error, from this calculation was 1.16%. As outliers were not removed some residuals were quite large as can be seen from the graphs in Chapter 4. Nevertheless, an error of 1.16% is considerably smaller than the mean errors obtained from the air- and mechanically-braked bicycle ergometers especially the mechanically-braked ergometer using the original weight cradle concept.

The accuracy and test - retest reliability of the mechanically-braked ergometer suffers as a result of variability in the frictional forces (Maxwell et al, 1988; Harman, 1989; Telford et al, 1980). The air-braked ergometers' accuracy and test - retest reliability are considerably better but nevertheless are affected by environmental factors (e.g. barometric pressure), airflow factors (see Section 5.3.1 above) and consistency in the resistance offered by the drive train. Provided the barometric pressure is measured and included in the calculation of power output, and the ergometer is maintained and used properly, this source of variability should be of minor concern. As a retest of the electronically-braked ergometry system was not carried out then a test-retest comparison of this system can not be made. It is not affected by variations in environmental factors and variability in the frictional forces between the bicycle rear wheel and the alternator drive roller was measured and included in the calculation of power output. From the design point of view, and as the only known source of variability had been accounted for, the electronically-braked ergometry system is inherently superior to the air and mechanically-braked ergometers.

CHAPTER 6

PILOT STUDY TO TEST THE OPERATION OF THE ELECTRONICALLY-BRAKED ERGOMETRY SYSTEM

6.1 INTRODUCTION

The objective of this pilot study was to perform an operational test under conditions similar to those that would be encountered in the main study in this thesis. This enabled any unexpected problems with the hardware and software to be solved prior to the main study.

Although this pilot study was conducted primarily as a means of testing the equipment prior to use in the main study it did reveal some important data relative to that obtained in the main study.

This study revealed very early that the frame used to hold the bicycle needed modification to allow volunteers with large shoes to pedal without their shoes hitting an obstruction. After this modification was carried out the ergometry system performed its function flawlessly. The positive feedback from the adolescent cyclists reassured the author that the correct decision had been made at the outset to construct an ergometry system that could utilise the athletes' own bicycles. Although the ergometry system does not have a flywheel to fully simulate road conditions, the athletes did not desire to stop pedalling and "coast" as is possible on the road. The cyclists, at times, did change to a lower gear and/or slow their cadence as is possible when riding in echelon on the road. After observing this practice in the pilot study the instructions to the volunteers, for the main study, were subsequently modified to ensure that athletes would choose a workload that they could maintain for the duration of the test but still work just below their anaerobic threshold. The design of the system did not allow the cyclists the option of ceasing pedalling if they so chose.

6.2 METHODS

The volunteers for this study were male (n = 18) and female (n = 16) endurance-trained adolescents (13 to 17 years). The study required the athletes to ride on their own bicycles using the electronically-braked ergometry system (Figure 6.1).

Figure 6.1 Adolescent cyclist being tested on the author's ergometry system

The protocols for the cyclists under and over 15 years were that they ride 4 km and 8 km, respectively. O_2 uptake (VO_2) was measured at regular intervals during the exercise and from this and the mechanical power measured by the system, biological efficiency was calculated. This was then compared to corresponding data in adults, as there is no comparative data published for adolescents. Athletes started at a self-selected mechanical resistance and cadence. If they were unable to maintain power then they were entitled to select a lower gear ratio or cadence in order to decrease power.

6.3 RESULTS AND DISCUSSION

The male and female adolescent volunteers of this study had a mean age of 14.3 and 14.4 years, respectively. The maximal oxygen uptakes (Table 6.1) of the adolescent male, although not significantly different ($t = 2.00$, $p = 0.054$) tended to be higher than for the females. There was no significant difference in the levels of self-selected exercise intensity during the performance of the duathlon for the males and females when expressed as a percentage of $VO_{2\ max}$ (83.9% and 82.0%, respectively; $t = 0.97$, $p = 0.34$).

	$\dot{V}O_{2\,max}$ (ml.Kg^{-1}.min^{-1})		Cycling test $\dot{V}O_2$ (ml.Kg^{-1}.min^{-1})	
	Mean	**±SD**	**Mean**	**±SD**
Female	55	6	44	5
Male	60	9	53	8

Table 6.1 Maximal effort and cycling performance VO_2 data.

The performance heart rate data showed that the males and females exercised at a level of 88.6% and 90.0% of their maximal heart rates, respectively. The mean power output during the ride was 187 and 140 watts for the males and females, respectively. Table 6.2 presents the data for the net efficiency of the two groups during the ride component of the duathlon. The data are comparable to previous efficiency values reported in the literature on adults. (22% ± 4%; Andersen et al., 1971; Rowland et al, 1990; and Springer et al, 1991). Statistical analysis of the net biological efficiencies revealed that there were no significant differences between males and females ($t = -0.13$, $p = 0.90$). The performance VO_2 and heart rates were both below the ventilatory thresholds, which were determined from the prior incremental $VO_{2\ max}$ test. These data indicate that the participant's freely chosen work effort in the performance was predominantly aerobic in nature.

	FEMALE		MALE	
	Mean	**±SD**	**Mean**	**±SD**
Efficiency %	20.44	4.51	20.61	2.20

Table 6.2 Biological efficiency (%).

In conclusion the data for the riding "leg" of the simulated duathlon reflects the distances for this group of adolescent male and females are quite well performed within their aerobic limitations.

As the cyclists were simulating racing conditions then this study provided important data (efficiency%) that could be compared with the adult athletes in the main study and used to validate the measurement of the electronically-braked ergometry system.

CHAPTER 7

EFFICIENCY OF ANKLING AND NORMAL PEDALLING

7.1 INTRODUCTION

The preceding chapters focused on the development of the author's ergometer (Chapter 4), a comparison of the accuracy and reliability of the ergometer in comparison with other ergometers in usage by sports science and research facilities (Chapter 5) and the trialing of the ergometer in the collection of physiological data (Chapter 6).

The present chapter is the essential chapter of the thesis and will address the central hypothesis that ankling improves biological efficiency.

7.2 METHODS

7.2.1 Participants

Forty-five volunteers were recruited for this study from varying cycling disciplines and backgrounds. As this study was focusing on a pedalling technique that could be of benefit to competitive cyclists, if adopted, only athletes who competed on a regular basis were included. Cycling disciplines included the traditional road and track racing as well as the relatively new disciplines of triathlon and mountain bike racing. Many cyclists participated in more than one discipline. Cycling ability ranged from the club athlete to the elite with one being a final contender for the 1996 Olympic games (Appendix R) and several others who had recently competed at World championship level. It would have been ideal, from a statistical sampling error perspective, to narrow the selection to only one discipline and only use elite cyclists. However, this would not permit generalisation of the results to other cycling disciplines and other levels of cycling ability. In order to minimise sampling error, the sample size used was as large as possible within the time and financial constraints of the study.

7.2.2 Methods (refer to the following table)

	Test sequence					
	1	2	3	4	5	6
TEST DESCRIPTION			**Pre**	**$Post_1$**	**$Post_2$**	**$Post_3$**
$VO_{2\,max}$ test (all athletes), refer to Section 3.6. The data from this test was used to determine the starting point for the following test (i.e. 50% of $VO_{2\,max}$)	—					
Anaerobic threshold (AE_{th}) test (all athletes), refer to Section 3.8. The data from this test was used to assist in determining the workload for the following tests.		—				
Efficiency test (Practice group; see Section 3.5) The pre-training test and post-training tests ($Post_1$, $Post_2$ and $Post_3$) were done one month apart, with training in between. Data: VO_2, mechanical power and video film. Biological efficiency was calculated.			—	—	—	—
Efficiency test (Control group; see Section 3.5) The pre-training test and post-training tests ($Post_1$, $Post_2$ and $Post_3$) were done one month apart. Data collected was the same as for the practice group.			—	—	—	—

Table 7.1 Methods used in obtaining biological efficiency and pedalling effectiveness data

Of the 45 volunteers recruited, 42 remained in the study after the VO_2 max test and 41 after the anaerobic threshold testing. Of the 42 athletes who were tested for the anaerobic threshold, 34 underwent lactate threshold tests, a further eight completing ventilatory threshold tests (see Section 3.8.2) The data collected in this phase of the study was analysed to locate the anaerobic threshold (see Section 3.8). The mechanical power output corresponding to 95% of this threshold was then used as an aid in setting the workload for the efficiency tests. On the first efficiency test the power output was only varied from this predetermined level if the volunteers indicated that they could not maintain the power output for the duration of the efficiency test (20 minutes). The power output was also adjusted if the RER was over 1.0 or below 0.9 at the start of the test. During subsequent efficiency tests, the cyclist's power output was identical to the previous test unless the cyclist indicated that they were unable to maintain that power. Thirty-three athletes completed all phases of the efficiency tests and a further eight athletes had incomplete data from the four efficiency tests due to illness, equipment failure or data corruption. Cyclists with missing data were censored from the data analysis.

Four efficiency tests were conducted over a three-month period (Section 3.5). Data collected was VO_2, respiratory exchange ratio, mechanical power output and angular kinematics of one lower limb while pedalling.

A full description of the efficiency calculations is found in Section 3.11.2. To quantify the ankling technique, the angle between the foot and the crank for the pedalling range of 0° and 180° of the crank cycle, top dead centre (TDC) to bottom dead centre (BDC), was measured. The mean of this data was subsequently calculated. These data were analysed using a spreadsheet and SPSS software.

7.3 RESULTS

7.3.1 $VO_{2\,max}$

There was a significant difference in the $VO_{2\,max}$ between males (n=31) and females (n=14; $t = 4.04$, $p = 0.000$;Table 7.2), as was expected, but no significant difference in $VO_{2\,max}$ between the control and experimental groups ($t = 0.16$, $p = 0.87$) nor the two age groups ($t = 0.82$, $p = 0.42$). A test of homogeneity also showed that the groups were of equal cycling fitness ($F(7,33) = 1.14$, $p = 0.36$).

	Male	Female	Under 35	Over 35	Control	Experimental
Mean	63.2*	53.7*	61.3	59.0	60.2	60.6
±SD	7.2	7.6	9.5	8.0	8.5	9.5
Max	81.0	70.7	81.0	72.2	72.2	81.0
Min	49.8	44.3	44.3	46.4	46.4	44.3
n	31	14	25	16	20	20

Table 7.2 $VO_{2\ max}$ statistical data. The Max and Min values represent the range of $VO_{2\ max}$ for each column

* significant difference between males and females ($p = 0.000$)

7.3.2 Anaerobic Threshold Tests

In the current study, individual lactate thresholds varied between 2.0 mmol.L^{-1} and 7.1 mmol.L^{-1}. Ventilatory thresholds ranged from 43.0 ml.kg^{-1}.L^{-1} to 62.5 ml.kg^{-1}.L^{-1}. The anaerobic threshold, expressed as a percentage of $VO_{2\ max}$, varied from 75% to 95%.

7.3.3 Efficiency Tests

7.3.3.1 Mean VO_2 for the Efficiency Tests Relative to $VO_{2\ max}$

The mean VO_2, of the efficiency tests, expressed as a percentage of $VO_{2\ max}$, was 75% with a minimum of 60% and a maximum of 94%. There were no significant differences between both males and females ($t = -0.82$, $p = 0.42$), the two age groups ($t = -0.70$, $p = 0.49$) and the two technique groups ($t = -0.19$, $p = 0.85$; Table 7.3).

	Male	Female	Under 35	Over 35	Experimental	Control
Mean	74.1%	76.4%	74.0%	75.9%	74.5%	75.0%
±SD	8.3%	7.6%	8.3%	7.8%	8.7%	7.5%
Max	93.8%	86.9%	93.8%	87.9%	93.8%	87.9%
Min	59.9%	64.8%	59.9%	61.4%	59.9%	60.8%
n	29	12	25	16	21	20

Table 7.3 VO_2 as a percentage of $VO_{2\,max}$ during the efficiency tests. Each statistic represents the mean data for all of the efficiency tests.

7.3.3.2 Power Output, During the Efficiency Tests, Relative to the Anaerobic Threshold

Mechanical power output was significantly lower than that corresponding to the anaerobic threshold ($t = 6.02$, $p = 0.000$) and that corresponding to 95% of the anaerobic threshold ($t = 3.68$, $p = 0.001$; Table 7.4). Mechanical power output measured during the 20 minute efficiency tests correlated significantly with the power determined previously at the anaerobic threshold ($r = 0.87$, $p = 0.000$; Table 7.4).

	Mechanical power output (watts)		
	Efficiency tests	**95% Anaerobic threshold**	**Anaerobic threshold**
Mean	208.8*#	224.3#	236.3*
±SD	40.3	53.1	55.8
Max	307.0	355.0	375.0
Min	121.9	135.0	142.0
n	41	41	41

Table 7.4 Comparison of mechanical power output during the efficiency tests and the mechanical power output at 95% of the anaerobic threshold and the anaerobic threshold. The mean, max and min mechanical power output values for the efficiency tests were derived from the mean power for each efficiency tests for the 41 cyclists.

* significant difference between mechanical power during the efficiency tests and the mechanical power at the anaerobic threshold (p = 0.000)

significant difference between mechanical power during the efficiency tests and the mechanical power at 95% of the anaerobic threshold (p = 0.001)

The mechanical power output also correlated significantly when subdivided (partial correlation) according to group (experimental and control; r = 0.87, p = 0.000; Table 7.5).

	Mechanical power output (watts)			
	efficiency tests		Anaerobic threshold	
Groups	Experimental	Control	Experimental	Control
Mean	208*	210*	232*	240*
±SD	47	32	60	52
Max	307	278	375	320
Min	122	141	142	155
n	21	20	21	20

Table 7.5 Comparison of mechanical power output during the efficiency tests and the mechanical power output at the anaerobic threshold subdivided according to groups (experimental and control)

* Significant correlation between mechanical power output during the efficiency tests and the mechanical power output at the anaerobic threshold subdivided (partial correlation) according to groups (experimental and control)

Similarly a significant partial correlation was found when the data was subdivided according to gender (r = 0.78, p = 0.000), age (r = 0.87, p = 0.000) and type of anaerobic test (r = 0.87, p = 0.000; Appendix S).

Mechanical power output during the efficiency tests was calculated as a percentage of the measured anaerobic threshold. The volunteers during these tests exercised at 91 ±8% (experimental group) and 89 ±10% (control group) of the anaerobic threshold. There was no significant difference between the two technique groups (experimental and control; t = 0.52, p = 0.61; Table 7.6).

	Experimental	Control
Mean	91%	89%
±SD	8%	10%
Max	100%	106%
Min	70%	68%
n	21	20

Table 7.6 Mechanical power output during the efficiency tests expressed as a percentage of the power output measured at the anaerobic threshold

There was a significant difference between males and females (t = 2.27, p = 0.03) but no significant difference between the two age groups (t = 0.42, p = 0.68) and between the two types of anaerobic threshold tests - lactate and ventilatory (t = 1.15, p = 0.26; Appendix T)

7.3.3.3 Respiratory Exchange Ratio (RER)

RER exceeded 1.0 in only nine out of 155 efficiency tests (Table 7.7).

	RER
Mean	0.96
±SD	0.02
Max	1.02
Min	0.92
n	41

Table 7.7 Respiratory exchange ratio (RER) descriptive statistics

There was no significant difference in RER between the two pedalling technique groups ($t = 0.55$, $p = 0.59$; Table 7.8).

	Pedalling technique	
	Ankling	**Normal**
Mean	0.96	0.96
S	0.02	0.03
Max	0.99	1.01
Min	0.92	0.92
n	21	20

Table 7.8 Comparison of the RER between the two pedalling technique groups

There also was no significant difference in RER between males and females ($t = 0.17$, $p = .86$) but there was a significant difference between those under and over 35 years ($t = -2.25$, $p = 0.03$) and the two types of anaerobic threshold tests ($t = 2.82$, $p = 0.008$; Appendix U).

7.3.3.4 Cycling Efficiency

The test of within subject effects, for each group, showed a significant difference in biological efficiency (corrected for O_2 drift) between the two pedalling techniques ($F_{(1,32)} = 25.21$, $p = 0.000$) with the efficiency for ankling being less than for normal pedalling (ankling - 22.2 ± 2.3%, normal - 22.5 ± 2.3%; Figures 7.1a & 7.1b). The degrees of freedom were corrected as the assumption of sphericity was violated (Mauchly's $W = 1.0$, $p = 0.000$; Huynh-Feldt Epsilon = 1.0). There was no significant difference in biological efficiency over time (i.e. the four repeated measures efficiency tests; $F_{(3,96)} = 0.461$, $p = 0.71$; Figures 7.1a & 7.1b). The assumptions of normality, homogeneity of variance and homogeneity of variance-covariance matrices (Box's M statistic not significant, $p = 0.165$) were not violated.

Test of between subjects' effects showed no significant difference between the ankling (experimental) group who practised ankling between tests and the control group ($F(1,32) = 1.15$, $p = 0.29$).

The differences in biological efficiency decreased between the two pedalling techniques over time for the experimental group but there was not a significant change between the first and last efficiency tests ($t = 0.79$, $p = 0.44$; Figure 7.1a & 7.2). The difference in biological efficiency between the two techniques, in the control group, varied between tests with the difference in efficiency in the third test reducing to almost zero (0.03%) from 0.21% in test one and 0.38% in test two but then increasing to 0.73% in the last (fourth) test (Figure 7.1b).

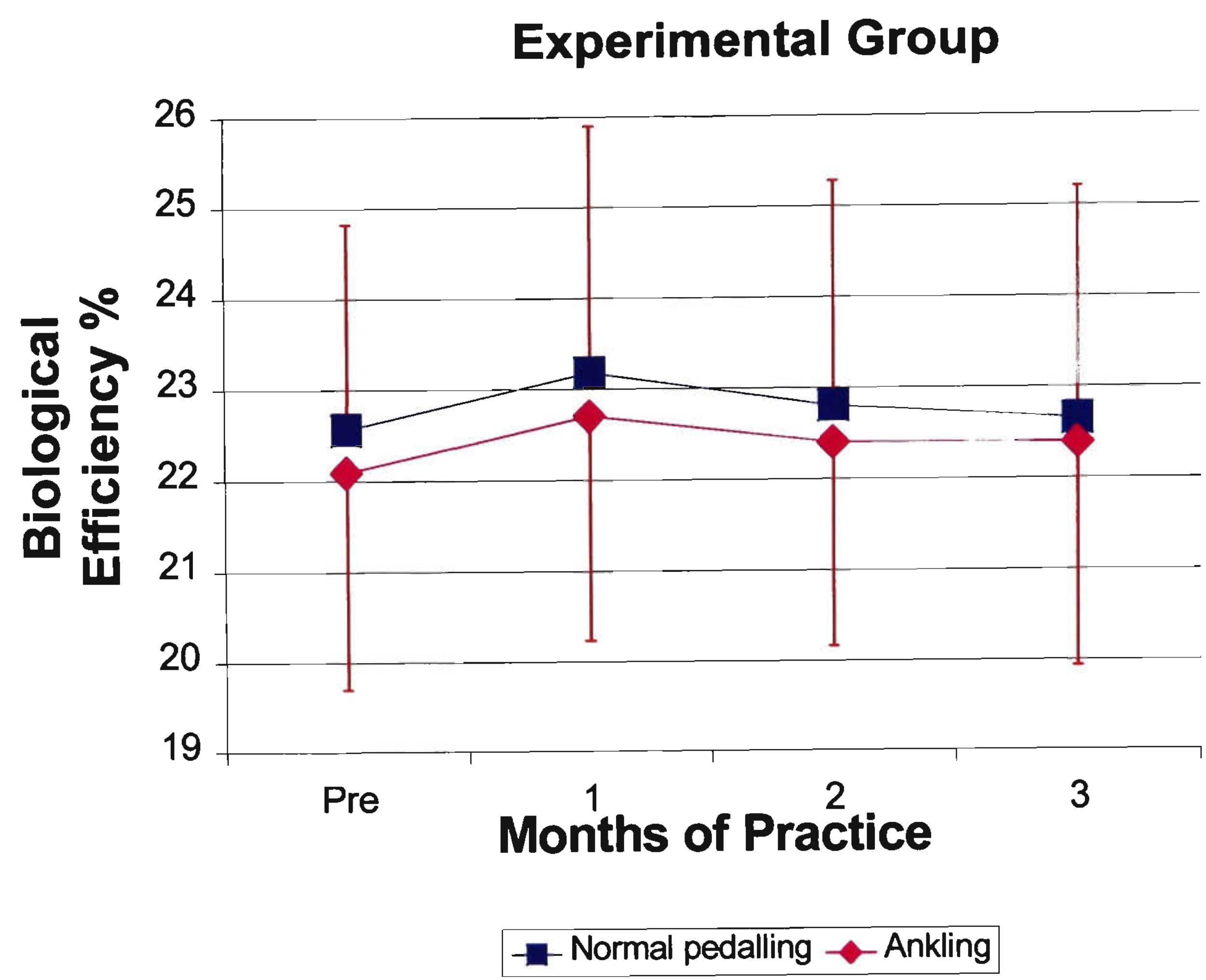


Figure 7.1a Biological efficiency of the experimental group for the four tests comparing efficiency of normal pedalling and ankling (mean, ±SD)

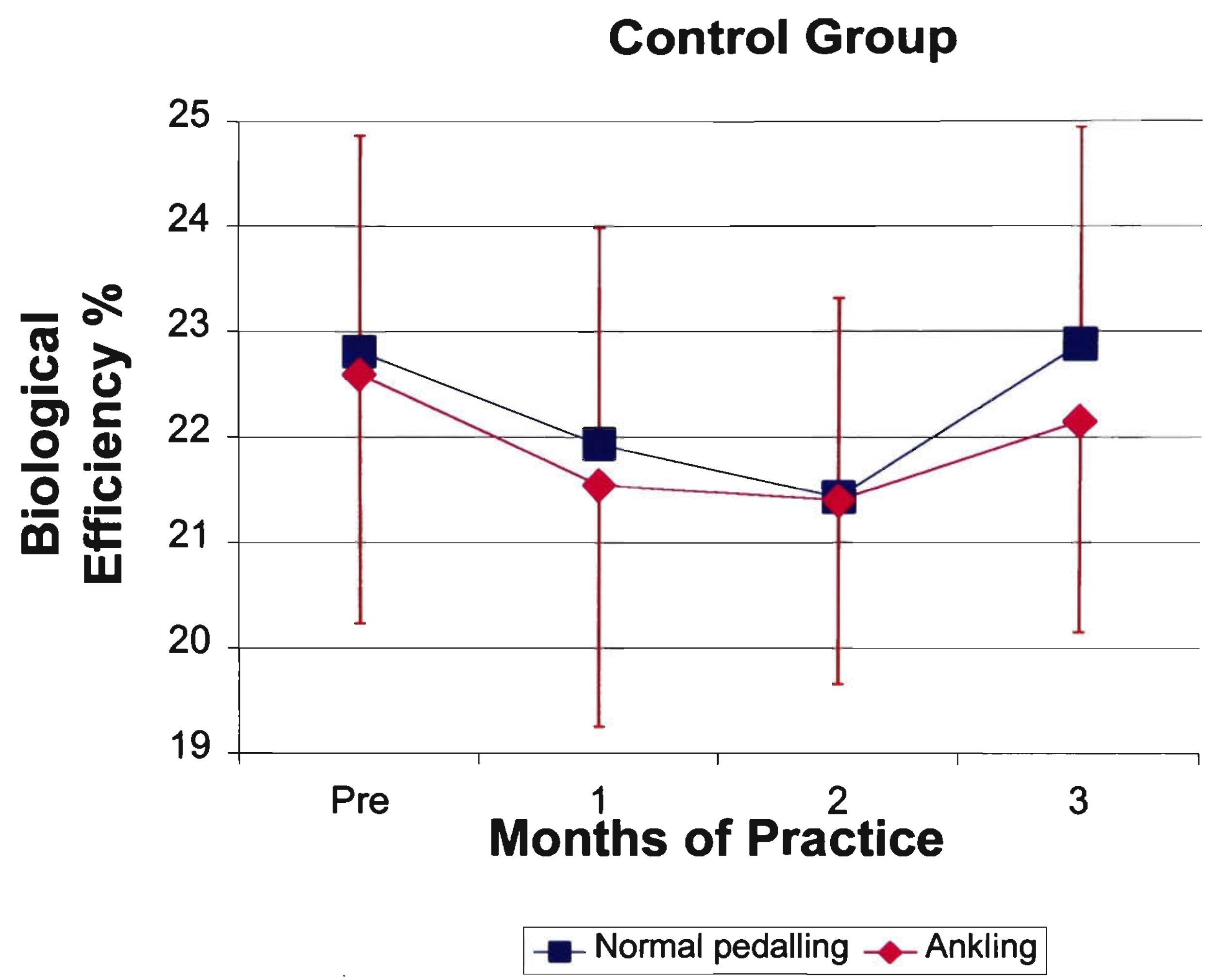


Figure 7.1b Biological efficiency of the control group for the four tests comparing efficiency of normal pedalling and ankling (mean, ±SD)

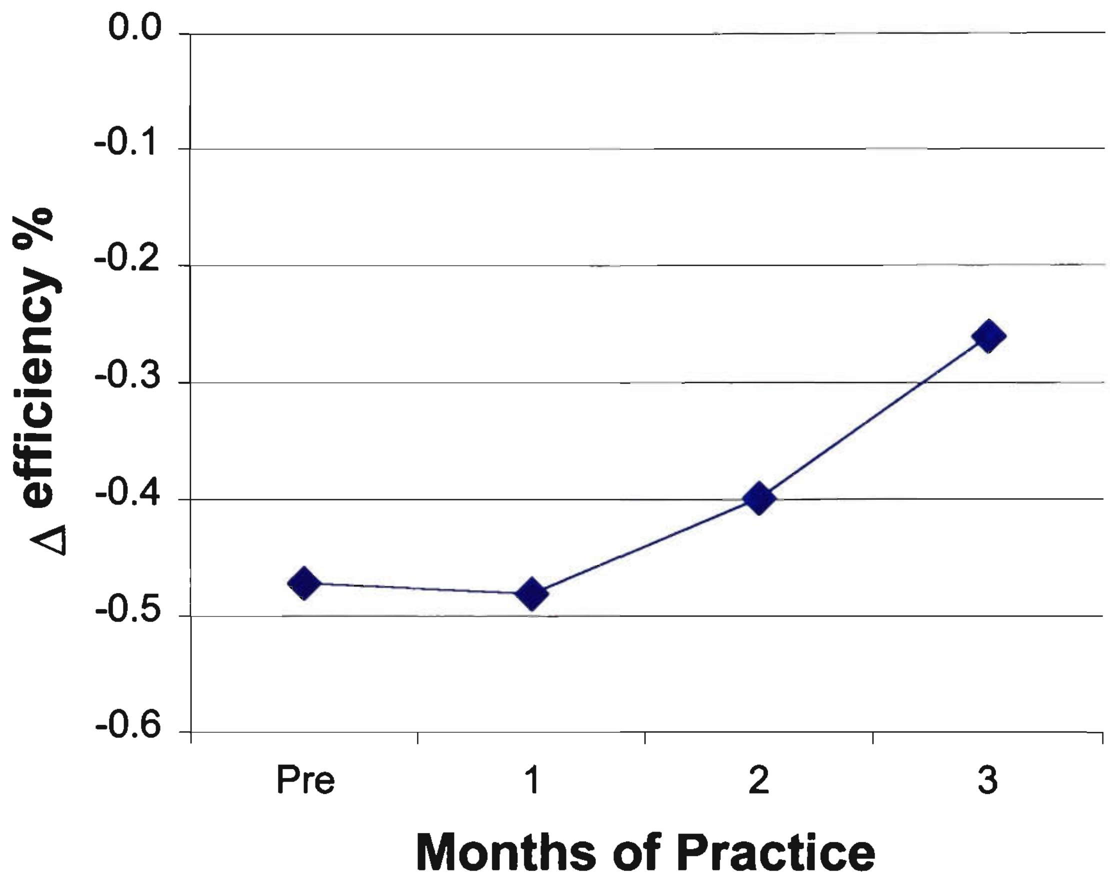


Figure 7.2 Delta efficiency (ankling efficiency - normal efficiency) of the experimental group for the four tests showing the trend in the difference between normal pedalling and ankling

7.3.3.5 Quantifying the Ankling Technique and a Comparison with Normal Pedalling

Kinematic analysis of the lower limbs while pedalling showed that there was a significant difference in the mean angle of the foot relative to the crank (between TDC and BDC) for the two pedalling techniques ($F_{(1,39)} = 8.3$, $p = 0.006$). The difference between the two groups for the normal pedalling technique was approaching significance ($F_{(1,39)} = 2.9$, $p = 0.1$) and there was a significant difference between the two groups for the ankling technique ($F_{(1,39)} = 5.6$, $p = 0.02$; Table 7.9a & b). Observation of the digitised video data showed progressively less variability in the movement of the ankle joint, in all athletes, as the study progressed.

	Ankling technique	**Normal pedalling**
Mean	52°*	54°*
±SD	3.8	1.9
Max	62°	58°
Min	46°	49°
n	41	41

Table 7.9a The mean angle between the foot and the crank for 0° to 180° (TDC to BDC) for all cyclists for ankling and normal pedalling techniques

* Significant difference between the two pedalling techniques ($p = 0.006$)

	Ankling Technique		Normal pedalling	
	Experimental	Control	Experimental	Control
Mean	51°**	54°**	53°	54°
±SD	3.2	3.9	1.8	1.9
Max	60	62	57	58
Min	46	48	49	50
n	21	20	21	20

Table 7.9b The mean angle between the foot and the crank for 0° to 180° (TDC to BDC) for the experimental and control groups for both pedalling techniques

** Significant difference between the experimental and control groups for the ankling technique ($p = 0.02$)

7.3.4 Data additional to the central hypothesis

7.3.4.1 Angle of Crank where Maximum Power Exhibited

The mean crank angle where maximum power was exhibited was 138° ±9.6°. There was a statistical significant difference in the crank angle where maximum power was measured between the two pedalling techniques ($t = -2.37$, $p = 0.02$; Table 7.10).

	Normal Pedalling	**Ankling**
Mean	137.2°#	138.2°#
±SD	9.3°	9.9°
Max	165.0°	176.0°
Min	112.0°	111.0°
n	304	304

Table 7.10 Angle of crank where maximum power exhibited

Significant difference (p = 0.02)

7.3.4.2 Range of Motion (ROM) of the Ankle and Knee Joints

There were significant differences between the two pedalling techniques for the maximum angle of plantarflexion ($t = 12.3$, $p = 0.000$), the maximum angle of dorsiflexion ($t = -3.1$, $p = 0.003$) and the range of motion (ROM) of the ankle joint ($t = -15.6$, $p = 0.000$). All athletes demonstrated a large ROM of the ankle joint while ankling (mean ± SD = 46.4° ± 9.3°) with the minimum being 25.3° and the maximum 65.2°. There was no significant difference in the ROM of the knee joint for the two pedalling techniques (Table 7.11). The data presented in the Tables 7.11a to 7.11d is the combined data for the two groups comparing the two techniques.

	Ankling	Normal Pedalling
Mean	27.4° *	11.3° *
±SD	8.2°	4.2°
Max	-50.5°	-21.9°
Min	-12.2°	-1.8°
n	41	41

Table 7.11a Maximum angle of plantarflexion for the two pedalling techniques

* Significant difference (p = 0.000)

	Ankling	Normal Pedalling
Mean	19.0° *	16.6° *
±SD	7.8°	6.2°
Max	35.1°	26.6°
Min	0.6°	1.5°
n	41	41

Table 7.11b Maximum angle of dorsiflexion for the two pedalling techniques

* Significant difference (p = 0.003)

	Ankling	Normal Pedalling
Mean	46.4° *	27.9° *
±SD	9.3°	5.3°
Max	65.2°	37.6°
Min	25.3°	14.3°
n	41	41

Table 7.11c Range of motion (ROM) of the ankle for the two pedalling techniques

* Significant difference (p = 0.000)

	Ankling	Normal Pedalling
Mean	73.3°	74.2°
±SD	5.1°	4.6°
Max	87.7°	82.1°
Min	63.3°	64.0°
n	41	41

Table 7.11d Range of motion (ROM) of the knee for the two pedalling techniques

There were no significant differences in the same data when analysed subdivided according to group (experimental and control; Table 7.12).

	Ankling		Normal Pedalling	
	Experimental	Control	Experimental	Control
Mean	-28.4°	-26.4°	-12.0°	-10.5°
±SD	6.0°	10.1°	4.4°	3.4°
Max	-41.0°	-50.5°	-21.9°	-19.2°
Min	-19.6°	-12.2°	-1.8°	-3.3°
n	21	20	21	20
	$t = -0.74$, $p = 0.47$		$t = -1.11$, $p = 0.27$	

Table 7.12a Maximum angle of plantarflexion for the two groups and the two pedalling techniques

	Ankling		Normal Pedalling	
	Experimental	Control	Experimental	Control
Mean	18.5°	19.5°	16.0°	17.3°
±SD	6.9°	8.8°	6.6°	5.9°
Max	28.1°	35.1°	24.7°	26.6°
Min	0.6°	3.2°	1.5°	-3.3°
n	21	20	21	20
	$t = -0.40$, $p = 0.70$		$t = -0.64$, $p = 0.53$	

Table 7.12b Maximum angle of dorsiflexion for the two groups and the two pedalling techniques

	Ankling		Normal Pedalling	
	Experimental	**Control**	**Experimental**	**Control**
Mean	72.8°	73.9°	74.6°	73.7°
±SD	3.5°	6.4°	3.7°	5.5°
Max	79.7°	87.7°	81.8°	82.1°
Min	66.7°	63.3°	68.1°	64.0°
n	21	20	21	20
	t = -0.67, p = 0.51		t = 0.64, p = 0.53	

Table 7.12c ROM of the knee joint for the two groups and the two pedalling techniques

	Ankling		Normal Pedalling	
	Experimental	**Control**	**Experimental**	**Control**
Mean	46.9°	45.9°	28.0°	27.8°
±SD	8.7°	10.1°	5.6°	5.1°
Max	60.4°	65.2°	37.6°	36.8°
Min	32.2°	25.3°	14.3°	15.4°
n	21	20	21	20
	t = 0.33, p = 0.74		t = 0.13, p = 0.90	

Table 7.12d ROM of the ankle joint for the two groups and the two pedalling techniques

7.3.4.3 Cadence Variations

Cadence varied sinusoidally with two complete cycles of a sinewave for each crank revolution (Figure 7.3).

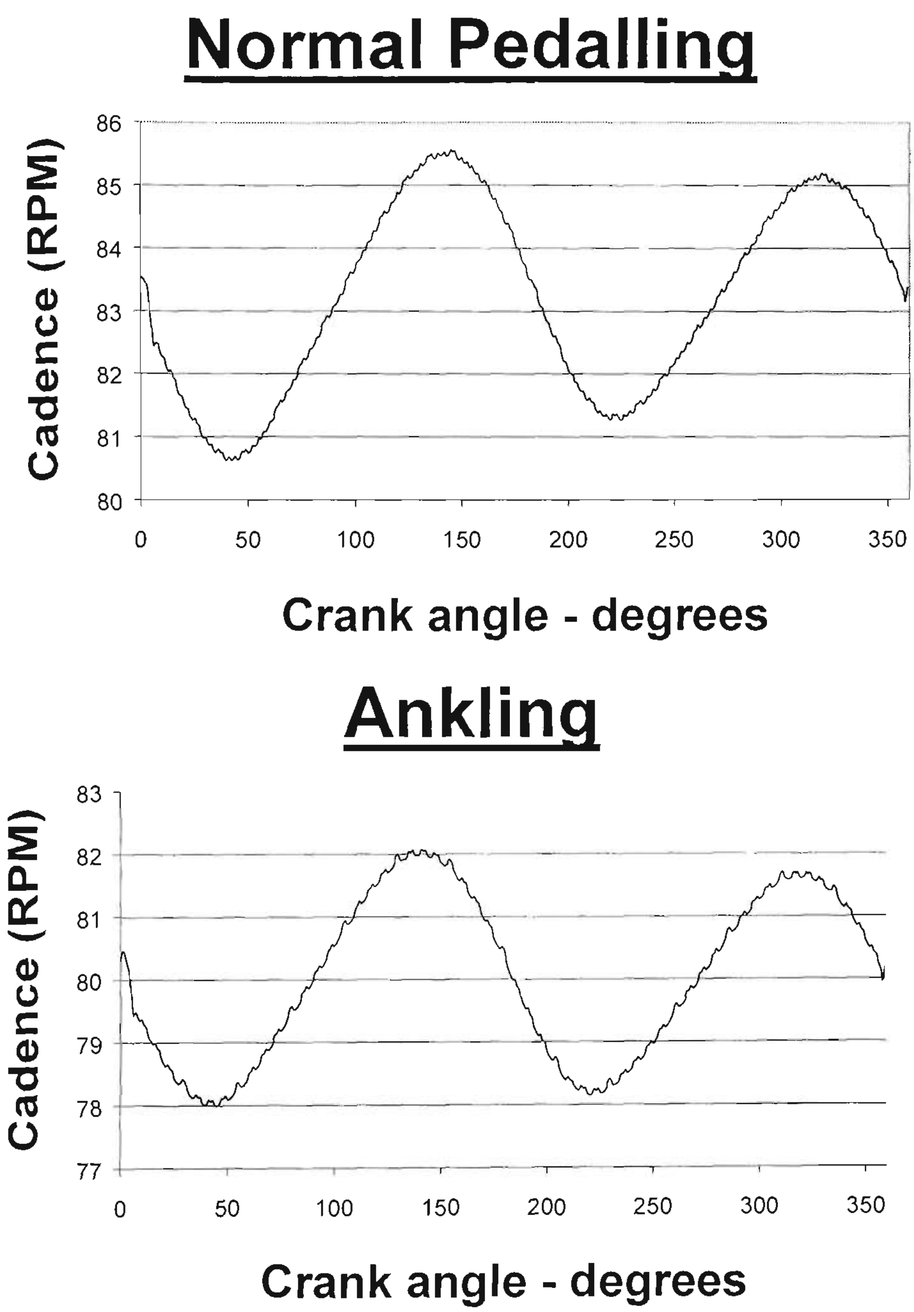


Figure 7.3 Cadence variations were sinusoidal with two complete cycles of a sinewave for each crank revolution (data from one test for cyclist # 4 shown)

7.3.4.4 Knee to Pedal Distance

Knee to pedal distance varied considerably between cyclists with a mean distance (± SD) between the centre of the pedal and the front of the knee (Figure 7.4) of -3.08mm ± 31.48mm. **Note:** A negative value indicates that the knee was behind the centre of the pedal. There was no significant difference between the two groups of volunteers ($t\ (36) = 1.20$, $p = 0.24$; (Table 7.13), the males and females ($t\ (36) = 0.24$, $p = 0.82$; and the under and over 35 years old athletes ($t\ (36) = 0.93$, $p = 0.36$; Appendix V). When this variable was added as a covariable in the analysis of biological efficiency (Section 7.3.3.4) it did not alter the results.

	Experimental	**Control**
Mean	3.00mm	-9.16mm
±SD	31.86mm	30.72mm
Max	85mm	50mm
Min	-45mm	-60mm
n	19	19

Table 7.13 Horizontal distance between a plumb line dropped from the front of the knee to the centre of the pedal with the crank at 90° for the experimental and control groups. A negative value indicates that the front of the knee was behind the pedal.

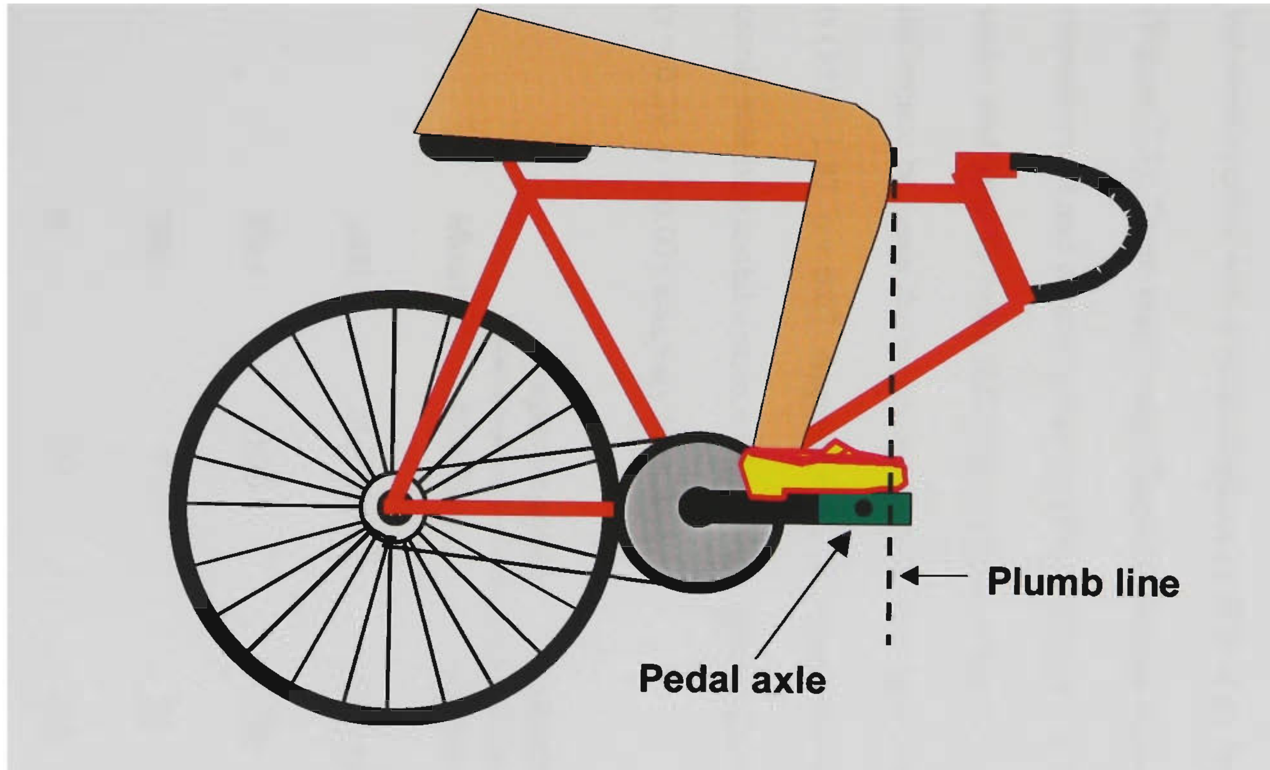


Figure 7.4 Horizontal distance measured between the plumb line dropped from the front of the knee to the centre of the pedal with the crank at 90°

7.3.4.5 Horizontal Distance Between the Bicycle Seat and the Centre of the Crank

The horizontal distance (R) measured using a plumb-line between the front of the bicycle seat and the centre of the crank varied considerably between cyclists with a mean distance (± SD) of 61.2mm ±22.5mm (Figure 7.5). There was no significant difference between the experimental and control groups ($t\ (39) = 0.22$, $p = 0.83$; Table 7.14), the under and over 35 years old cyclists ($t\ (39) = 0.66$, $p = 0.51$; but the difference between the males and females was approaching significance ($t\ (39) = 1.87$, $p = 0.07$; Appendix W). The correlation (r) between the crank-seat horizontal distance (R) and the knee-pedal horizontal distance ($r = 0.29$, $p = 0.07$) was very low.

	Experimental	**Control**
Mean	62.0mm	60.4mm
±SD	24.7mm	20.7mm
Max	100	100
Min	5	25
n	21	20

Table 7.14 Horizontal distance measured using a plumb line between the front of the bicycle seat and the centre of the crank

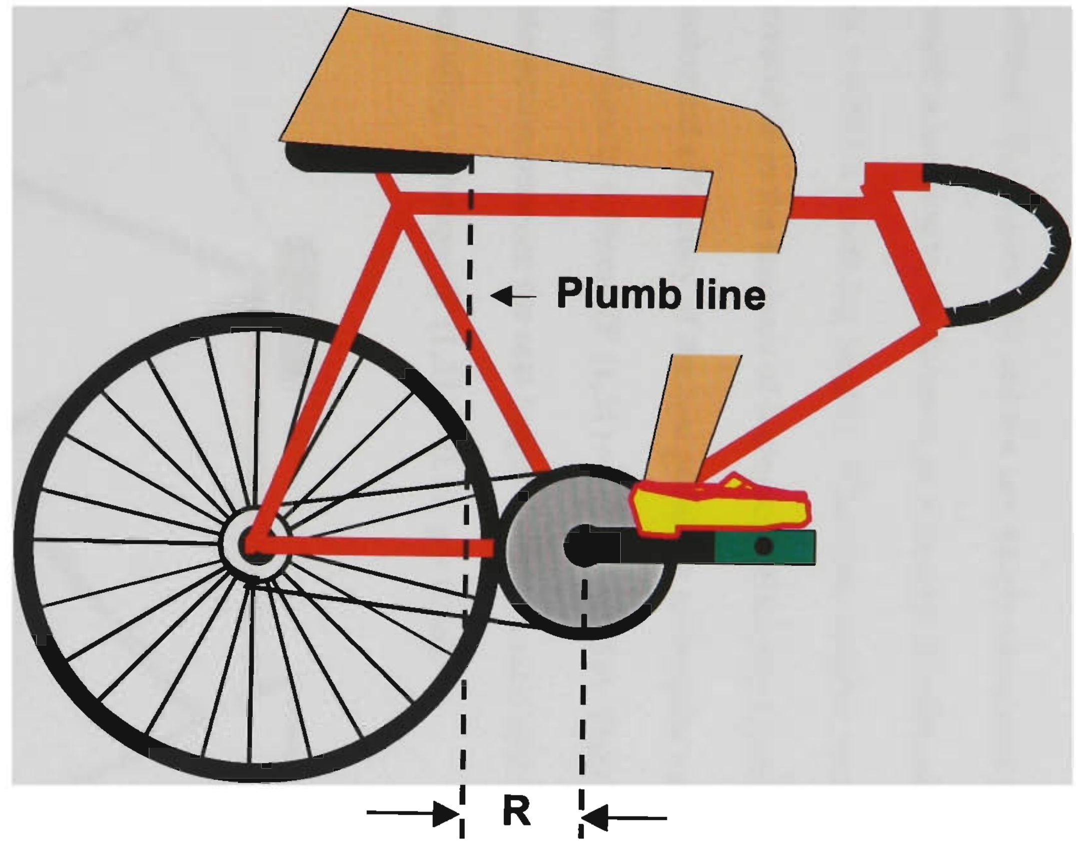


Figure 7.5 Horizontal distance (R) measured between the plumb line dropped from the front of the seat to the centre of the crank

7.3.4.6 Seat Height/Inside leg Length Ratio

The mean (±SD) seat height/inside leg length ratio was 0.87 ± 0.03 with a maximum of 1.03 and a minimum of 0.81. There was no significant difference ($t = 1.08$, $p = 0.28$) between the measured seat height of the athletes (H_S; Figure 7.6) and the seat height calculated from the inside leg length according to the formula provided by Hinault and Genzling (1987; $H_S = 0.855 \times$ inside leg length). When this variable was added as a covariable in the analysis of biological efficiency (Section 7.3.3.4) the biological efficiency of the two pedalling techniques was no longer significantly different ($F(1,31) = 0.39$, $p = 0.54$). There was no interaction between the seat height/leg length ratio and the type of pedalling technique ($F(1,31) = 0.19$, $p = 0.66$).

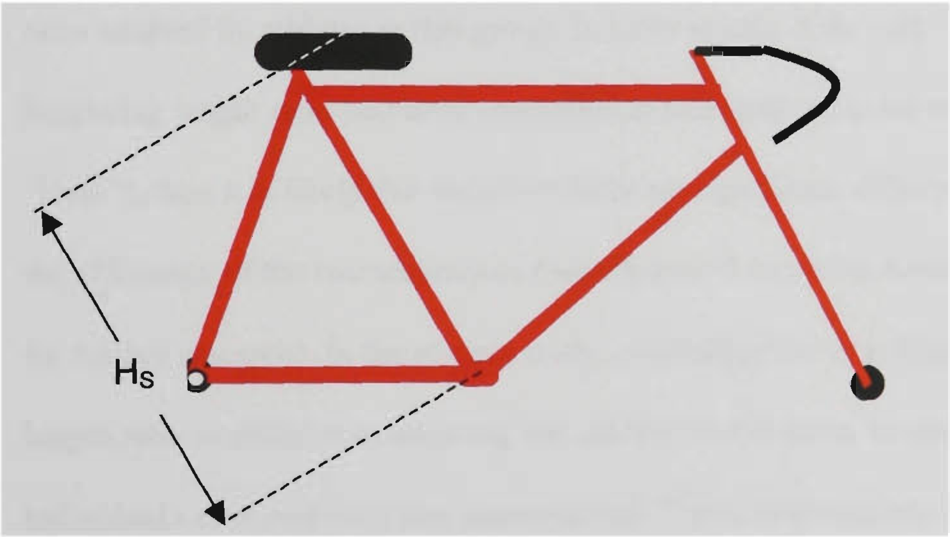


Figure 7.6 Seat height measurement (H_S) measured from the top surface of the centre of the seat to the centre of the crank axle

7.4 DISCUSSION

7.4.1 Biological Efficiency of Experimental and Control Groups for the two Pedalling Techniques

Ankling was significantly less efficient than normal pedalling, which rejects the null hypothesis of the thesis. Furthermore, this data argues against the adoption of ankling as a more effective method of force production than normal pedalling despite its support (Francis, 1986; Hull and Davis, 1981; Sharp, 1896; and Watters, 1987). When the seat height/leg length ratio was added as a covariable, analysis of covariance (ANCOVA) resulted in no significant difference in biological efficiency between the two pedalling techniques. This indicates that the statistical difference in biological efficiency with ankling was not solely attributed to the technique itself, but at least in part to the seat height/leg length ratio adopted by athletes in this group. In other words, if the seat height/leg length ratio had been controlled in this study (e.g. set at "ideal"), then it is likely that there would be no significant difference in the efficiency of the two techniques (see Chapter 9 on recommendations for further research). In the present study, controlling the seat height/leg length ratio in addition to adopting the ankling would have, in many individual cases, imposed two interventions. These interventions would have needed to be imposed sequentially and would have added tremendous complexity to the research design. Finally, the rationale for not controlling for the seat height/leg length ratio was that it was intended to replicate each volunteer's preferred set-up (own bicycles, self-selection of seat position, handlebar height, cadence etc.), with the sole intervention being ankling.

Efficiency studies have previously determined a strong relationship between the seat height/leg length ratio and biological efficiency (Hamley and Thomas, 1967; Nordeen-Snyder, 1977; and Shennum and deVries, 1976). Anecdotal evidence has shown that all cyclists do not necessarily follow the recommendations in the above research studies. The popular cycling literature recommends the use of the ideal seat height/leg length ratio (.855 x inside leg length; Hinault and Genzling, 1987) for setting up individual cyclists, but this advice is not always followed. Others have recommended that the seat height be adjusted from the ideal to suit individual preference. (Faria and Cavanagh, 1978; Hinault and Genzling, 1988a).

Analysis of the difference in efficiency between the two techniques, in the present thesis, for the four tests (repeated measures) found that the mean difference (delta efficiency) was decreasing. Although this decreasing trend in the delta efficiency between the two techniques was not significant, the trend indicates that a longer training intervention may eliminate the gap between the two techniques. However, there is no evidence to suggest that ankling would be more efficient with long-term training.

Although all possible attempts were made to control for variability in the data, variability of the participants is outside of the researchers' control unless selection criteria is very narrow, for example elite cyclists only. Random stratified allocation of volunteers to the experimental and control groups was successful in that homogeneity between the groups was maintained. As the range of cycling ability was quite extensive, with some being low level club athletes to others of Olympic standard, the results of this study can be applied to all competitive cyclists. On the one hand, wide inclusion criteria results in greater inter-subject variability, but on the other hand it offers the advantage in that the results can be more generally applied.

Anecdotal evidence from one volunteer in the control group indicated that he practised the ankling technique on several occasions between the first and second tests to assess the merits of the technique. He was instructed to cease practising the ankling technique until after the end of the study.

Cavanagh and Sanderson (1986) stated that a rider who remained seated would not be able to use the ankling technique while pedalling as suggested in the popular literature. Kinematic analysis of the movement patterns of the lower right leg of the participants in this thesis confirmed that ankling did take place while cyclists were seated, contrary to the previous findings of Cavanagh and Sanderson (1986). Their study used pursuit riders (4000 metres) who may not be representative of the general competitive cycling population. As their study was published more than a decade ago, seat heights may have changed since then and therefore permit the ankling technique to be practised in the seated position. It has been shown that a change in seat height can alter the range of motion of the ankle joint (Ericson et al., 1988).

The kinematic analysis showed that the mean plantarflexion (27.4^{o}) was greater than the mean plantarflexion (23^{o}) reported by Sammarco et al. (1973). The mean angle of dorsiflexion was 19.0^{o}. This data compares favourably with the range of reported data (Boone and Azen, 1979, 13^{o}; Glanville and Kreezer, 1937, 40^{o}).

7.4.2 Range of motion and angular velocity relationship of the hip, knee and ankle joints

Under normal cycling conditions the range of motion (ROM) of the hip, knee and ankle joints are inter-related as the foot is attached to the pedal with a lock-in cleat or toe straps and provided the cyclist remains seated. Increasing the ROM of the ankle, during ankling, will decrease the ROM of the hip and knee joints. This decrease in the ROM will result in a decrease in angular velocity of those joints. Due to the force velocity relationship existing in a muscle, the decrease in angular velocity will allow a greater force to be exerted by the muscles acting on those joints. However, increasing the ROM of the ankle joint increases the angular velocity of that joint and therefore decreases the force able to be exerted by the muscles acting on that joint (i.e. in plantar flexion during ankling). Whether the reduction in force at the ankle joint totally negates any increase available at the hip and knee joints is not clear, however may be quite likely. This could lead to the conclusion that any increase in force was as a result of the improved FE index with the foot oriented more appropriately to allow the resulting force to be applied tangential to the movement of the crank.

7.4.2 Strengths of the Experimental Design

7.4.2.1 Competitive Athletes and a Large Sample Size

Many studies that have been reported in the literature have been conducted with recreational cyclists and small sample sizes (Table 7.15).

Author/s	Year	Cyclist type	n
Zommers	**1999**	**competitive**	**41**
Black et al	1993	licensed racers category 3 or higher	5
Cavanagh and Sanderson	1986	elite	7
Davis and Hull	1981	varied from inexperienced to elite	6
Ericson and Nisell	1988	recreational	6
Ericson et al	1988	recreational	6
Francis	1986	elite	1
Gregor	1976	competitive	5
Houtz and Fischer	1959	experienced	3
Hull and Jorge	1985	experienced	3
Jorge and Hull	1984	2 recreational, 1 ex racer	3
Jorge and Hull	1986	varying from touring to elite	6
Lafortune and Cavanagh	1983	college students	20
Lafortune et al	1983	elite	6
Sanderson	1987	recreational	8
Sanderson and Cavanagh	1990	recreational	6
Soden and Adeyefa	1979	elite	2

Table 7.15 Studies of cyclists from various backgrounds together with the sample size used

This study has attempted to meet the above criteria by using only competitive cyclists and a sample larger than all previous studies.

7.4.2.2 Exercise Intensity at Which Efficiency was Measured

The oxygen consumption during the efficiency tests averaged 75% of $VO_{2\ max}$. Since the anaerobic threshold occurs at about 70-80% of $VO_{2\ max}$ in endurance-trained athletes (Farrell et al, 1979), the thesis participants were exercising close to their anaerobic thresholds. This observation is supported by the mechanical power output measured during the anaerobic threshold tests that show that the anaerobic threshold varied from 75% to 95% of the mechanical power output corresponding to $VO_{2\ max}$. The mean respiratory exchange ratio (RER), during the efficiency tests, was 0.96, adding further evidence to argue that the workload of these athletes during the efficiency tests was simulating competition conditions.

The workload used in the biological efficiency tests was primarily based on the measured results of the anaerobic threshold tests. Despite the small adjustments to this work-load to ensure high performance conditions, the high correlation ($r = 0.87$, $p = 0.000$) between the mean mechanical power output of the efficiency tests and the mechanical power output at the anaerobic threshold indicates that the work-load finally used during the efficiency tests was indeed representative of competition performance conditions. The mechanical power output during the efficiency tests was an average of 90% of the power output measured at the anaerobic threshold with several athletes exercising close to or just over their anaerobic threshold. This data adds to the evidence that these cyclists were simulating competition conditions during the efficiency tests.

When the mechanical power output during the efficiency tests was expressed as a percentage of the mechanical power output at the anaerobic threshold, no significant difference was found between the experimental and control groups. This adds further evidence that the stratified random allocation of volunteers to the various sub-groups did not affect the homogeneity of the experimental and control groups. Although there was also no significant difference between the under and over 35 years of age participants, a significant difference was found between the males (87.8% ± 9.3%) and females (94.6% ± 6.8%) although no significant difference in the RER. It is suggested that the female participants in this study were either more willing or more able to exercise closer to their anaerobic thresholds and that the RER may not be an accurate indicator of exercise intensity in all cases.

7.4.2.3 Accurate Measurement of the Mechanical Power Output at Each Degree of Crank Revolution Using the Cyclists' own Bicycles

Two major objectives were achieved with the construction of the electronically-braked ergometry system developed by the author of this thesis. The regression R^2 obtained from the final calibration was (0.999; p = 0.000, F (14) = 114061) with a standard error of only 3.8 watts

(Section 4.3.3) that indicates that the measured variables are able to predict the mechanical power output of the cyclist being tested with a high degree of accuracy. A comparison with other ergometers, including those commonly used in exercise testing, revealed that the electronically-braked ergometry system, in this thesis, was superior in accuracy and reliability to both the mechanically-braked and air-braked ergometers. The air-braked ergometer was considerably more accurate and reliable than the mechanically-braked ergometer, provided environmental variables were measured and the ergometer was set-up to allow an unrestricted airflow for the turbine (Section 5.3.1).

The ability of the present system to be able to measure mechanical power output (and also cadence) at each degree of crank revolution meant that analysis of the angle where maximum power output was exhibited would be able to be measured. This feature also allowed the cadence to be graphed accurately showing the variation in cadence in one crank revolution. The ability to have such a high data-sampling rate has been possible with the rapid advances in the development of the personal computer (PC).

The second objective was to be able to test athletes on their own bicycle instead of an ergometer that may be uncomfortable and/or unfamiliar. Most sports institutes have constructed ergometers that replicate the cyclists' own racing bicycle. The seat and handlebar positions for each cyclist are recorded. This data is then used each time the cyclists attend the laboratory for a test. Some cyclists also bring with them their own seat, possibly for comfort. The ergometry system designed and constructed for this thesis has progressed this practice even more, by allowing the cyclists to be tested on their own bicycle rather than a facsimile. Anecdotal evidence from the adolescents tested during the pilot study (Chapter 6) indicated a large amount of satisfaction with being able to ride their own bicycles. During the recruitment for the main study, mountain bike cyclists were pleased that the ergometry system could accommodate a wide range of bicycles. This fact alone increased the range and number of athletes that volunteered for the study.

7.4.3 Physiological Variables

7.4.3.1 $VO_{2\ max}$

Burke (1982) reported unsourced data from other researchers that $VO_{2\ max}$ of international cycling teams varied between 65 $ml.kg^{-1}.min^{-1}$ for the U.S. junior national team to 74 $ml.kg^{-1}.min^{-1}$ for the U.S. Men's national team and a surprisingly low 57 $ml.kg^{-1}.min^{-1}$ for the U.S. women's national team. The $VO_{2\ max}$ data presented in Table 7.2 (Section 7.3.1) shows that some cyclists, both male and female, were of elite standard with the average of the athletes at least of a high competitive standard.

This would permit generalisation of the main thrust of this study (that is the question of whether ankling is more efficient than normal pedalling) to the broad range of competitive cyclists.

7.4.3.2 Anaerobic Threshold

There appears to be little agreement on which is the ideal method to define the anaerobic threshold by analysing the blood lactate concentrations (4 $mmol.L^{-1}.min^{-1}$ (AT4), individual anaerobic threshold (IAT), maximum lactate steady-state (MLSS) and lactate turnpoint; Davis et al, 1983; and Beneke, 1995). This study adopted the analysis procedure as defined by Davis et al (1983) that they termed the "lactate turnpoint". Commencement of the anaerobic threshold test at 50% of $VO_{2\,max}$ allowed a larger number of blood samples to be collected and enabled the identification of this lactate breakpoint. Nominating a workload of 95% (224.3 ± 53.1 watts) of this breakpoint was statistically significantly different to the eventual mechanical power output measured during the efficiency tests (208.8 ± 40.3 watts) but not significantly different from a practical point of view (difference only of 6.9%). As the RER measured during the efficiency tests was 0.96 ± 0.02 there was still room for a workload approaching the anaerobic threshold without exceeding an RER of 1.0.

There was a statistically significant difference in the RER between the under-35 years (0.96 ± 0.02) and over-35 years (0.97 ± 0.02) volunteers but this was a minuscule difference that will not have practical relevance.

In the current study, individual lactate thresholds occurred over a wide range of lactate concentrations (2.0 $mmol.l^{-1}$ - 7.1 $mmol.l^{-1}$). Although the mean lactate concentration at anaerobic threshold (4.1 $mmol.l^{-1}$ corresponding to 232 watts) was similar to the arbitrary 4 $mmol.l^{-1}$ level, the large variability between individuals precludes the application of this criterion (Heck et al, 1985; Kindermann et al, 1979 and Sjodin and Jacobs, 1981).

Eight volunteers either could not be cannulated or did not consent to blood sampling. The analysis of the efficiency tests, with the data grouped according to the type of anaerobic threshold test found no significant difference between the two types of anaerobic threshold test indicating that, in this study, the use of a ventilatory anaerobic threshold test for some athletes was a valid alternative to the lactate anaerobic threshold test. The relationship between the different methods of measurement of the anaerobic threshold has been discussed extensively (Chicharro et al, 1997; Dickstein et al, 1990; Neary et al, 1985; Simon et al, 1983; Yeh, et al, 1983; and Yoshida et al, 1981). The two methods used in this study produced comparable workloads for the efficiency phase of the study with athletes exercising at least at a training level if not at race performance (anecdotal evidence).

7.4.4 Data Additional to the Central Hypothesis

7.4.4.1 Crank Angle Where Maximum Mechanical Power Output Exhibited

The maximum mechanical power output measured during this study occurred at crank angles varying between 130° and 150°. Force at the rear wheel was calculated from the power and cadence. The crank angle where maximum force occurred at the rear wheel did not differ significantly from the crank angle where maximum power was exhibited despite the sinusoidal variation in crank angular velocity (cadence). Although only average power has been measured before, due to the lack of suitable equipment, the angle of the crank where maximum force is exhibited has been measured extensively using force pedals (Brooke et al, 1981, 1984; Davis and Hull, 1981; Patterson and Moreno, 1990; Sargeant and Davies, 1977; and Sargeant et al, 1978). Some authors have combined the force data from the left and right pedals which shows that the combined effective force at the rear wheel occurs at a crank angle of about 90° using elite 4000 metre pursuit cyclists (Cavanagh and Sanderson, 1986). The resultant maximum force occurred at about 115°. Brooke et al (1981) have provided applied force data for both pedals using what they termed as "enthusiastic ... subjects". Combining the results of both pedals indicates that the maximum force at the rear wheel occurred at a crank angle of 118°. The data from these two studies appear to be similar in the crank angle where maximum applied force is exhibited. Most ergometers used in studies of force application throughout the crank cycle have employed friction-braked ergometers – predominantly the Monark. These ergometers do not replicate the traditional racing bicycle. The horizontal seat to pedal position, on the Monark ergometer, is significantly more rearward than on a racing bicycle (Somerville, 1987). This would lead to a completely different pattern of force application. Also, the crank angle where maximum angular velocity occurred, in this study, varied between 142 and 149 degrees. As power is a function of angular velocity and force, then this may explain why the crank angle where maximum power occurred was not similar to the angle of maximum force application in previously reported studies.

The crank angle where maximum power was obtained at the rear wheel differed significantly between the two pedalling techniques. This was expected as the foot orientation also changed significantly. However this change in crank angle was only 1.04 degrees. This equates to a 0.7% change in crank angle. Although ankling resulted in large differences in foot/crank angle relativity, this had virtually no effect on the angle at which peak power occurred. This could explain the failure to find a significant difference in biological efficiency between the two pedalling techniques.

7.4.4.2 Cadence Variations During one Crank Cycle

Cadence varied sinusoidally in all athletes with a mean variation (refer Section 7.3.4.3) of approximately 7%. This variation has been dismissed by several researchers as insignificant and by others as non-existent (Hoes et al, 1968; Jorge and Hull, 1984; Kautz et al, 1991; Newmiller et al, 1988; and Redfield and Hull, 1986). This assumption may be valid with an ergometer that utilises a flywheel together with a fixed wheel. The fixed wheel does not allow the athlete to ease up on the pedalling as may occur on the road on a bicycle. The inertia of the flywheel, together with the fixed wheel, will smooth out most of the fluctuations in cadence. This smoothing of cadence fluctuations will not occur to the same extent on a bicycle. Although the forward velocity of the bicycle may be relatively constant on the road, due to the forward momentum, the velocity of the cranks can only be constant if the cyclist is using a "fixed wheel" with any movement in the crank being transmitted directly to the rear wheel of the bicycle. As the bicycles used on the road would have a free-wheel, that enables the rear wheel to keep rotating even though the cyclist may stop pedalling, any variations in cadence would not be apparent in the forward velocity of the bicycle.

7.4.4.3 Relationship Between the Horizontal Position of the Knee and the Bicycle Pedal With the Crank at 90°

Daniel Clement, a former French national cycling coach, defined one method of determining the ideal seat position (Hinault and Genzling, 1988b). He recommended dropping a plumb line from the front of the femur, with the bicycle cranks horizontal. The plumb line should fall over the pedal axle. In this study, the plumb line was dropped from the front of the kneecap as the front of the femur is more difficult to identify. The mean values from the experimental and control groups were 3.0 ± 31.86 mm and -9.16 ± 30.72 mm respectively (a negative value indicates that the front of the knee was behind the pedal axle). This suggests that volunteers were seated approximately according to the recommendations of Daniel Clement. However there was a large amount of variability in the data and it appears that there is little agreement amongst cyclists on the ideal position. Bernard Hinault followed Clement's recommendation for some time until later ergonometric studies found that moving the seat back several centimetres was better for him. Several ideas have emerged over time on the ideal position of the seat including one by Hinault himself. He recommends that the horizontal seat position, measured from the centre of the crank, should be relative to the inside leg length (measured standing with feet slightly apart wearing cambered shoes). This method assumes that all athletes have the same ratio between the length of the lower and upper legs. In this study the mean inside leg length was 848.2 ± 53.3 mm and the seat setback was 61 ± 23 mm. This is in close agreement with Hinault's recommendation of 72 mm for the same inside leg length (Hinault and Genzling, 1987a). However, the large variability in the measurements suggests that athletes either are not aware of the recommendations or simply have chosen to be different. Anecdotal evidence suggests that cyclists are encouraged to experiment by varying the parameters of their bicycle from the "ideal" in small amounts until they find what suits them best.

7.4.4.4 Seat Height - Inside leg Length Ratio

Cyclists once had the seat height low compared to their inside leg length. They would adjust the seat height so that the heel could reach the pedal with the crank at bottom dead centre (BDC). Changes in shoe design permitted the seat height to be raised (Hinault and Genzling, 1988b). This changed the range of motion of the hip allowing the cyclist to adopt a more streamlined upper body position. The participants in this study had a seat height - leg length ratio of 0.87 ± 0.03. There was no significant difference to that recommended by Hinault (0.885; Hinault and Genzling, 1987). Two other methods have been used and recommended by researchers (Gonzalez and Hull, 1989; Hamley and Thomas, 1967; Lafortune and McLean, 1989; Nordeen-Snyder, 1977; Shennum and deVries, 1976; Welbergen and Clijsen, 1990).

- Seat height (refer to Figure 7.6) plus crank arm length should be 97 - 100% of the distance measured from the greater trochanter to the floor standing in bare feet.
- Seat height (refer to Figure 7.6) plus crank arm length should be 107 - 109% of the medial aspect of the inside leg from the symphysis pubis to the floor standing in bare feet.

The centre of the greater trochanter is difficult to find in some volunteers, especially females, who may exhibit a high concentration of fat tissue in this region. The second method is similar to that proposed by Hinault and

Genzling (1987). This method does include the crank length measurement, however the crank length in this study was predominantly 170mm (n = 28) with a mean crank length of 171.3 mm. Variations to this length (165mm - 180mm) are only necessary with very short or very tall cyclists. Both of the above methods ignore the different sizes and types of shoes worn by cyclists (both used bare feet). The ideal method would be to measure the seat height (refer to Figure 7.6) plus crank length relative to the inside leg length measured while wearing cycling shoes. No studies have reported this method of measurement.

7.4.4.5 Efficiency Percentage

The biological net efficiency percentage during the study, was 22.4% ±2.3%. The net efficiency reported in the literature during the last 30 years varies from 20% to 24% (Table 7.16) and it is comparable to the data in this thesis (Chapter 6, 7; Zommers et al 1995c). Gross efficiency was also measured in this study however it was not deemed to be a valid measurement as the resting VO_2 did not vary linearly relative to the VO_2 measured during the efficiency tests. Athletes with a low resting VO_2 could display a higher biological efficiency without necessarily being more efficient while exercising. A high level of validity is displayed in the efficiency results as the volunteers were tested on their own bicycles on an accurate ergometry system.

Author(s)	n	Subject type	Notes	Gross efficiency		Net efficiency		Work efficiency		Delta efficiency	
				% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)
Zommers (1999)	**41**	**Competitive**				22.4	209 watts (80.4)				
Dickinson (1929)	1	Unknown				11.4	178 watts (222 rpm)				
						22	280 watts (67 rpm)				
Garry and Wishart (1931)	2	Experimenters	Mean data calculated from selected samples [1,2]	12.75	89 watts (96.5 rpm)	14.95	89 watts (96.5 rpm)	33.35	89 watts (96.5 rpm)		
				17.15	111 watts (52 rpm)	20.55	111 watts (52 rpm)	25.25	111 watts (52 rpm)		
Banister and Jackson (1967)	1	Olympic Gold Medal oarsman	Mean data calculated [1]	18.2	195 watts (85 rpm)						
Whipp and Wasserman (1969)	8	Healthy male volunteers				20.2	57 watts (60 rpm)	29.8	57 watts (60 rpm)		

Table 7.16 Studies on biological efficiency of cycling

Notes:

1. Net efficiency was calculated in this thesis (see Section 2.2 for justification).

2. The author of this thesis calculated mean efficiency from the data of Garry and Wishart (1931) to simplify the table presentation.

3. Samples were selected from the published data to show relative efficiency for the same cadence. The data chosen does not necessarily show the highest and lowest efficiency for each method of calculation.

Author(s)	n	Subject type	Notes	Gross efficiency		Net efficiency		Work efficiency		Delta efficiency	
				% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)	% eff	Power output (cadence)
Asmussen and Bonde-Petersen (1974)	3	1. average fitness skin diver 2. experienced middle/long distance runner 3. a fit cyclist	Study used a bicycle on a motorised treadmill					25.1	8.36 km/h (65 rpm)		
Gaesser and Brooks (1975)	12	Well conditioned males	Selected samples shown [2]	16.6	65.4 watts (60 rpm)	22.3	65.4 watts (60 rpm)	26.8	65.4 watts (60 rpm)	29	65.4 watts (60 rpm)
				20.4	130.8 watts (60 rpm)	24.1	130.8 watts (60 rpm)	26.6	130.8 watts (60 rpm)	26.3	130.8 watts (60 rpm)
Seabury et al. (1977)	3	Two trained distance runners and one active recreational cyclist		15.4	40.8 watts (42 rpm)						
				23.6	326.8 watts (64 rpm)						
Faria et al. (1982)	4	Elite cyclists		14	142 watts (130 rpm)						
			Mean data calculated [1]	22	291 watts (100 rpm)						
Zommers et al. (1995c)	15	Endurance trained adolescent triathletes	Mean data calculated [1]			21	184 watts (100 rpm)				

Table 7.16 Studies on biological efficiency of cycling (cont)

CHAPTER 8

SUMMARY OF RESULTS AND CONCLUSIONS

8.1 OVERVIEW

This research focused on whether cyclists were biologically more efficient if they adopted a pedalling technique known as ankling. The following discussion examines the results of the central hypotheses and the supporting hypotheses. The central hypotheses include two concepts. First, whether the ankling technique is anatomically feasible and second, whether use of the ankling technique leads to greater biological efficiency. The supporting hypotheses focused on whether the crank angle at peak power varied with the two techniques, whether cadence varied within one crank cycle and whether biological efficiency was similar to that previously reported in the literature. The discussion ends with a summary of the calibration results of the bicycle ergometry equipment used in this study together with a comparison with other ergometers traditionally used in exercise physiology laboratories.

8.2 CENTRAL HYPOTHESES

Despite claims that the ankling technique was not anatomically possible, video analysis showed that all athletes were able to move their ankle joint through a large range of motion (ROM) demonstrating that the ankling technique was a viable alternative to other pedalling techniques exhibited by elite cyclists (Cavanagh and Sanderson, 1986). Seat heights have increased over the years and this may have prevented the large ROM of the ankle joint and therefore led Cavanagh and Sanderson (1986) to claim that ankling was not possible if the cyclist remained seated (Hinault and Genzling, 1987a). In the present study the athletes remained seated and indeed were able to demonstrate the ankling technique as instructed.

Ankling was practised for three months but the biological efficiency was significantly less than for normal pedalling. All efforts had been made to ensure that the measurement equipment was accurate and reliable so that a small effect size had a good chance of being detected. When the seat height/leg length ratio was added as a covariable, there was no longer a significant difference between the biological efficiency of the two pedalling techniques. There was a non-significant trend of convergence of the two techniques for the experimental (ankling) group. This raises the possibility that if pedalling technique had been studied for a longer follow-up (say 6-9 months), and the ideal seat height/leg length ratio was used for all athletes, then ankling may have proved to be more efficient than normal pedalling. However there is no evidence to support this hypothesis.

8.3 SUPPORTING HYPOTHESES

The ankling technique was designed to increase the effective force applied to the cranks between top dead centre and bottom dead centre. As the effectiveness of the force would be greater especially between 90° and 180° it is reasonable to expect that the angle where maximum power was exhibited would be different between the two pedalling techniques. The difference of one degree was statistically significant but trivial in practical terms.

Several researchers had assumed that cadence did not vary within one crank revolution (Hoes et al, 1968; Jorge and Hull, 1984; Kautz et al, 1991; Newmiller et al, 1988; and Redfield and Hull, 1986). This assumption was based on relatively crude measurements of crank velocity, by today's standards. The ergometry equipment used in this study was capable of measuring cadence every degree of crank revolution. This enabled the observation that cadence varied sinusoidally in all cyclists with two complete cycles of the sine wave for every crank revolution.

8.4 ACCURACY AND RELIABILITY OF THE ELECTRONICALLY-BRAKED BICYCLE ERGOMETRY SYSTEM USED IN THIS STUDY AND A COMPARISON WITH EXISTING AIR- AND MECHANICALLY-BRAKED BICYCLE ERGOMETERS.

The results in Chapter 4 indicated that the electronically-braked bicycle ergometry system was not only highly accurate and reliable, more so than either the air- or mechanically-braked ergometers discussed in Chapter 5. This was achieved by using high quality components in the design of the electronics interface and the inclusion of all possible variables in the prediction of mechanical power. Not only was accurate measurement of mechanical power possible but also participants were able to use their own bicycles thereby removing the need for extensive familiarisation with a laboratory ergometer.

8.5 BIOLOGICAL EFFICIENCY

Cycling is known to be the most efficient form of transportation (Wilson, 1973) with net efficiency of 22 ± 4% (Andersen et al., 1971). Several studies have reported similar efficiencies, including studies of adolescent triathletes (Dickinson, 1929; Garry and Wishart, 1931; Gaesser and Brooks, 1975; Rowland et al, 1990; Springer et al, 1991; Whipp and Wasserman, 1969; and Zommers et al, 1995c). In the present study net efficiency, for ankling, averaged 22.2 ± 2.3% which compares favourably with the reported literature.

8.6 LIMITATIONS OF THE STUDY

Although the study was designed to minimise variability in the data, several limitations became apparent during the study and after the study was completed. First, as the athletes were tested on two occasions prior to the efficiency testing phase ($VO_{2\,max}$ and anaerobic threshold test); it was felt that pedalling technique instructions immediately prior to the first efficiency test would be adequate. Visual observation of the pedalling technique of each cyclist during the first efficiency test revealed that this decision was correct. Observation of the digitised video data at the completion of the study revealed the volunteers conformed better to the ankling pedalling pattern as the study progressed (Efficiency test 1 to test 4). With the seat height set to the ideal position for each cyclist it is believed that a familiarisation period may have significantly decreased the variability in the pedalling pattern and may have shown that the ankling technique was more efficient than normal pedalling supporting the central hypothesis. This familiarisation phase may also have allowed a more accurate selection of the ideal work-load for the remainder of the study.

In order to minimise drop-out from the study the number of efficiency tests was limited to four with one month of practice in between each test. The trend in the delta efficiency indicated that more practice time might have resulted in ankling being as or even more efficient than normal pedalling.

One major problem with any study is compliance with instructions. There is no way of ensuring that they follow instructions in regard to technique practice, and preparation (food, fluid and rest) for each efficiency test. Even a diary could be inaccurate as entries could be made several hours or days later.

The volunteer cyclists in the experimental group were instructed to practise the ankling technique during training sessions and if possible use the technique during competition. Prior to each efficiency test cyclist were asked to report how much time was spent practising ankling. Feedback was then given on the percentage of time to be spent practising ankling prior to the next efficiency test. From personal experience, adopting ankling as a preferred pedalling technique takes time and practice. Athletes were not expected to use ankling 100% of the time during training but were instructed to practise ankling all of the time, during training, by the fourth efficiency test. Having the volunteers document the actual amount of time spent ankling may have served as a daily reminder of the objective and ensured a greater compliance with the instructions.

After analysis of the data it was clear that another limitation of the study was the inability to measure the actual forces applied to the pedals. This may have shown that despite the changes in pedal orientation, as a result of the ankling technique, the resultant force did not change and therefore an increase in mechanical and therefore metabolic efficiency could not be expected.

CHAPTER 9

RECOMMENDATIONS FOR FURTHER RESEARCH

The current study has assisted in addressing the longstanding controversy concerning the efficiency of ankling technique. Previous studies had only examined the force application from a theoretical perspective without analysis of the biological efficiency of the human body. From an empirical perspective, ankling was not considered the ideal technique as elite cyclists rarely adopted it as their preferred method of pedalling.

Several studies have attempted to modify the pedalling technique of cyclists using real-time feedback of the force pattern on the pedals during the "pull-up" phase (i.e. from bottom dead centre (BDC) to top dead centre (TDC) of the crank cycle; McLean, 1989; and Sanderson and Cavanagh, 1990). The biofeedback succeeded in reducing the counter-productive forces during the "pull-up" phase and although the feedback was progressively removed the cyclists in one study did not revert to their previous technique after 10 days of practice (Sanderson and Cavanagh, 1990). Another study that had similar results tested eight cyclists two months after the completion of the initial study and found no retention loss (Broker et al, 1993). It is therefore concluded that further studies on changes to the pedalling technique from TDC to BDC (such as ankling) could benefit from a longer period in the practice phase together with feedback during each testing session. This is supported by the present observation of a non-significant convergence of the two techniques. However there is no suggestion of a cross-over effect (i.e. ankling becoming more efficient after say 6 months, compared to normal technique).

Studies on the effect of seat height on biological efficiency have generally agreed on the ideal height despite small differences in the measurement of leg-length (Hamley and Thomas, 1967; Nordeen-Snyder, 1977; and Shennum and deVries, 1976). When this variable was included in the analysis of biological efficiency in this study, the results changed from ankling being significantly less efficient to there being no significant difference, despite the difference in the means. Further studies on the seat height/leg length ratio comparing the biological efficiency of ankling and normal pedalling would certainly aid in resolving whether ankling was a more efficient technique than normal pedalling. This intervention would need to be imposed sequentially (i.e. prior to the technique intervention) and would add tremendous complexity to the research design.

Research by others using a commercially-available cycling ergometers is limited by the accuracy of these ergometers (see Section 5.3.1). One of the strengths of the present thesis was the development of a highly accurate and reliable ergometry system that could be used in other studies of cycling technique or physiology where the expected effect sizes are small.

REFERENCES

Andersen, K.L., Shephard, R.J., Denolin, H., Varnauskas, E., & Masironi, R. (1971). *Fundamentals of Exercise Testing.* Geneva, Switzerland: World Health Organisation. p 74.

Asmussen, E., & Bonde-Petersen, F. (1974). Apparent efficiency and storage of elastic energy in human muscles during exercise. *Acta Physiol. Scand., 92: 537-545.*

Attaway, R., Bartoli, W.P., Pate, R.R., & Davis, J.M. (1992). Physiologic and perceptual responses to exercise on a new cycle ergometer. *Canadian Journal of Sports Science, 17(1): 56-59.*

Banister, E.W., & Jackson. R.C. (1967). The effect of speed and load changes on oxygen intake for equivalent power outputs during bicycle ergometry. *Internationale Zeitschrift fur Angewandte Physiologie Einschliesslich Arbeitsphysiologie, 24: 284-290.*

Bartlett, R.M. (1989). *Biomechanical Assessment of the Elite Athlete.* UK: British Association of Sports Sciences.

Behring Diagnostics Inc. Stat-Pack TM Rapid Lactate Test: Lactate Reagents.

Beneke, R. (1995). Anaerobic threshold, individual anaerobic threshold, and maximal lactate steady state in rowing. *Medicine and Science in Sports and Exercise, 27(6): 863-867.*

Black, A.H., Sanderson, D.J., & Hennig, E.M. (1993). *Kinematic and kinetic changes during an incremental exercise test on a bicycle ergometer.* Melbourne, Australia: (Annual Scientific Conference in Sports Medicine)

Boone, D.C., & Azen, S.P. (1979). Normal range of motion of joints in male subjects. *Journal of Bone and Joint Surgery, 61A(5):756-759.*

Broker, J.P., Gregor, R.J., & Schmidt, R.A. (1993). Extrinsic feedback and the learning of kinetic patterns in cycling. *Journal of Applied Biomechanics, 9:111-123.*

Broker, J.P. & Gregor, R.L. (1996). Cycling Biomechanics. In: Burke, E.R. *High Tech Cycling.* Champaign, Illinois, USA: Human Kinetics Books. pp. 145-165.

Brooke, J.D., Hoare, J., Rosenrot, P, & Triggs, R. (1981). Computerized system for measurement of force exerted within each pedal revolution during cycling. *Physiology and Behaviour, 26:139-143.*

Brooke, J.D., Dalton, J., Hoare, J., Rosenrot, P., & Wilson, B.A. (1984). Variability of pedalling force - inversely related to force magnitude within the movement cycle but positively correlated to that magnitude as it increases with increased power output. *Journal of Human Ergology, 13:121-128.*

Burke, E. (1982). In Case You're Curious. *Bicycling, 23:*29.

Carpenter, T.M. (1964). *Tables, Factors and Formulas for Computing Respiratory Exchange and Biological Transformations of Energy. 4th Ed.* DC, USA: Carnegie Institute of Washington.

Casaburi, R., Storer, T.W., Ben-Dov, I., & Wasserman, K. (1987). Effect of endurance training on possible determinants of VO_2 during heavy exercise. *Journal of Applied Physiology, 62(1):199-207.*

Cavanagh, P.R., & Nordeen, K.S. (1976). Biomechanical studies of cycling: instrumentation and application. *Medicine and Science in Sports, 8(1):61-62.*

Cavanagh, P.R., & Sanderson, D.J. (1986). The biomechanics of cycling: Studies of the pedaling mechanics of elite pursuit riders. In: Burke, E. *Science of Cycling*. Champaign, Illinois, USA: Human Kinetics Books. pp. 91-122.

Chapman, A.E. (1985). The mechanical properties of human muscle. *Exercise & Sports Sciences Reviews, 13: pp. 443-501.*

Chicharro, J.L., Perez, M., Vaquero, A.F., Lucia, A., & Legido, J.C. (1997). Lactic threshold vs ventilatory threshold during a ramp test on a cycle ergometer. *Journal of Sports Medicine and Physical Fitness, 37(2):117-121.*

Coyle, E.F., Feltner, M.E., Kautz, S.A., Hamilton, M.T., Montain, S.J., Baylor, A.M., Abraham, L.D., & Petrek, G.W. (1991). Physiological and biomechanical factors associated with elite endurance cycling performance. *Medicine and Science in Sports and Exercise, 23(1):93-107.*

Craig, N., & Walsh, C. (1989). Cycling. In: Draper, J., & Telford, R. (Eds.). *Sport Specific Guidelines for the Physiological Assessment of the Elite Athlete.* ACT Australia: Australian Sports Commission.

Cullen, L.K., Andrew, K., Lair, K.R., Widger, M.J., & Timson, B.F. (1992). Efficiency of trained cyclists using circular and noncircular chainrings. *International Journal of Sports Medicine, 13:264-269.*

Davis, H.A., Bassett, J., Hughes, P., & Gass, G.C. (1983). Anaerobic threshold and lactate turnpoint. *European Journal of Applied Physiology, 50:383-392.*

Davis, R.R., & Hull, M.L. (1981). Measurement of pedal loading in bicycling: II. Analysis and results. *Journal of Biomechanics, 14(12):857-872.*

DeGroot, G., Welbergen, E., Clijsen, L., Clarijs, J., Cabri, J., & Antonis, J. (1994). Power, muscular work, and external forces in cycling. *Ergonomics, 37(1): pp. 31-42.*

DeVries, H.A., & Housh, T.J. (1994). *Physiology of Exercise for Physical Education, Athletics and Exercise Science.* (5th Edition). IA, USA: Brown and Benchmark.

Dickerson, A. (1972). *Calibration Check on Repco Cycle Ergometer.* Melbourne, Australia: Repco Research Pty Ltd.

Dickinson, S. (1929). The efficiency of bicycle-pedalling as affected by speed and load. *Journal of Physiology, 67:242-255.*

Dickstein, K., Barvic, S., Aarsland, T., Snapinn, S., & Karlsson, J. (1990). A comparison of methodologies in detection of an anaerobic threshold. *Circulation, 8(supplement II):38-46.*

Durham, R. (February, 1975). The new elliptical chainwheel. *Bike World Magazine, pp 20-21.*

Eastop, T.D., and McConkey, A. (1970). *Applied Thermodynamics for Engineering Technologists.* London: Longman Group Ltd.

Ericson, M.O., Bratt, A., Nisell, R., Arborelius, U.P., & Ekholm, J. (1986). Power output and work in different muscle groups during ergometer cycling.,*European Jurnal of Applied Physiology, 55: pp. 229-235.*

Ericson, M.O., & Nisell, R. (1988). Efficiency of pedal forces during ergometer cycling. *International Journal of Sports Medicine, 9:118-122.*

Ericson, M.O., Nisell, R., & Nemeth, G. (1988). Joint motions of the lower limb during ergometer cycling. *The Journal of Orthopaedic and Sports Physical Therapy, 9(8):273-278.*

Faria, I., & Cavanagh, P.R. (1978). *The Physiology and Biomechanics of Cycling.* Canada: John Wiley & Sons, Inc.

Faria, I., Sjojaard, G., & Bonde-Petersen, F. (1982). Oxygen cost during different pedalling speeds for constant power output. *Journal of Sports Medicine, 22:295-299.*

Farrell, P.A., Wilmore, J.H., Coyle, E.F., Billings, J.E., & Costill, D.L. Plasma lactate accumulation and distance running performance. *Medicine in Science Sports and Exercise, 11:338-344.*

Flood, D.K. (1996). *Practical Math for Health Fitness Professionals.* IL, USA: Human Kinetics.

Francis, P.R. (1986). Injury prevention for cyclists: A biomechanical approach. In: Burke, E. (Ed.). *Science of Cycling.* Champaign, Illinois, USA: Human Kinetics Books. pp. 145-184.

Freund, R.J., & Littell, R.C. *SAS System for Regression.* NC. USA: SAS Institute Inc.

Gaesser, G.A., & Brooks, G.A. (1975). Muscular efficiency during steady-rate exercise: effects of speed and work rate. *Journal of Applied Physiology, 38(6):1132-1139.*

Garas, M. (1998). *Self excitation of alternators.* Bosch, Melbourne, Australia. (Email communication).

Garry, R.C., & Wishart, G.M. (1931). On the existence of a most efficient speed in bicycle pedalling, and the problem of determining human muscular efficiency. *Journal of Physiology, 72: 425-437.*

Glanville, A.D., & Kreezer, G. (1937). The maximum amplitude and velocity of joint movements in normal male human adults. *Human Biology, 9:197-211.*

Gonzalez, H., & Hull, M.L. (1989). Multivariate optimization of cycling biomechanics. *Journal of Biomechanics, 22(11/12):1151-1161.*

Gravel, D., Richards, C.L., & Filion, M. (1990). Angle dependency in strength measurements of the ankle plantar flexors. *European Journal of Applied Physiology, 61:182-187.*

Gregor, R.J. (1976). *A Biomechanical Analysis of Lower Limb Action during Cycling at Four Different Loads.* PhD, Pennsylvannia State University.

Gulch, R.W. (1994). Force-velocity relations in human skeletal muscle. *International Journal of Sports Medicine, 15: pp. S2-S10.*

Hagberg, J.M., Mullin, J.P., & Nagle, F.J. (1978). Oxygen consumption during constant-load exercise. *Journal of Applied Physiology Respiration and Environmental Exercise Physiology, 45(3):381-384.*

Hamley, E.J., & Thomas, V. (1967). Physiological and postural factors in the calibration of the bicycle ergometer. *Journal of Physiology, 191:55-57.*

Harman, E. (1989). Letter to the editor-in-chief. *Medicine and Science in Sports and Exercise, 21(4):487.*

Harrison, J.Y. (1970). Maximizing human power output by suitable selection of motion cycle and load. *Human Factors, 12(3):315-329.*

Hay, J.G., & Reid, G.J. (1982). *The Anatomical and Mechanical Bases of Human Motion.* New Jersey, USA: Prentice Hall.

Heck, H., Mader,, A., Hess, G., Mücke, S., Müller, R., & Hollmann, W. (1985). Justification of the 4-mmol.l^{-1} lactate threshold. *International Journal of Sports medicine. 6:117-130.*

Henderson, S.C., Ellis, R.W., Klimovitch, G., & Brooks, G.A. (1977). The effects of circular and elliptical chainweels on steady-rate cycle ergometer work efficiency. *Medicine and Science in Sports, 9(4):202-207.*

Hinault, B., & Genzling, C. (1987a). There's more to pedaling than you think. *Velo News - A Journal of Bicycle Racing. 16(13) pp:16-17.*

Hinault, B., & Genzling, C. (1987b). Increase Efficiency with better position. *Velo News - A Journal of Bicycle Racing. 16(18):12-13.*

Hinault, B., & Genzling, C. (1988a). Body Measurements Help Refine Position. *Velo News - A Journal of Bicycle Racing. 17(1):8-9*

Hinault, B., & Genzling, C. (1988b). *Road racing: technique and training.* UK: Springfield Books Limited.

Hoes, M.J., Binkhorst, R.A., Smeekes-Kuyl, A.E., & Vissers, A.C. (1968). Measurement of forces exerted on pedal and crank during work on a bicycle ergometer at different loads. *Internationale Zeitschrift fur Angewandte Physiologie Einschliesslich Arbeitsphysiologie, 26:33-42.*

Hooper, L.A. (1977). *Report on the Calibration of the "Lode" Eddy Current Bicycle Ergometer.* Melbourne, Australia: Repco Research Pty Ltd.

Hope, J.S. (1975). *Middle Level Physics.* Australia: Pitman.

Houtz, S.J., & Fischer, F.J. (1959). An analysis of muscle action and joint excursion during exercise on a stationary bicycle. *Journal of Bone and Joint Surgery, 41A(1):123-131.*

Hull, M.L., & Davis, R.R. (1981). Measurement of pedal loading in bicycling: I. Instrumentation. *Journal of Biomechanics, 14(12):843-855.*

Hull, M.L., & Gonzalez, H. (1988). Bivariate optimization of pedalling rate and crank arm length in cycling. *Journal of Biomechanics, 21(10):839-849.*

Hull, M.L., & Gonzalez, H. (1990). The effect of pedal platform height on cycling biomechanics. *International Journal of Sport Biomechanics, 6:1-17.*

Hull, M.L., & Jorge, M. (1985). A method for biomechanical analysis of bicycle pedalling. *Journal of Biomechanics, 18(9):631-644.*

Hull, M.L., Kautz, S., & Beard, A. (1991). An angular velocity profile in cycling derived from mechanical energy analysis. *Journal of Biomechanics, 24(7):577-586.*

Jackson, K.M. (1979). Fitting of mathematical functions to biomechanical data. *IEEE Transactions on Biomedical Engineering.* pp 122-124.

Jorge, M., & Hull, M.L. (1984). Biomechanics of bicycle pedalling. In Terauds, J., (Ed.). *Sports Biomechanics, Proceedings of ISBS.*

Jorge, M., & Hull, M.L. (1986). Analysis of EMG measurements during bicycle pedalling. *Journal of Biomechanics, 19(9):683-694.*

Kautz, S., Feltner, M.E., Coyle, E.F., & Baylor, A.M. (1991). The pedaling technique of elite endurance cyclists: Changes with increasing workload at constant cadence. *International Journal of Sport Biomechanics, 7(1):29-53.*

Kautz, S.A., & Hull, M.L. (1993). Theoretical basis for interpreting the force applied to the pedal in cycling. *Journal of Biomechanics, 26(2):155-165.*

Keppel, G. (1991). *Design and Analysis - A Researcher's Handbook.* (3rd Edition). New Jersey, USA: Prentice-Hall, Inc.

Kindermann, W., Simon, J., & Keul, J. (1979). The significance of the aerobic-anaerobic transition for the determination of workload intensities during endurance training. *European Journal of Applied Physiology, 42:25-34.*

Kolin, M.J., & De la Rosa, D.M. (1979). *The Custom Bicycle.* Emmaus, PA, USA: Rodale Press.

Kyle, C.R., & Edelman, W.E. (1974). Manpowered vehicle design criteria. *Proceedings of the 3rd International Conference Vehicle System Dynamics, pp.20-30.*

Lafortune, M. (1986). Cycling from a biomechanical perspective. *Sports Science and Medicine Quarterly, 2(3):8-10.*

Lafortune, M., & Cavanagh, P.R. (1980). Force effectiveness during cycling. *Medicine and Science in Sport, 12(95):95.*

Lafortune, M., & Cavanagh, P.R. (1983). Effectiveness and efficiency during bicycle riding. In Matsui, H., & Kobayashi, K., (Eds.). *Biomechanics VIII A & B: Proceedings of the Eighth International Congress of Biomechanics.*

Lafortune, M., & McLean, B. (1989). Biomechanical investigation of pedalling techniques of elite and recreational cyclists. In: Australian Sports Commission, *Technical Report.*

Lafortune, M.A., Cavanagh, P.R., Valiant, G.A., & Burke, E.R. (1983). A study of the riding mechanics of elite cyclists. *Medicine and Science in Sports and Exercise, 15(1):113.*

Lieb, T. (1980). A look at Bernard Hinault's pedaling motion under the microscope. *Bicycling, 21(9):22,24.*

Lowry, O.H., & Passoneau, J.V. (1972). A *flexible system of enzymatic analysis.* New York, USA: Academic Press.

Marsh, A.P., & Martin, P.E. (1993). The association between cycling experience and preferred and most economical cadences. *Medicine and Science in Sports and Exercise, 25(110:1269-1274.*

Maxwell, B.F., Withers, R.T., Ilsley, A.H., Wakim, M.J., Woods, G.F., & Day, L. (1998). Dynamic calibration of mechanically, air- and electromagnetically braked cycle ergometers. *European Journal of Applied Physiology, 78: pp. 346-352.*

McLean, B. (1989). Pull-up theory is now pedalling history. *Australian Cycling and Triathlon News, 5(4): 26.*

McLean, B., & Lafortune, M.A. (1988). Improving pedalling technique with "real time" biomechanical feedback. *Excel, 5(1):15-18.*

McLellan, T.M. (1987). The anaerobic threshold: Concept and controversy. *The Australian Journal of Science and Medicine in Sport, 19(2):3-8.*

McLoughlin, P., Popham, P., Linton, R.A., Bruce, R.C., & Band, D.M. (1992). Use of arterialized venous blood sampling during incremental tests. *Journal of Applied Physiology, 73(3):937-940.*

Milesis, C.A., Sprecker, T.B. & Chumbley, R.A. (1991). Physiological and perceptual responses to friction-braked vs. air-braked ergometer cycling. *Clinical Kinesiology, 45(1):3-8.*

Miller, N.R., & Ross, D. (1980). The design of variable- ratio chain drives for bicycles and ergometers - application to a maximum power bicycle drive. *Journal of Mechanical Design, 102(4):711-717.*

Neary, P.J., MacDougall, J.D., Bachus, R., & Wenger, H.A. (1985). The relationship between lactate and ventilatory thresholds: coincidental or cause and effect? *European Journal of Applied Physiology, 54:104-108.*

Newmiller, J., & Hull, M.L. (1988). A mechanically decoupled two force component bicycle pedal dynamometer. *Journal of Biomechanics, 21(5):375-386.*

Nordeen-Snyder, K.S. (1977). The effect of bicycle seat height variation upon oxygen consumption and lower limb kinematics. *Medicine and Science in Sports, 9(2):113-117.*

Ohlsson, O. (1989). Response to Letter to editor-in-chief. *Medicine and Science in Sports and Exercise, 21(4):487-488.*

Okajima, S. (1983). Designing chainweels to optimize the human engine. *Bike Tech, 2(4):1-7.*

Palmer, G.S., Dennis, S.C., Noakes, T.D., & Hawley, J.A. (1996). Assessment of the reproducibility of performance testing on an air-braked cycle ergometer. *International Journal of Sports Medicine, 17(4): pp. 293-298.*

Papadopoulos, J.M. (1987). Forces in bicycle pedalling. In: Rekow, E.D., (Ed.). *Biomechanics in Sport - A 1987 Update.* Boston, USA: The American Society of Mechanical Engineers.

Patterson, R.P., & Moreno, M.I. (1990). Bicycle pedalling forces as a function of pedalling rate and power output. *Medicine and Science in Sports and Exercise, 22(4):512-516.*

Philips Components - Signetics Corporation. (1991). *80C51 and derivative microcontrollers: data handbook / Philips Components.* Eindhoven, The Netherlands: Philips Components.

Press, W.H. (1988). *Numerical Recipes in C: The Art of Scientific Computing.* New York, USA: Cambridge University Press.

Price, D., & Donne, B. (1997). Effect of variation in seat tube angle at different seat heights on submaximal cycling performance in man. *Journal of Sports Sciences, 15: pp. 395-402.*

Redfield, R., & Hull, M.L. (1986). On the relation between joint moments and pedalling rates at constant power in bicycling. *Journal of Biomechanics, 19(4):317-329.*

Robert Bosch. (1989). *Initial Performance Curve for 28Volt K1 Alternator.* Australia: Robert Bosch.

Roosa, D. (1988). Shimano biopace. *Bicycle Guide, 5(8):79-85.*

Rowland, T.W., Staab, J.S., Unnithan, V.B., Rambusch, J.M., & Siconolfi, S.F. (1990). *Mechanical efficiency during cycling in prepubertal and adult males, 11(6): pp. 452-455.*

Sammarco, G.J., Burstein, A.H., & Frankel, V.H. (1973). Biomechanics of the ankle: a kinematic study. *Orthopaedic Clinics of North America, 4(1):75-96.*

Sanderson, D.J. (1987). Training with biofeedback. *Bike Tech, 6(1):10-13.*

Sanderson, D.J., & Cavanagh, P.R. (1990). Use of augmented feedback for the modification of the pedaling mechanics of cyclists. *Canadian Journal of Sports Science, 15(1):38-42.*

Sargeant, A.J., & Davies, C.T.M. (1977). Forces applied to cranks of a bicycle ergometer during one- and two-leg cycling. *Journal of Applied Physiology Respiration and Environmental Exercise Physiology, 42(4):514-518.*

Sargeant, A.J., Charters, A., Davies, C.T.M., & Reeves, E.S. (1978). Measurement of forces applied and work performed in pedalling a stationary bicycle ergometer. *Ergonomics, 21(1):49-53.*

Savitzky, A., & Golay, M.J. (1964). Smoothing and differentiation of data by simplified least squares procedures. *Analytical Chemistry, 36(8):1627-1639.*

Seabury, J.J., Adams, W.C., & Ramey, M.R. (1977). Influence of pedalling rate and power output on energy expenditure during bicycle ergometry. *Ergonomics, 20(5):491-498.*

Sharp, A. (1896). *Bicycles and Tricycles.* London: Longmans, Green and Company.

Shennum, P.L., & deVries, H.A. (1976). The effect of saddle height on oxygen consumption during bicycle ergometer work. *Medicine and Science in Sports, 8(2):119-121.*

Simon, J., Young, J.L., Gutin, B., Blood, D.K., & Case, R.B. (1983). Lactate accumulation relative to the anaerobic and respiratory compensation thresholds. *Journal of Applied Physiology: Respiration and Environmental Exercise Physiology, 54(1):13-17.*

Sjodin, B., & Jacobs, I. (1981). Onset of blood lactate accumulation and marathon running performance. *International Journal of Sports Medicine, 2:160-165.*

Skinner, J.S., & McLellan, T.M. (1980). The transition from aerobic to anaerobic metabolism. *Research Quarterly for Exercise and Sport, 51:234-248.*

Soden, P.D., & Adeyefa, B.A. (1979). Forces applied to a bicycle during normal cycling. *Journal of Biomechanics, 12:527-541.*

Somerville, K.A., & Quinney, H.A. (1987). A modified cycle ergometer. *Canadian Journal of Sports Science, 12(4): pp. 225-228.*

Springer, C., Barstow, T.J., Wasserman, K., & Cooper, D.M. (1991). *Oxygen uptake and heart rate responses during hypoxic exercise in children and adults, 23(1): pp. 71-79.*

Telford, R.D., Hooper, L.A., & Chennells, M.H. (1980). Calibration and comparison of air-braked and mechanically-braked bicycle ergometers. *Australian Journal of Sports Medicine, 12(2):40-46.*

Thompson, C.W. (1985). *Manual of Structural Kinesiology.* St. Louis, Missouri, USA: Times Mirror/Mosby College Publishing.

Vander, A.J., Sherman, J.H., & Luciano, D.S. (1985). *Human Physiology - The Mechanisms of Body Function.* (4th Edition). USA: McGraw-Hill, Inc.

Walz, T. (1984). Shimano's Biopace - not just another elliptical chainring. *Bicycling, 25(2):142,144.*

Watters, P. (Ed.). (1987). *Coaching Manual Level I.* Australian Cycling Federation.

Weir, W.J. (1987). *Electronic Circuit Fundamentals.* New Jersey, USA: Prentice Hall, Inc.

Welbergen, E., & Clijsen, L.P.V.M. (1990). The influence of body position on maximal performance in cycling. *European Journal of Applied Physiology, 61:138-142.*

Wendell, S.N., Gladden, B.L., Barklay, J.K., & Wilson, B.A. (1980). Exercise efficiency: validity of base-line subtractions. *Journal of Applied Physiology: Respiration and Environmental Exercise Physiology, 48(3):518-522.*

Whipp, B.J., & Wasserman, K. (1969). Efficiency of muscular work. *Journal of Applied Physiology, 26(5):644-648.*

Wilkie, D.R. Man as a source of mechanical power. *Ergonomics, 3: pp 1-8.*

Wilmore, J.H., Constable, S.H., Stanforth, P.R., Buono, M.J., Tsao, Y.W., Roby, F.B. (Jr.), Lowdon, B.J., & Ratliff, R.A. (1982). Mechanical and physiological calibration of four cycle ergometers. *Medicine and Science in Sports and Exercise, 14(4):322-325.*

Wilson, S.S. (1973). Bicycle technology. *Scientific American, 228: 81-91.*

Yeh, M.P., Gardner, R.M., Adams, T.D., Yanowitz, F.G., & Crapo, R.O. (1983). "Anaerobic threshold": problems of determination and validation. *Journal of Applied Physiology: Respiration and Environmental Exercise Physiology, 55(4):1178-1186.*

Yoshida, T., Nagata, A., Muro, M. Takeuchi, N. & Suda, Y. (1981). The validity of anaerobic threshold determination by a Douglas bag method compared with arterial blood lactate concentration. *European Journal of Applied Physiology, 46:423-430.*

Zommers, A. (1987). *A quantitative analysis of the ankle action in cycling, comparing the traditional "toes down" technique with the theoretical "ankling" technique.* Deakin University, Melbourne, Australia: (unpublished).

Zommers, A., Gibbs, M., & Selig, S. (1995a). *Calibration and comparison of bicycle ergometers used in physiological testing.* Hobart, Australia: (Australian Conference of Science and Medicine in Sport).

Zommers, A., Gibbs, M., Selig, S., & Best, R. (1995b). *Design and calibration of bicycle ergometers used in the physiological and biomechanical testing of athletes.* Jyvaskyla, Finland: XVth Congress of the International Society of Biomechanics.

Zommers, A., Naughton, G. A., Carlson, J. S., & Gibbs, M. (1995c). *Physiological efficiency of adolescents measured during the cycling leg of a simulated duathlon.* Hobart, Australia: Australian Conference of Science and Medicine in Sport.

APPENDIX A

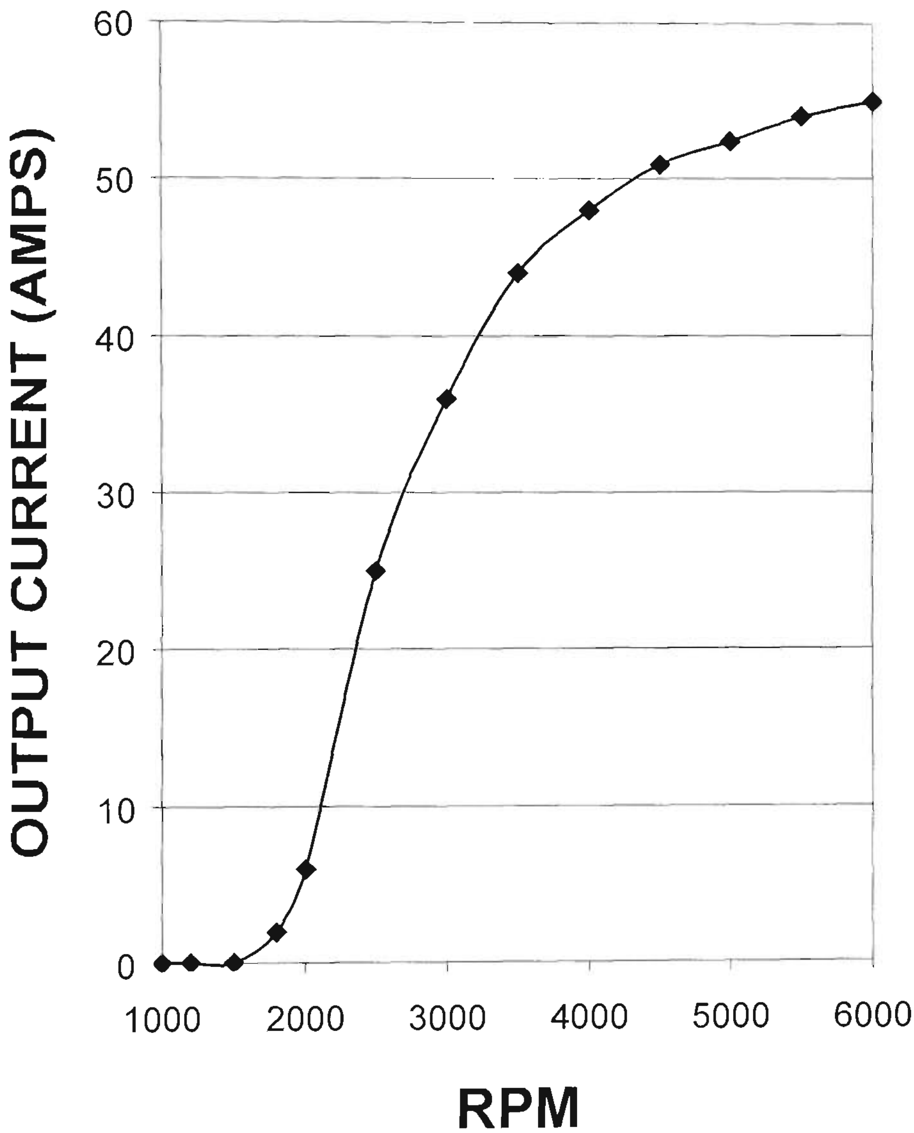


Figure A.1 24 Volt 55 Amp alternator (part no BXU2455) initial performance curve. Robert Bosch (Australasia)

APPENDIX B

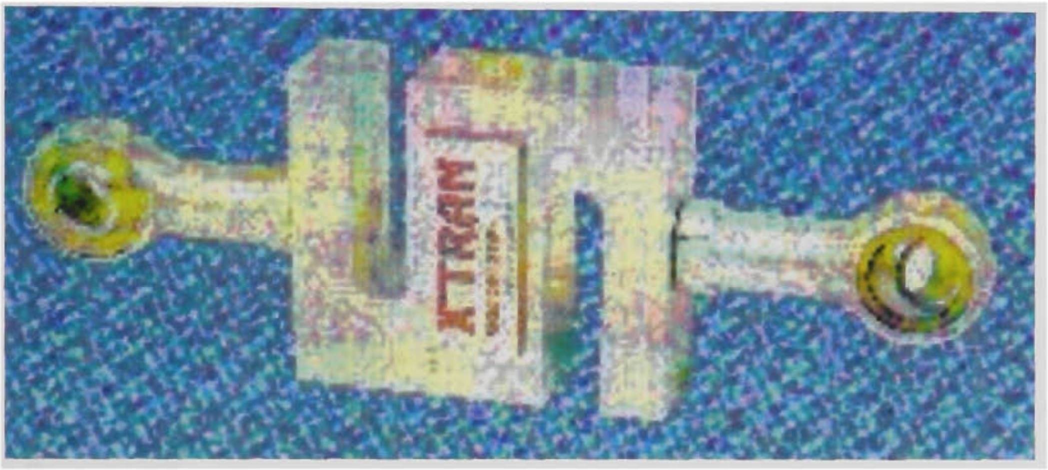


Figure B.1 XTRAN Load Cell. Applied Measurement, Melbourne Australia.

Type	S1
Protection Class	Sealed, Splashproof
Load Direction	Tension & Compression
Rated Capacity	250N
Safe Overload (% of Rated Capacity)	±150
Max Overload (% of Rated Capacity)	±500
Non-Linearity (% Full Scale)	±0.03
Hysteresis (% Full Scale)	±0.02
Non-Repeatability (% Full Scale)	±0.02
Creep (% Full Scale at full load after 20 minutes)	±0.03
Rated Output (nominal)	3 mV/V
Excitation Voltage (AC or DC)	Recommended 10 V Maximum 20 V
Impedance	Input 370 Ohm Output 350 Ohm
Zero Balance (% Full Scale)	±1
Compensated Temperature Range °C	- 15 to +70
Safe Operating Temperature Range °C	-55 to +100
Thermal Sensitivity Coefficient (% of Reading per °C)	±0.0015
Thermal Zero Coefficient (% of Full Scale Output per °C)	±0.0015
Physical Deflection with full load applied	0.013
Electrical Isolation of Bridge from Cell Structure at 100 Volts DC	5000 megohm min.
Cable	3m, 4 conductor shielded
Construction	Proprietary heat treated anodized aluminium

Table B.1 XTRAN S1 Load Cell specifications. Applied Measurement, Melbourne Australia.

APPENDIX C

C.1 STRAIN GAUGE AMPLIFIER (RS STOCK NO. 846-171 AND PRINTED CIRCUIT BOARD (RS STOCK NO. 435-692) RS COMPONENTS LTD. CORBY, NORTHANTS, UK.

The following description has been extracted from technical information supplied by RS Components, Corby, UK.

C.1.1 Description and operation

The strain gauge amplifier is a purpose designed hybrid low noise, low drift dc amplifier in a 24 pin DIL package, specifically configured for resistive bridge measurement.

Foil strain gauges when attached to a specimen, produce very small changes in resistance (typically 0.2 Ω in 120 Ω per microstrain), and are thus normally connected in a Wheatstone bridge. Overall outputs of less than 1mV on a common mode voltage of 5 volts may be encountered, requiring exceptional common mode rejection which cannot be provided by conventional means.

The strain gauge amplifier overcomes the problem of common mode rejection by removing the common mode voltages. This is achieved by controlling the negative bridge supply voltage in such a manner that the voltage at the negative input terminal is always zero. Thus for a symmetrical bridge, a negative bridge supply is generated equal and opposite to the positive bridge supply, hence zero common mode voltage.

The advantages of such a system are:

- No floating power supply needed.
- Bridge supply easily varied with remote sense if necessary.
- Wire remote sense system.
- Freedom from common mode effects.
- Very high stability dc amplifier enables numerous configurations to be assembled
- Low noise.
- High speed (at low gains).

Supply voltage	±2 to ±20V dc
Input offset voltage	200 μV max
Input offset voltage/temperature	0.5 μV/°C
Input offset voltage/supply	3 μV/V max
Input offset voltage/time	0.3 μV/month max
Input impedance	>5MΩ min
Input noise voltage	0.9 μV p-p max
Band width (unity gain)	450 kHz
Output current	5 mA
Output voltage span	±(V-2)V
Closed loop gain (adjustable)	3 to 60,000
Open loop gain	> 120 dB
Common mode rejection ratio	> 120 dB
Bridge supply voltage/temperature	20 μV/°C
Maximum bridge supply current	12 mA
Power dissipation	0.5 W
Warm up time	5 mins
Operating temperature range	-25°C to +85°C

Table C.1 Specifications (At 25°C ambient and ±12V supply unless otherwise stated) RS Components Ltd. Corby, Northants, UK.

APPENDIX D

D.1 GENERAL TECHNICAL DESCRIPTION OF THE ELECTRONICALLY-BRAKED ERGOMETRY SYSTEM NOT INCLUDED IN THE BODY OF THE THESIS

The microcomputer sent raw voltage, current and time for each degree to the PC via a parallel port in binary format that was then stored in a file on the hard disk drive. The raw electrical and time data was also converted to mechanical power using stored regression beta constants. Analysis of the raw data to convert to mechanical power was done using a floating-point processor. Maximum possible accuracy was achieved using double precision floating point arithmetic (64 bit) that is, approximately 15 decimal places (this is the default of a Microsoft Excel spreadsheet). A buffer on the microcomputer stored data temporarily whenever an interrupt occurred in the PC. This was to prevent the loss of any data. Normal "hand shaking" occurred between the microcomputer and the PC; however there was no cyclic redundancy checking because this would have seriously retarded data transfer. Any errors in data transfer were obvious to the operator as errors were also displayed on the computer screen. Any sudden change in power or cadence data would also be detected in the stored data and this could be "smoothed out manually".

Graphing and analysis software was developed. This software had the capabilities of removing/ignoring any rogue values and replacing them with averaged data.

Four status light emitting diodes (LED'S) were included to inform the operator of the status of the microcomputer or alert the operator to any errors.

1. Green LED flashed after a reset had been carried out. The LED flashed until a re-calibration had occurred. The LED went out whenever any further re-calibration was under way that is when the load resistance was changed. The LED stayed on whenever data was being sent from the micro to the PC.

2. A red LED was used to indicate a buffer overflow. When the micro was undergoing a re-calibration, data acquisition was still under way and this data was sent to the buffer. If the "UP" or "DOWN" arrows were pressed too many times before the system could respond with a re-calibration, the re-calibration was further delayed. This could cause the buffer to overflow.

3. Amber LED was used to indicate whenever the tacho count exceeded 50,000 pulses. This was a warning LED only.

4. Another red LED was used to indicate that the tacho pulse count had exceeded 65,536 pulses. The microcomputer, being eight bit, could only handle a pulse count of this amount.

D.1.1 Microcontroller 80C552 - General overview

The following description has been extracted from the Philips data handbook (Philips Components - Signetics Corporation, 1991)

"The 80C552/83C552 (hereafter generically referred to as 8XC552) Single-Chip 8-Bit Microcontroller is manufactured in an advanced CMOS process and is a derivative of the 80C51 microcontroller family. The 8XC552 has the same instruction set as the 80C51. Three versions of the derivative exist:

- 83C552-8k bytes mask programmable ROM
- 80C552-ROMless version of the 83C552
- 87C552-8k bytes EPROM (described in a separate chapter)

The 8XC552 contains a non-volatile 8k × 8 read-only program memory (83C552), a volatile 256 × 8 read/write data memory, five 8-bit I/O ports, one 8-bit input port, two 16-bit timer/event counters (identical to the timers of the 80C51) an additional 16-bit timer coupled to capture and compare latches, a 15-source, two-priority-level, nested interrupt structure, an 8-input ADC, a dual DAC pulse width modulated interface, two serial interfaces (UART and I2PC-bus), a 'watchdog' timer and on-chip oscillator and timing circuits. For systems that require extra capability, the 8XC552 can be expanded using standard TTL compatible memories and logic.

In addition, the 8XC552 has two software selectable modes of power reduction—idle mode and power-down mode. The idle mode freezes the CPU while allowing the RAM, timers, serial ports, and interrupt system to continue functioning. The power-down mode saves the RAM contents but freezes the oscillator, causing all other chip functions to be inoperative.

The device also functions as an arithmetic processor having facilities for both binary and BCD arithmetic plus bit-handling capabilities. The instruction set consists of over 100 instructions: 49 one-byte, 45 two-byte, and 17 three-byte. With a 16MHz (24MHz) crystal, 58% of the instructions are executed in 0.75ms (0.5μs) and 40% in 1.5ms (1μs). Multiply and divide instructions require 3ms (2μs).

Features:

- 80C51 central processing unit
- 8k × 8 ROM expandable externally to 64k bytes

- ROM code protection
- An additional 16-bit timer/counter coupled to four capture registers and three compare registers
- Two standard 16-bit timer/counters
- 256 × 8 RAM expandable externally to 64k bytes
- Capable of producing eight synchronized timed outputs
- A 10-bit ADC with eight multiplexed analog inputs
- Two 8-bit resolution pulse width modulation outputs
- Five 8-bit I/O ports plus one 8-bit input port shared with analog inputs
- I 2 C-bus serial I/O port with byte oriented master and slave functions
- Full-duplex UART compatible with the standard 80C51
- On-chip watchdog timer
- Three speed ranges:
    - 3.5 to 16MHz
    - 3.5 to 24MHz (ROM ROMless only)
    - 3.5 to 30MHz (ROM ROMless only)
- Three operating ambient temperature ranges:
    - P83C552xBx: 0°C to +70°C
    - P83C552xFx: -40°C to +85°C (XTAL frequency max. 24 MHz)

- P83C552xHx: -40°C to +125°C (XTAL frequency max. 16 MHz)" (Philips Components - Signetics Corporation, 1991)

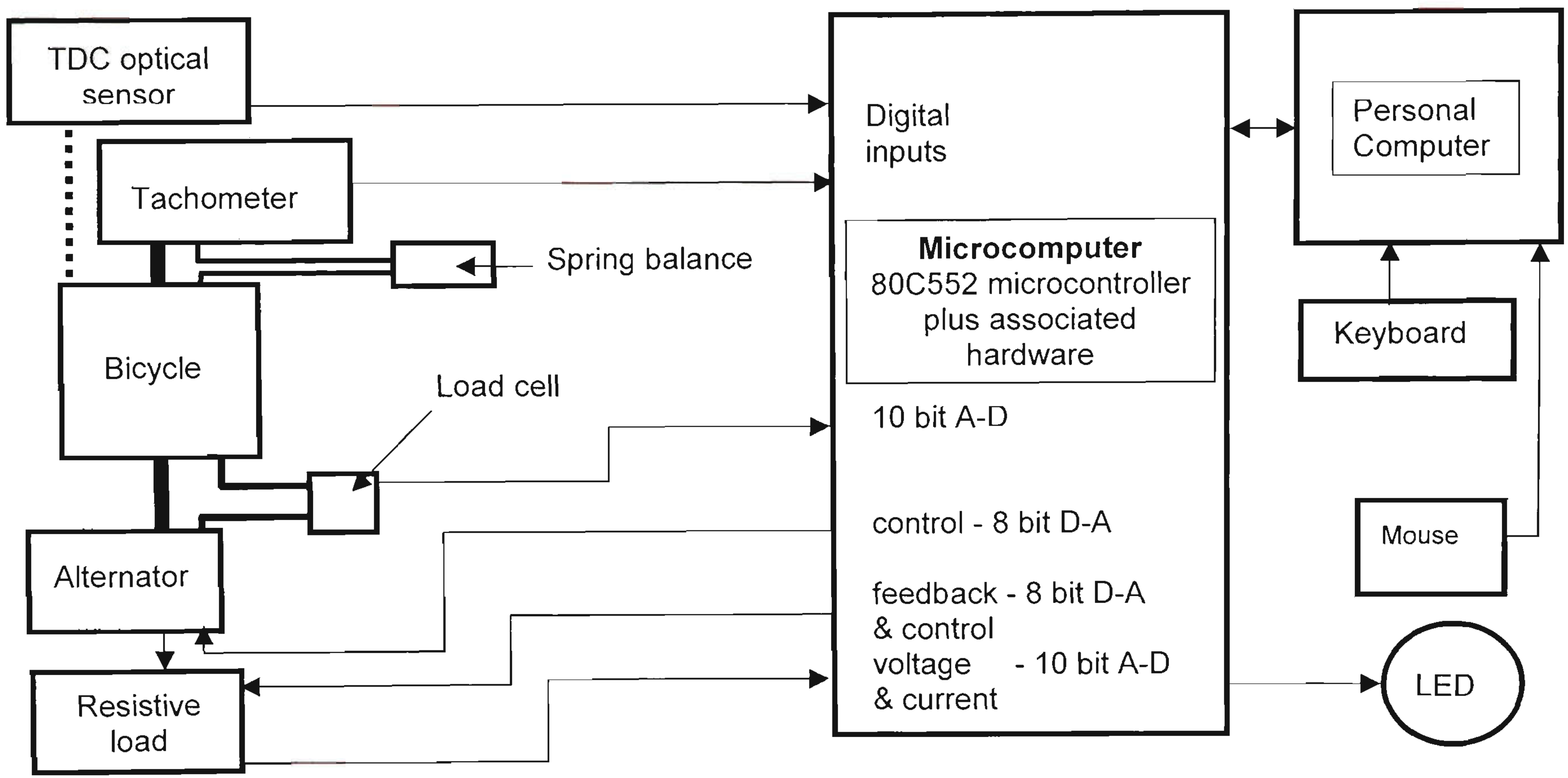


Figure D.1 Block diagram of electronically-braked ergometry system

APPENDIX E

```
[WinLog]
Regression=0
Physiology=0
Normal=1
PhaseShift=0
RaceMode=0

AltForce=11.602
CalForce=11.707
AltOffset=2.297
CalOffset=0.566
AltTrueWeight=11.660
CalTrueWeight=11.660
AltScale=0.993053
CalScale=0.997942

TachoRollerSize=54.88
AltRollerSize=73.8
RaceDistance=12.0
LogFile=Winlog.log

RegStep0=1
RegStep1=16
RegStep2=41
RegStep3=65
RegStep4=87
RegStep5=108
RegStep6=128
RegStep7=146
RegStep8=163
RegStep9=178
RegStep10=192
```

Table E.1 Sample WINLOG lookup table (winlog.ini file) used by the software to look up variables that were entered into the computer during the setup procedure

```
RegStep11=205
RegStep12=216
RegStep13=226
RegStep14=234
RegStep15=241
RegStep16=247
RegStep17=251
RegStep18=254
RegStep19=255

[Regression]
Offset=-61.468361
AltForceFactor=18.971907
AltForceSqdFactor=-1.079604
AltRPMFactor=0.029005
AltRPMSqdFactor=6.02529E-6
DacValueFactor=0.052858
DacValueSqdFactor=2.13276E-5
ElectPowerFactor=1.411764
ElectPowerSqdFactor=1.24072E-5
AltForceTimesDacFactor=-0.010688
AltForceTimesElectPowerFactor=0.014309
AltForceTimesAltRPMFactor=-1.31431E-4
ElectPowerTimesDacFactor=1.34638E-4
AltRPMTimesDacFactor=5.73447E-5
AltRPMTimesElectPowerFactor=-3.98143E-5

[Graph]
GraphOffset=-21.360173
GraphAltForceFactor=1.143152
GraphAltRPMFactor=0.056832
GraphDacValueFactor=0.065762
GraphElectPowerFactor=1.641397
```

Table E.1 (cont) Sample WINLOG lookup table (winlog.ini file) used by the software to look up variables that were entered into the computer during the setup procedure

APPENDIX F

VICTORIA UNIVERSITY OF TECHNOLOGY

STANDARD CONSENT FORM FOR SUBJECTS INVOLVED IN EXPERIMENTS

Variations in “ankling” technique of competitive cyclists: the relationship between physiological and mechanical efficiency.

Please return this Consent Form and retain for your records the Information Sheet that is attached.

CERTIFICATION BY SUBJECT

I,

....of

....

certify that I have the legal ability to give valid consent and that I am voluntarily giving my consent to participate in the experiment entitled:

Variations in "ankling" technique of competitive cyclists: the relationship between physiological and mechanical efficiency.

being conducted at Victoria University of Technology by:
Steve Selig, Alfred Zommers.

I certify that the objectives of the experiment, together with any risks to me associated with the procedures listed hereunder to be carried out in the experiment, have been fully explained to me by:

Steve Selig, Alfred Zommers.

and that I freely consent to participation involving the use on me of these procedures.

Procedures
Risk Factor Assessment
VO_2 max test
Lactate threshold test
Venous catheterisation and blood sampling

Physiology Efficiency test for up to 20 minutes at a workload equivalent to 95% of the workload required to obtain the lactate threshold
Monitoring of ECG, heart rate and rhythm, perceived exertion and lung ventilation.
Biodex test of range of motion of the ankle joint and angle where maximum plantar - flexor force exhibited.

I certify that I have had the opportunity to have my questions answered and that I understand that I can withdraw from this experiment at any time and that this withdrawal will not jeopardise me in any way.

I have been informed that the confidentiality of the information I provide will be safeguarded.

Signed: ....)
)
Witness other than the experimenter:) Date: ....
)
....)

CONSENT FOR SUBJECTS OVER THE AGE OF 35 YEARS

Variations in "ankling" technique of competitive cyclists: the relationship between physiological and mechanical efficiency.

Please return this Consent Form and retain for your records the Information Sheet that is attached.

FREEDOM OF CONSENT

Your permission to perform these tests is voluntary. You are free to deny consent now or *withdraw your consent at any time* (including during the tests) if you so desire.

You and your doctor will need to complete this form and return to us:

MEDICAL BACKGROUND and CONTRA-INDICATIONS TO EXERCISE

(i) details of any medical condition, disability or illness which will reduce this subject's capacity for exercise or make it unsafe for him/her to exercise at the intensity that is proposed in this study.

____

____

(ii) details of exercises that are contra-indicated for each subject

____

____

(iii) prescribed drugs currently being taken

____

(iv) any other information that you think will increase the safety of the participation of this subject in this study.

____

____

SUBJECT'S CONSENT

I have read this form and I understand the procedures involved and the conditions under which the study will be conducted. I consent to participate in this study **WITHOUT/WITH** medical supervision (**delete** inapplicable word).

....

Name of Subject **Signature of Subject** **Date**

DOCTOR'S CONSENT
I have read this form and, in my opinion, it is safe for this subject to participate in the study WITHOUT/WITH medical supervision (**delete** inapplicable word).

....

Name of Doctor **Signature of Doctor** **Date**

CONSENT FOR SUBJECTS <u>UNDER</u> THE AGE OF 35 YEARS

Variations in "ankling" technique of competitive cyclists: the relationship between physiological and mechanical efficiency.

Please return this <u>Consent Form</u> and retain for your records the <u>Information Sheet</u> that is attached.

FREEDOM OF CONSENT

Your permission to perform these tests is voluntary. You are free to deny consent now or *<u>withdraw your consent at any time</u>* (including during the tests) if you so desire.

SUBJECT'S CONSENT

I have read this form and I understand the procedures involved and the conditions under which the study will be conducted. I am under the age of 35 and consent to participate in this study **WITHOUT** medical supervision.

....

Name of Subject **Signature of Subject** **Date**

....

Name of Witness **Signature of Witness** **Date**

APPENDIX G

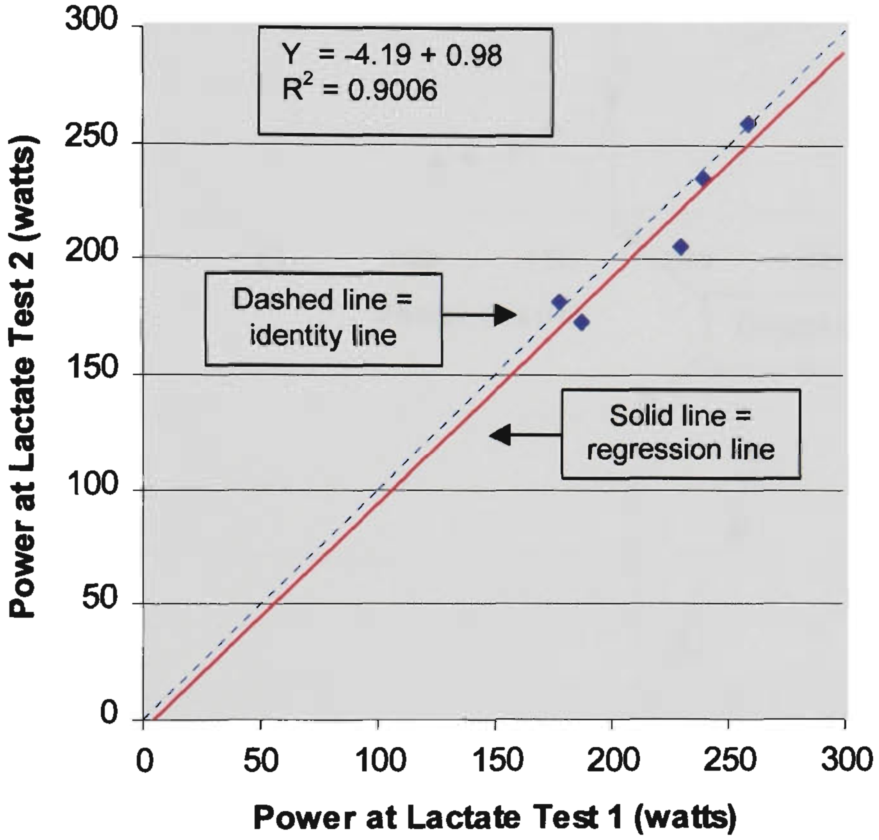


Figure G.1 Reliability of lactate threshold protocol. Test 1 & Test 2 were detections of the lactate threshold on five cyclists (#1, #5, #6, #10, and #19) on two occasions, each one week apart.

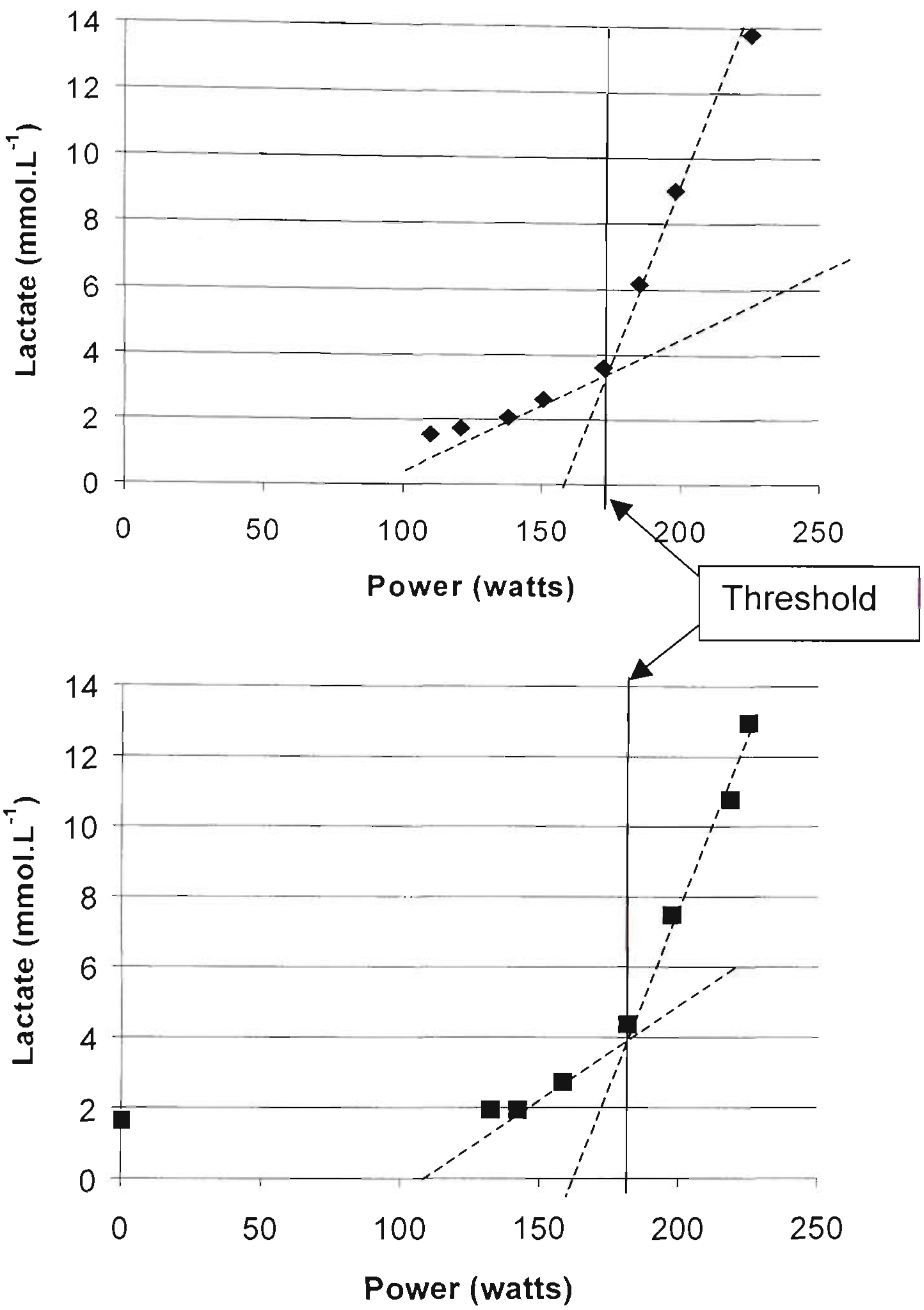


Figure G.2 Lactate threshold detections for subject #1 who underwent duplicate testing (♦ represents Test 1, ■ represents Test 2, one week later)

APPENDIX H

Sample ventilatory threshold graph of one of the cyclists who could-not or would-not be cannulated (cyclist # 4).

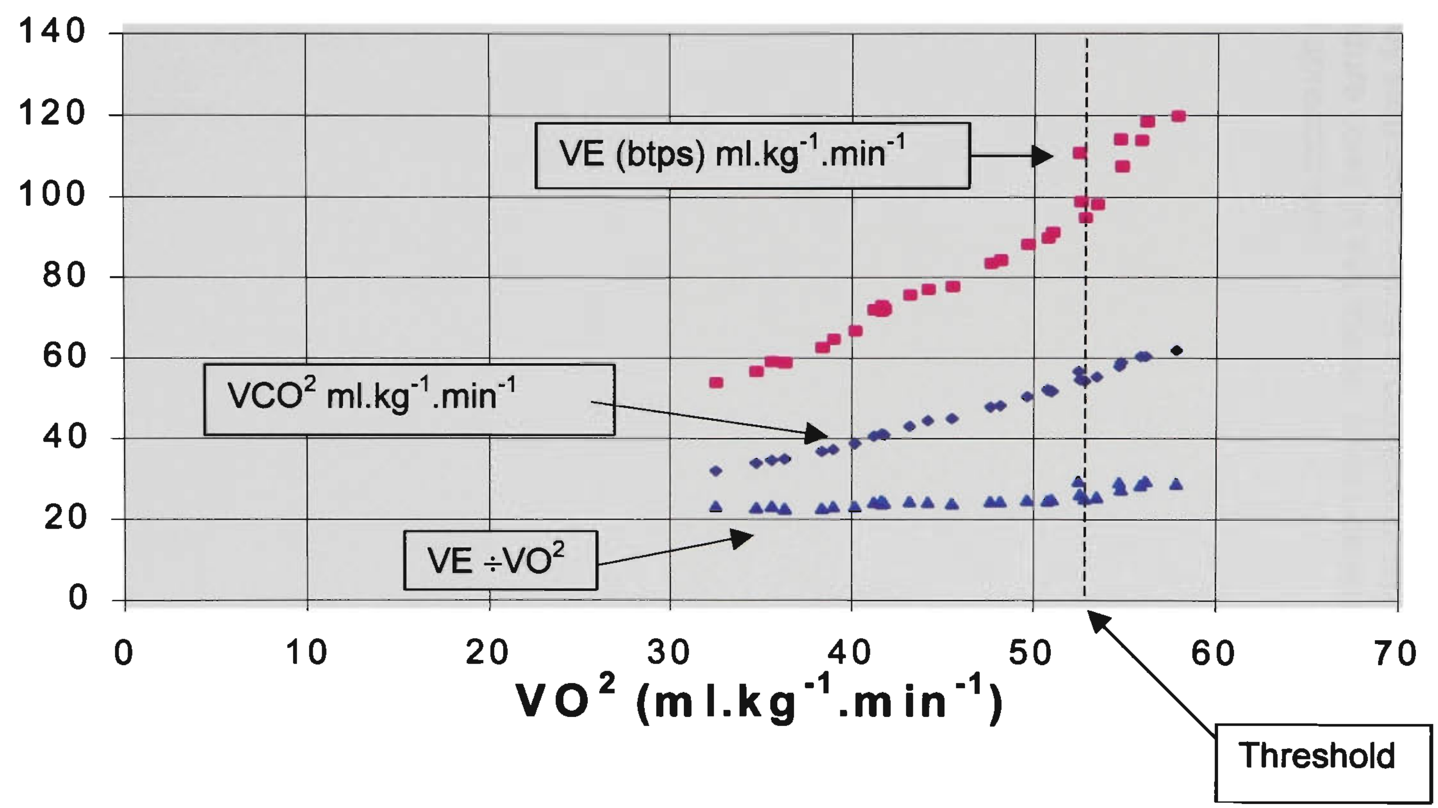


Figure H.1 Sample Ventilatory threshold graph (cyclist # 4)

APPENDIX I

Step by step instructional procedure of the lactate statistics procedure used in this thesis. It was developed using the Microsoft Excel spreadsheet.

Step 1. Enter data from standards in the green and white cell locations.

Name **Tom Jones** **Copyright © Alfred Zommers, Steve Selig, 1996.**
Date **17/03/96**

Cal	tube	Ao	An	An-Ao	An-Ao-Arb	(An-Ao-Arb)*Exp2	(An-Ao-Arb)*Exp3	Predicted [La]
H2O	0.0	0.2180	0.2228	0.0048	0.0000	0.0000	0.0000	0.4624
Cal	1.0	0.3138	0.3683	0.0545	0.0497	0.0025	0.0001	1.1091
Cal	2.5	0.3774	0.5236	0.1462	0.1414	0.0200	0.0028	2.3024
Cal	5.0	0.4544	0.7822	0.3278	0.3230	0.1043	0.0337	4.6655
CS	8.0	0.5122	1.0625	0.5503	0.5455	0.2976	0.1623	7.5608
Cal	10.0	0.6364	1.4086	0.7722	0.7674	0.5889	0.4519	10.4483
Cal	15.0	0.7407	1.9037	1.1630	1.1582	1.3414	1.5536	15.5337
Cal	20.0	0.7934	2.3387	1.5453	1.5405	2.3731	3.6558	20.5084
Cal	25.0	0.9599	2.8451	1.8852	1.8804	3.5359	6.6489	24.9314
Cal	30.0	0.9731	3.2048	2.2317	2.2269	4.9591	11.0434	29.4403

Standards used

Absorption values from Cobas Bio analysis

Note:

(enter 0 in any unused cell locations)

0.0000 is automatically entered in yellow cells

Cal	0.0	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000

Step 2. Input subjects lactate data in the white cell locations.

tube	Ao	An	An-Ao	An-Ao-Arb	(An-Ao-Arb)*Exp2	(An-Ao-Arb)*Exp3	Predicted [La#1]	Predicted [La#2]	average An-Ao-Arb	average [La]
1	0.3277	0.4133	0.0856	0.0808	0.0065	0.0005	1.5138	1.4969	0.0802	1.5054
2	0.3212	0.3976	0.0764	0.0716	0.0051	0.0004	1.3941	1.4019	0.0719	1.3980
3	0.3289	0.4077	0.0788	0.0740	0.0055	0.0004	1.4253	1.4370	0.0745	1.4312
4	0.3518	0.4490	0.0972	0.0924	0.0085	0.0008	1.6648	1.6283	0.0910	1.6465
5	0.3667	0.4875	0.1208	0.1160	0.0135	0.0016	1.9719	2.0109	0.1175	1.9914
6	0.3861	0.5330	0.1469	0.1421	0.0202	0.0029	2.3115	2.3245	0.1426	2.3180
7	0.4111	0.6102	0.1991	0.1943	0.0378	0.0073	2.9908	3.0012	0.1947	2.9960
8	0.4997	0.8665	0.3668	0.3620	0.1310	0.0474	5.1730	5.2276	0.3641	5.2003

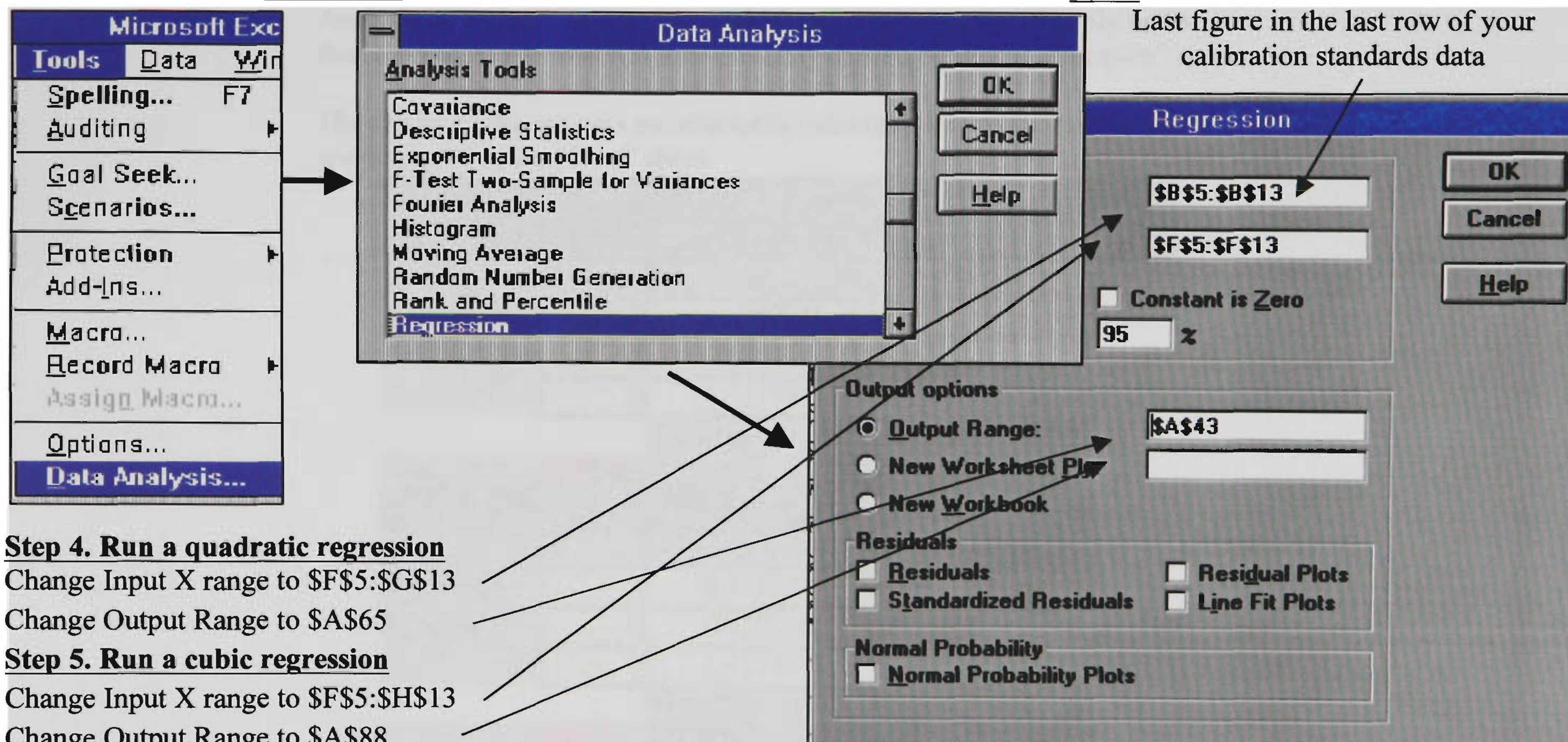
	Ao	An	An-Ao	An-Ao-Arb	(An-Ao-Arb)*Exp2	(An-Ao-Arb)*Exp3
D-1	0.3258	0.4101	0.0843	0.0795	0.0063	0.0005
D-2	0.3187	0.3957	0.0770	0.0722	0.0052	0.0004
D-3	0.3228	0.4025	0.0797	0.0749	0.0056	0.0004
D-4	0.3640	0.4584	0.0944	0.0896	0.0080	0.0007
D-5	0.3854	0.5092	0.1238	0.1190	0.0142	0.0017
D-6	0.3978	0.5457	0.1479	0.1431	0.0205	0.0029
D-7	0.4163	0.6162	0.1999	0.1951	0.0381	0.0074
D-8	0.5019	0.8729	0.3710	0.3662	0.1341	0.0491

Sample data

Duplicate sample data

Step 3. Run a linear regression.

Note:
Last figure in the last row of your calibration standards data



Step 4. Run a quadratic regression
Change Input X range to \$F\$5:\$G\$13
Change Output Range to \$A\$65

Step 5. Run a cubic regression
Change Input X range to \$F\$5:\$H\$13
Change Output Range to \$A\$88

Notes: The Regression results will be displayed in the areas marked on the "Data" sheet Analysis of variance to determine which polynomial is significantly better than the Linear regression is automatically carried out in the "Decision" sheet.

The results of this process automatically calculates the lactate results which are displayed in the "Data" sheet.

Linear vs Quadratic vs Cubic Regression decision making				
F CHANGE TEST - Linear vs Quadratic Polynomial				
	df	SS	MS	F
Regression	1	0.610221	0.610221	4.532689
Residual	6	0.80776	0.134627	
Total	7	1.417981		
		Significance	Regression	
F crit =	5.99	0.05	LINEAR	
F CHANGE TEST - Linear vs Cubic Polynomial				
	df	SS	MS	F
Regression	2	0.978225	0.489112	5.561173
Residual	5	0.439756	0.087951	
Total	7	1.417981		
		Significance	Regression	
F crit =	5.79	0.05	LINEAR	
	DECISION		LINEAR	

APPENDIX J

DETERMINATION OF PLASMA LACTATE

USING THE COBAS BIO SPECTROPHOTOMETER

Preparation of lactate standards

Whether using Lactate or Lactate diluted 1:2 with 0.6M Perchloric acid, care must be taken to accurately measure out the relevant quantities of Lactic acid stock and H_2O. Weighing the quantities, instead of pipetting, has produced the most accurate standards.

A. To make 25 of each lactate standard (1mM, 2.5mM, 5mM, 10mM, 15mM, 20mM, 25mM, 30mM) diluted 1:2 with 0.6M Perchloric acid.

Method:

1. To make 100 mls of 0.05M Lactic acid, mix 0.48gms 96MW Lactic acid (stored in the refrigerator) with 100mls milli-Q H_2O. Unused amount of Lactic acid "stock" to be stored in the refrigerator until required again.

3. Mix the "Lactate" stock with milli-Q water in the following ratios to obtain the required standards.

Standard	Lactate	H_2O
1.0mM	1ml	49ml
2.5mM	1ml	19ml
5.0mM	1ml	9ml
10mM	2ml	8ml
15mM	3ml	7ml
20mM	4ml	6ml
25mM	5ml	5ml
30mM	6ml	4ml

2. **Stir well** before pipetting 100μL into each of the orange caps.

3. Pipette 200μL 0.6M Perchloric acid into each of the orange caps.

4. Store each standard, in clearly labelled plastic resealable bags, in the freezer at -80°C until required for use.

B. To make 25 of each lactate standard (1mM, 2.5mM, 5mM, 10mM, 15mM, 20mM, 25mM, 30mM)

Method:

1. Measure out 25mls of 50mM (0.05M) Lactic acid (stored at -80°C).

2. Mix the "Lactate" stock with milli-Q water in the following ratios to obtain the required standards.

Standard	Lactate	H_2O
1.0mM	1ml	49ml
2.5mM	1ml	19ml
5.0mM	1ml	9ml
10mM	2ml	8ml
15mM	3ml	7ml
20mM	4ml	6ml
25mM	5ml	5ml
30mM	6ml	4ml

3. **Stir well** before pipetting 400μL into each of the orange caps.

4. Store each standard, in clearly labelled plastic resealable bags, in the freezer at -80°C until required for use.

Reagent preparation.
Method:

To make 12mls reagent, which will suffice for < 48 samples (including standards).

NOTE: Due to the high expense of reagent ingredients, make only enough for the number of tests to be carried out. Any excess can be stored in the refrigerator overnight.

1. Weigh out 0.02gms NAD (stored in the refrigerator). NAD is anhydrous so replace the cap quickly and replace in the refrigerator.
2. Add 5mls milli-Q H_2O.
3. Add 7mls glycine-hydrazine buffer (stored in the refrigerator).
4. Add 0.4mls Lactate Dehydrogenase (stored in the refrigerator).
5. Mix gently until NAD fully dissolved.

To make a solution of 1M glycine and 0.8M hydrazine:

1. Weigh 7.505 gms of glycine.
2. Add to 100ml flask.
3. Rinse beaker several times to expel glycine.
4. Nearly fill 100ml flask with milli-Q H_2O.
5. Add 3.9mls hydrazine.
6. Wait until dissolved
7. Fill flask to 100ml milli-Q H_2O.
8. Store in the refrigerator until required.

Setting up the COBAS BIO spectrophotometer, lactate standards, samples and reagent.

1. Clean the H_2O reservoir by siphoning out any residue water.
2. Replace the water with fresh milli-Q H_2O.
3. Select Multi-run and depress the button between the siphon plungers to flush fresh water through the system. Flush for 10 seconds.
4. Select Single-run.
5. Make up a 8mM Lactate or Lactate/PCA standard and a "blank" standard as follows:

   Lactate (adequate for 20 subjects @ 5 µL / subject)
   100 µL 8mM Lactate (from the refrigerator) into one cap. Place into position marked CS in the rotor.
   100 µL fresh milli-Q H_2O into another cap. Place into position marked 1 in the rotor.

   Lactate/PCA (adequate for 20 subjects @ 15 µL / subject)
   100 µL 8mM Lactate (from the refrigerator) mixed with 200 µL 0.6M PCA into one cap. Place into position marked CS in the rotor.
   100 µL fresh milli-Q H_2O mixed with 200 µL 0.6M PCA into another cap. Place into position marked 1 in the rotor.

5. Load the rotor with the standards that have been stored at -80°C with the 1mM Lactate standard in position 2 etc. and then followed by the samples to be tested.

Note: Samples and standards must be unfrozen and close to room temperature prior to testing otherwise the results will be useless. If you don't get close to a straight line with the regression of the absorption figures obtained from the standards against the measured lactate concentrations, then it is likely that the standards were not defrosted adequately. Repeat the test after the samples and standards are at room temperature.

6. Place the reagent into the "boat" and insert carefully into the holding rack.

7. Replace the "reagent sampling" tip carefully but firmly on the nozzle.

8. Set-up the analyser parameters as follows:

1	UNITS	MMOL/L
2	CALCULATION FACTOR	131
3	STANDARD 1 CONC	0
4	STANDARD 2 CONC	0
5	STANDARD 3 CONC	0
6	LIMIT	35
7	TEMPERATURE [DEG . C]	37.0
8	TYPE OF ANALYSIS	5
9	WAVELENGTH [NM]	340
10	SAMPLE VOLUME [UL]	05 (Lactate)
		15
(Lactate/PCA)		
11	DILUENT VOLUME [UL]	10
12	REAGENT VOLUME [UL]	250
13	INCUBATION TIME [SEC]	30
14	START REAGENT VOLUME [UL]	0
15	TIME OF FIRST READING [SEC]	.5
16	TIME INTERVAL [SEC]	300
17	NUMBER OF READINGS	02
18	BLANKING MODE	1
19	PRINTOUT MODE	3

9. **LIST** the parameters and when satisfied that they are correct, press the start button on the analyser. Note: Hold the start button down firmly for several seconds until the analyser starts. The reagent sampling tip should move to the right.

10. Observe that the reagent sampling tip sucks up 250 μL of reagent. If it doesn't, press the stop button immediately. Rectify the fault by replacing the tip. Press Button 9 on the control panel, followed by Button 1 and press start again. If you have been quick enough, you should only lose one cuvette "space".

11. When the analyser stops, it should make a noticeable noise and the following should print out on the chart paper:

    **COMPUTER NOT READY TO RECEIVE

12. Press "print" twice to print out the results of the test.

13. If there were not enough empty cuvette spaces, lift the caps that have been tested in the rotor (this ensures that they are not sampled again) and run the analysis again on the remaining caps **after** the cuvette has been replaced with a new one. If it is desired to run a duplicate analysis, leave the standards lifted and press down all of the sample caps into position in the rotor. Press start again for another analysis.

14. When finished testing, wash the reagent "boat" and then rinse with milli-Q H_2O and leave to dry upside down on tissue paper.

APPENDIX K

Sample graph (cyclist # 51) of the test-retest reliability of the lactate analysis procedure using the automated Cobas Bio spectrophotometer and sample graphs of the lactate standards and lactate samples collected from one subject (cyclist # 28).

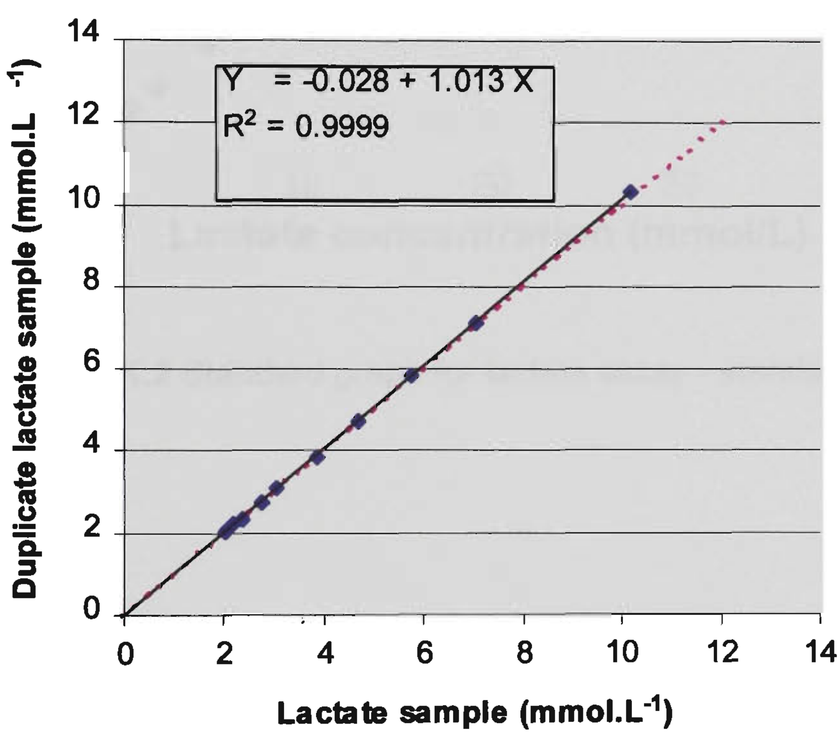


Figure K.1 Test - retest reliability of the lactate assay

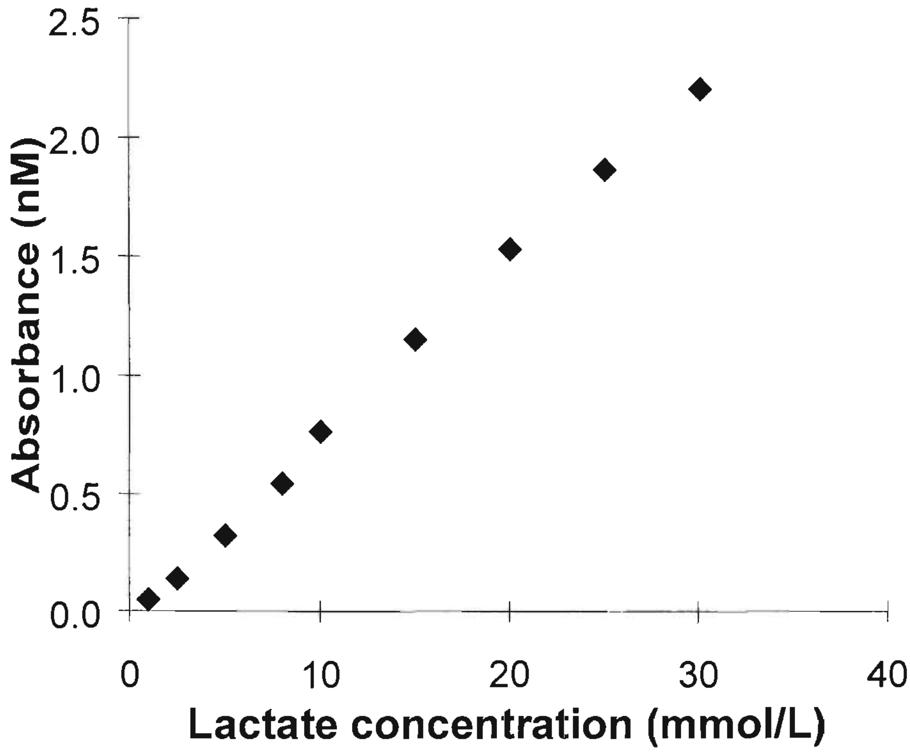


Figure K.2 Standard graph for lactate assay - standards

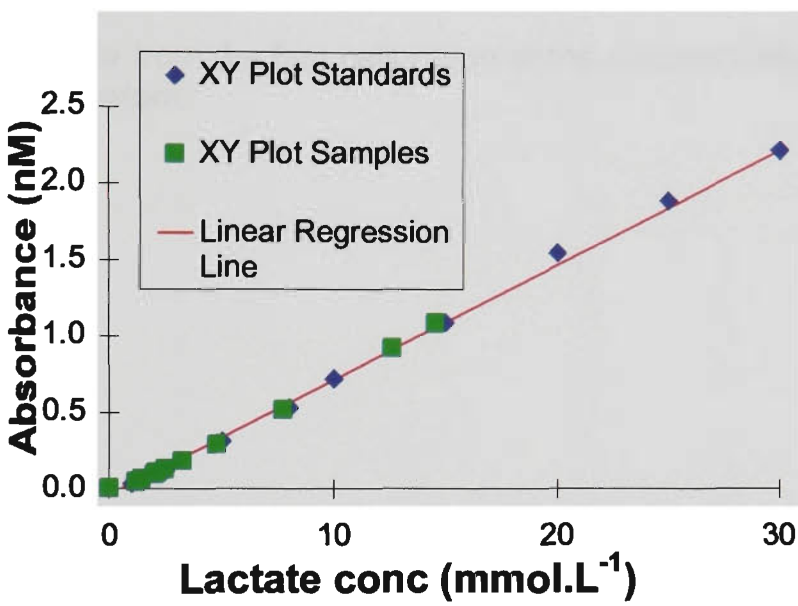


Figure K.3 Standard graphs for lactate assay - samples with linear regression line (cyclist # 28)

APPENDIX L

Statistics data from the first calibration of the electronically-braked ergometry system.

R	R Square	Adjusted R Square	Std. Error of the Estimate
.996	.992	.992	9.9028

Predictors: (Constant), Elec.Power, CADENCE, Tacho.Count, DACVALUE, Alt.Force
Dependent Variable: CALPOWER

Table L.1 First calibration of electronically-braked ergometry system - Model summary

	Sum of Squares	df	Mean Square	F	Sig.
Regression	33387140.666	5	6677428.133	68090.929	.000
Residual	253599.552	2586	98.066		
Total	33640740.218	2591			

Predictors: (Constant), Elec.Power, CADENCE, Tacho.Count, DACVALUE, Alt.Force
Dependent Variable: CALPOWER

Table L.2 First calibration of electronically-braked ergometry system - Anova table

	Unstandardized Coefficients		Standardized Coefficients			Collinearity Statistics	
	B	Std. Error	Beta	t	Sig.	Tolerance	VIF
(constant)	-86.760	1.938		-44.778	.000		
elec.power	1.549	.006	.770	240.243	.000	.284	3.521
cadence	1.331	.010	.297	137.010	.000	.622	1.608
tacho.count	2.196E-03	.000	.134	68.021	.000	.746	1.340
dacvalue	4.977E-02	.004	.037	13.340	.000	.388	2.576
alt.force	.840	.187	.008	4.505	.000	.966	1.035

Dependent Variable: CALPOWER

Table L.3 First calibration of electronically-braked ergometry system - Beta Coefficients

Dimension	Eigenvalue	Condition Index	Variance Proportions (Constant)	Elec.Power	CADENCE	Tacho.Count	DACVALUE	Alt.Force
1	5.176	1.000	.00	.00	.00	.00	.00	.00
2	.574	3.002	.00	.22	.00	.00	.02	.00
3	.153	5.808	.00	.16	.08	.00	.44	.00
4	6.483E-02	8.936	.00	.07	.40	.27	.08	.00
5	2.383E-02	14.739	.00	.18	.24	.31	.18	.62
6	7.363E-03	26.513	1.00	.37	.28	.41	.28	.37

a Dependent Variable: CALPOWER

Table L.4 First calibration of electronically-braked ergometry system - Collinearity Diagnostics

APPENDIX M

Statistics data from the second calibration of the electronically-braked ergometry system.

R	R Square	Adjusted R Square	Std. Error of the Estimate
.998	.997	.997	9.6827

a Predictors: (Constant), DAC, CAD, TACHO, ALTF, ELECPOW
b Dependent Variable: MECHPOW

Table M.1 Second calibration of electronically-braked ergometry system - Model summary

	Sum of Squares	df	Mean Square	F	Sig.
Regression	93219878.871	5	18643975.774	198857.686	.000
Residual	304892.461	3252	93.755		
Total	93524771.332	3257			

a Predictors: (Constant), DAC, CAD, TACHO, ALTF, ELECPOW
b Dependent Variable: MECHPOW

Table M.2 Second calibration of electronically-braked ergometry system - ANOVA table

	Unstandardized Coefficients		Standardized Coefficients			Collinearity Statistics	
	B	Std. Error	Beta	t	Sig.	Tolerance	VIF
(Constant)	-142.685	2.255		-63.286	.000		
ALTF	1.357	.186	.007	7.275	.000	.977	1.024
TACHO	3.458E-03	.000	.167	122.986	.000	.543	1.842
CAD	1.469	.012	.169	124.372	.000	.544	1.839
ELECPOW	1.674	.005	.855	358.086	.000	.176	5.691
DAC	3.320E-02	.004	.016	7.492	.000	.228	4.384

a Dependent Variable: MECHPOW

Table M.3 Second calibration of electronically-braked ergometry system - Beta Coefficients

			Variance Proportions					
Dimension	Eigenvalue	Condition Index	(Constant)	ALTF	TACHO	CAD	ELECPOW	DAC
1	5.443	1.000	.00	.00	.00	.00	.00	.00
2	.394	3.717	.00	.00	.00	.00	.12	.01
3	9.162E-02	7.708	.00	.00	.03	.02	.16	.30
4	4.973E-02	10.462	.00	.00	.27	.24	.00	.00
5	1.762E-02	17.574	.01	.82	.12	.17	.12	.14
6	4.029E-03	36.758	.99	.18	.58	.57	.59	.54

a Dependent Variable: MECHPOW

Table M.4 Second calibration of electronically-braked ergometry system - Collinearity Diagnostics

APPENDIX N

Statistics data from the second calibration of the electronically-braked ergometry system (alternator RPM added as an independent variable).

R	R Square	Adjusted R Square	Std. Error of the Estimate
.999	.998	.998	6.6342

a Predictors: (Constant), ALTRPM, DAC, ALTF, ELECPOW
b Dependent Variable: MECHPOW

Table N.1 Second calibration of electronically-braked ergometry system (alternator RPM added as an independent variable) - Model Summary

	Sum of Squares	df	Mean Square	F	Sig.
Regression	93381596.629	4	23345399.157	530419.005	.000
Residual	143174.703	3253	44.013		
Total	93524771.332	3257			

a Predictors: (Constant), ALTRPM, DAC, ALTF, ELECPOW
b Dependent Variable: MECHPOW

Table N.2 Second calibration of electronically-braked ergometry system (alternator RPM added as an independent variable) - ANOVA table

	Unstandardized Coefficients		Standardized Coefficients			Collinearity Statistics	
	B	Std. Error	Beta	t	Sig.	Tolerance	VIF
(Constant)	-21.360	1.083		-19.722	.000		
ALTF	1.143	.128	.006	8.951	.000	.978	1.023
ELECPOW	1.641	.003	.838	505.776	.000	.171	5.839
DAC	6.576E-02	.003	.031	21.338	.000	.221	4.518
ALTRPM	5.683E-02	.000	.244	218.156	.000	.377	2.655

a Dependent Variable: MECHPOW

Table N.3 Second calibration of electronically-braked ergometry system (alternator RPM added as an independent variable) - Beta Coefficients

Dimension	Eigenvalue	Condition Index	Variance Proportions (Constant)	ALTF	ELECPOW	DAC	ALTRPM
1	4.496	1.000	.00	.00	.00	.00	.00
2	.356	3.551	.00	.01	.12	.01	.00
3	.122	6.072	.00	.00	.07	.17	.10
4	1.861E-02	15.543	.00	.53	.38	.41	.44
5	7.282E-03	24.846	.99	.46	.43	.41	.45

a Dependent Variable: MECHPOW

Table N.4 Second calibration of electronically-braked ergometry system (alternator RPM added as an independent variable) - Collinearity Diagnostics

APPENDIX O

Statistics data from the second calibration of the electronically-braked ergometry system (cubic polynomial - stepwise method).

R	R Square	Adjusted R Square	Std. Error of the Estimate
.999	.999	.999	5.5635

Predictors: (Constant), ELECPOW, ALTRPM, DAC2, ALTF_RPM, ALTF3, EP3, EP_ALTF, ALTF_DAC, ALTF, EP_RPM, ALTRPM2, ALTRPM3, DAC3
Dependent Variable: MECHPOW

Table O.1 Second calibration of electronically-braked ergometry system (cubic polynomial - stepwise method) - Model Summary

	Sum of Squares	df	Mean Square	F	Sig.
Regression	93424362.391	13	7186489.415	232180.237	.000
Residual	100408.941	3244	30.952		
Total	93524771.332	3257			

Predictors: (Constant), ELECPOW, ALTRPM, DAC2, ALTF_RPM, ALTF3, EP3, EP_ALTF, ALTF_DAC, ALTF, EP_RPM, ALTRPM2, ALTRPM3, DAC3
Dependent Variable: MECHPOW

Table O.2 Second calibration of electronically-braked ergometry system (cubic polynomial - stepwise method) - ANOVA table

	Unstandardized Coefficients		Standardized Coefficients			Collinearity Statistics	
	B	Std. Error	Beta	t	Sig.	Tolerance	VIF
(Constant)	79.541	5.956		13.355	.000		
ELECPOW	2.007	.023	1.025	86.876	.000	.002	420.607
ALTRPM	-3.830E-02	.004	-.164	-9.538	.000	.001	897.052
DAC2	-7.370E-04	.000	-.100	-7.529	.000	.002	535.073
ALTF_RPM	5.778E-03	.000	.182	28.805	.000	.008	120.738
ALTF3	-6.602E-02	.010	-.040	-6.320	.000	.008	122.222
EP3	7.178E-07	.000	.031	15.269	.000	.081	12.372
EP_ALTF	-3.591E-02	.002	-.123	-16.306	.000	.006	171.360
ALTF_DAC	2.286E-02	.002	.073	13.791	.000	.012	84.749
ALTF	-5.369	1.197	-.029	-4.486	.000	.008	127.726
EP_RPM	-5.932E-05	.000	-.100	-11.762	.000	.005	220.464
ALTRPM2	2.343E-05	.000	.515	15.889	.000	.000	3173.047
ALTRPM3	-2.796E-09	.000	-.257	-15.003	.000	.001	889.786
DAC3	1.631E-06	.000	.058	6.633	.000	.004	231.509

a Dependent Variable: MECHPOW

Table O.3 Second calibration of electronically-braked ergometry system (cubic polynomial - stepwise method) - Beta Coefficients

APPENDIX P

Statistics data from the second calibration of the electronically-braked ergometry system (quadratic polynomial - enter method).

R	R Square	Adjusted R Square	Std. Error of the Estimate
.999	.999	.999	5.7575

a Predictors: (Constant), ALTF2, DAC, ALTRPM, EP2, DAC2, EP_ALTF, ALTRPM2, EP_RPM, ALTF_RPM, ALTF_DAC, EP_DAC, DAC_RPM, ALTF, ELECPOW
b Dependent Variable: MECHPOW

Table P.1 Second calibration of electronically-braked ergometry system (quadratic polynomial - enter method) - Model Summary

	Sum of Squares	df	Mean Square	F	Sig.
Regression	93417268.614	14	6672662.044	201292.055	.000
Residual	107502.718	3243	33.149		
Total	93524771.332	3257			

a Predictors: (Constant), ALTF2, DAC, ALTRPM, EP2, DAC2, EP_ALTF, ALTRPM2, EP_RPM, ALTF_RPM, ALTF_DAC, EP_DAC, DAC_RPM, ALTF, ELECPOW
b Dependent Variable: MECHPOW

Table P.2 Second calibration of electronically-braked ergometry system (quadratic polynomial - enter method) - ANOVA table

	Unstandardized Coefficients		Standardized Coefficients			Collinearity Statistics	
	B	Std. Error	Beta	t	Sig.	Tolerance	VIF
(Constant)	27.422	8.096		3.387	.001		
ALTF	1.562	2.412	.008	.648	.517	.002	484.303
ELECPOW	1.985	.056	1.014	35.171	.000	.000	2344.450
DAC	-.104	.031	-.049	-3.341	.001	.002	612.898
ALTRPM	1.462E-02	.002	.063	6.457	.000	.004	266.401
EP2	6.587E-04	.000	.096	7.598	.000	.002	447.751
DAC2	3.361E-05	.000	.005	.500	.617	.004	235.619
ALTRPM2	1.795E-06	.000	.039	5.786	.000	.008	131.104
EP_ALTF	-3.607E-02	.003	-.123	-11.185	.000	.003	343.192
EP_DAC	-3.701E-04	.000	-.046	-2.079	.038	.001	1393.686
EP_RPM	-8.220E-05	.000	-.139	-7.093	.000	.001	1086.808
ALTF_DAC	2.332E-02	.003	.074	7.888	.000	.004	251.663
ALTF_RPM	5.613E-03	.000	.177	22.033	.000	.005	181.842
DAC_RPM	2.516E-05	.000	.033	2.633	.008	.002	432.541
ALTF2	-1.151	.201	-.076	-5.739	.000	.002	493.339

a Dependent Variable: MECHPOW

Table P.3 Second calibration of electronically-braked ergometry system (quadratic polynomial - enter method) - Beta Coefficients

APPENDIX Q

Statistics data from the third calibration of the electronically-braked ergometry system (quadratic polynomial - enter method).

R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
				R Square Change	F Change	df1	df2	Sig. F Change
1.000	.999	.999	3.8146	.999	114061.063	14	1136	.000

a Predictors: (Constant), RPM-EP, ALTF, DAC2, RPM2, EP2, RPM-DAC, ALTRPM, AF-DAC, EP-DAC, ALTF2, DAC, AF-EP, AF-RPM, EPOW

Table Q.1 Third calibration electronically-braked ergometry system (quadratic polynomial - enter method) - Model Summary

	Sum of Squares	df	Mean Square	F	Sig.
Regression	23235507.851	14	1659679.132	114061.063	.000
Residual	16529.703	1136	14.551		
Total	23252037.554	1150			

a Predictors: (Constant), RPM-EP, ALTF, DAC2, RPM2, EP2, RPM-DAC, ALTRPM, AF-DAC, EP-DAC, ALTF2, DAC, AF-EP, AF-RPM, EPOW
b Dependent Variable: MPOW

Table Q.2 Third calibration electronically-braked ergometry system (quadratic polynomial - enter method) - ANOVA Table

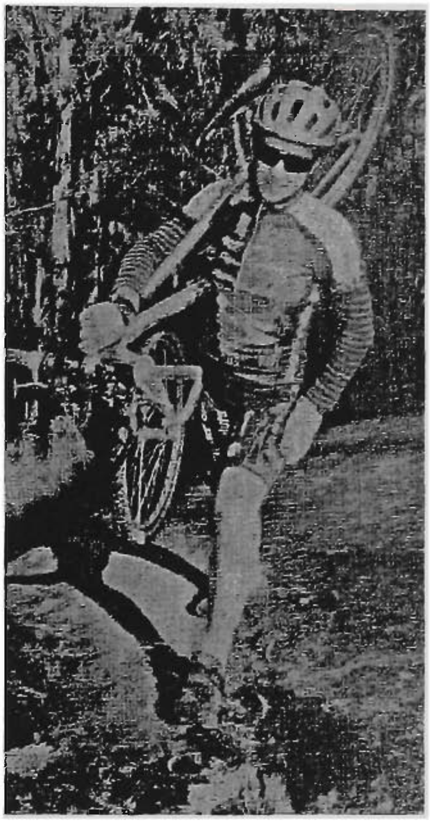
	Unstandardized Coefficients		Standardized Coefficients			Collinearity Statistics	
	B	Std. Error	Beta	t	Sig.	Tolerance	VIF
(Constant)	-61.468	7.185		-8.555	.000		
ALTF	18.972	2.312	.138	8.207	.000	.002	448.967
ALTRPM	2.900E-02	.003	.148	11.178	.000	.004	278.895
EPOW	1.412	.062	.764	22.672	.000	.001	1814.956
DAC	5.286E-02	.032	.031	1.639	.102	.002	559.519
ALTF2	-1.080	.228	-.101	-4.741	.000	.001	729.482
RPM2	6.025E-06	.000	.155	10.585	.000	.003	341.611
DAC2	2.133E-05	.000	.003	.294	.768	.004	223.054
EP2	1.241E-05	.000	.002	.124	.901	.004	283.115
AF-DAC	-1.069E-02	.005	-.042	-2.244	.025	.002	563.957
AF-EP	1.431E-02	.006	.058	2.441	.015	.001	890.312
AF-RPM	-1.314E-04	.001	-.006	-.205	.838	.001	1488.257
EP-DAC	1.346E-04	.000	.018	.682	.496	.001	1065.869
RPM-DAC	5.734E-05	.000	.081	4.955	.000	.002	428.511
RPM-EP	-3.981E-05	.000	-.064	-2.925	.004	.001	775.892

a Dependent Variable: MPOW

Table Q.3 Third calibration electronically-braked ergometry system (quadratic polynomial - enter method) - Beta Coefficients

APPENDIX R

R.1 NEWSPAPER ARTICLE RELATING TO ONE OF THE ELITE CYCLISTS WHO VOLUNTEERED FOR THE STUDY



SHOULD you, perhaps, be out walking one day, negotiating the steep downward slope of a muddy bridle path on the hills around Olinda, and hear from behind you a rasping, growing growl, do not panic.

In all likelihood, the noise will signal the imminent presence of ...., .., out for a ride on her bicycle. Stand to the edge of the path; any second now, she'll pass you, doing 80 kmh.

Many people own mountain bikes: sturdy, no-nonsense contraptions that look like BMX bicycles after six months in a gym and a good course of steroids. Some people even take them for a genteel treadle across a field or over a foothill. Ms .... literally takes them up and down mountains — over the rocks, up dramatic inclines where even hikers think twice — against the clock, as fast as possible, pushing the machine to the limits of its 24 gears.

If her speedy passing disturbs you, do not think ill of her. She is doing it not for her, but for her country. Mountain biking makes its debut as an Olympic medal event in Atlanta. Ms .... is on the squad.

"Downhill mountain biking is a real thrills-and-spills event," she said. "Everyone loves to see someone fall off and the bike flying through the air. It's a really good spectator sport . . .

"You can injure yourself doing it. There have been a couple of people killed in America, doing downhill ...

"The most common injuries are broken collar bones. I've been really lucky: the only accident I've had, I broke my finger, and that was on the road. I like downhilling, but I haven't got that complete absence of fear.

"You need to have that frontal lobotomy attitude of not caring. You see, I worry, because my main event is cross country. If I have an accident, that could be the end of my Olympics hopes. I'm pretty conservative, but, of course, if there's a downhill in a cross country race then, you just go for it — no brakes — because you can pick up a lot of time."

You can also, of course, pick up a lot of multiple fractures.

Although she took up the sport only two years ago — switching from equestrian competitions — Ms .... seems born to it. Blithely she'll tell you her winter training regime includes 700 to 800 kilometres a week on a road bike, including pedalling to and from central Melbourne to her job at .... three days out of every seven.

In spring and summer, the racing season, she gets really serious.

"The national series starts each October," she said. "That's when everyone starts to get really fit. You cut back on the kilometres and start to do the really intense, lung busting stuff.

"You still do a couple of maybe 150 or 200 kilometre rides in a day, and then do the more uphill stuff. There's a place in Ferntree Gully called One Tree Hill. It's a notorious climb, takes about 25 minutes in your smallest gear. You do three of those. It trains your heart to become much stronger and your legs to put up with lactic acid burn."

All in all, it's a long way from a sedate pedal around the park and a free-wheel run down the hill on the way home.

But, then again, the Atlanta Olympics are a long way from a quick sprint up and down Mt Buller or around Thredbo. Such is the growth of the sport that the Olympic administrators allowed it to debut with full medal status, rather than as a demonstration event.

When the final team is picked from the Australian squad, it will probably comprise five men and two women, most of them from Victoria. .... is in with a good chance.

There can be few people in modern sport who have gone from beginner to Olympic medal prospect in just two years. If you see her barrelling down the path towards you, therefore, don't be offended if she doesn't stop to chat.

APPENDIX S

Comparison of mechanical power output during the efficiency tests and the mechanical power output at the anaerobic threshold subdivided according to gender, age, and type of anaerobic threshold test.

	Mechanical power output (watts)			
	efficiency tests		Anaerobic threshold	
gender	male	female	male	female
Mean	225*	169*	259*	179*
±SD	30	26	47	31
Max	289	211	375	254
Min	153	135	162	142
n	29	11	29	11
age	Under 35	Over 35	Under 35	Over 35
Mean	207*	214*	234*	243*
±SD	41	35	63	45
Max	271	162	375	300
Min	135	289	142	170
n	24	16	24	16
AEth type	Lactate	Ventilatory	Lactate	Ventilatory
Mean	208*	216*	233*	253*
±SD	37	45	56	57
Max	271	289	375	318
Min	135	143	142	155
n	32	8	32	8

Table S.1 Comparison of mechanical power output during the efficiency tests and the mechanical power output at the anaerobic threshold subdivided according to gender, age, and type of anaerobic threshold test

* Significant correlation between mechanical power output during the efficiency tests and the mechanical power output at the anaerobic threshold subdivided (partial correlation) according to gender, age, and type of anaerobic threshold test.

APPENDIX T

gender	Male	Female
Mean	87.8*	94.6*
±SD	9.3	6.8
Max	100.9	106.2
Min	67.7	83.1
n	29	11

age	Under 35	Over 35
Mean	90.3	89.0
±SD	9.5	8.7
Max	106.2	99.7
Min	69.7	67.7
n	24	16

AEth type	Lactate	Ventilatory
Mean	91	86
±SD	9	10
Max	106	100
Min	70	68
n	32	8

Table T.1 Mechanical power output during the efficiency tests expressed as a percentage of the power output measured at the anaerobic threshold

* Significant difference ($t = 2.27$, $p = 0.03$)

APPENDIX U

gender	Male	Female
Mean	0.96	0.96
±SD	0.02	0.02
Max	1.0	1.01
Min	0.92	0.92
n	29	12

age	Under 35	Over 35
Mean	0.95*	0.97*
±SD	0.02	0.02
Max	1.0	1.01
Min	0.92	0.93
n	25	16

AEth type	Lactate	Ventilatory
Mean	0.96#	0.94#
±SD	0.02	0.02
Max	1.01	0.98
Min	0.93	0.92
n	33	8

Table U.1 Comparison of the RER between the males and females, the two age groups, and the two types of anaerobic threshold tests

* Significant difference ($t = -2.25$, $p = 0.03$)

Significant difference ($t = 2.82$, $p < 0.01$)

APPENDIX V

gender	Male	Female
Mean	-2.25	-4.875
±SD	26.31	41.886
Max	35	85
Min	-60	-55
n	26	12

age	Under 35	Over 35
Mean	-6.91	2.80
±SD	34.93	25.29
Max	85	35
Min	-60	-35
n	23	15

Table V.1 Horizontal distance between a plumb line dropped from the front of the knee to the centre of the pedal with the crank at 90° for the males and females and the two age groups. A negative value indicates that the front of the knee was behind the pedal.

APPENDIX W

gender	Male	Female
Mean	65.3	51.3
±SD	21.2	23.4
Max	100	95
Min	30	5
n	29	12

age	Under 35	Over 35
Mean	59.3	64.1
±SD	25.4	17.5
Max	100	95
Min	5	40
n	25	16

Table W.1 Horizontal distance measured using a plumb line between the front of the bicycle seat and the centre of the crank