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The effect of cycling on muscle activation in the running leg of an Olympic distance triathlon

Tamika Heiden
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EDITH COWAN UNIVERSITY
School of Biomedical and Sports Science

**THE EFFECT OF CYCLING ON MUSCLE
ACTIVATION IN THE RUNNING LEG OF AN
OLYMPIC DISTANCE TRIATHLON**

THESIS

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Degree: Bachelor of Science with Honours

Date of Submission: 16 November 2001

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ABSTRACT

Anecdotal reports from triathletes, highlight the transition from cycling to running as the most difficult due to the change from non-weight bearing cycling activity to weight bearing running activity. The aim of this study was to examine the effects of prior cycling on activation of lower limb muscles in running during an Olympic distance triathlon. Ten elite level triathletes underwent two conditions; a 40km ride followed by a 2km run and a 10km run followed by a 2km run, at their Olympic distance race pace. Testing was carried out in the field with at least one week between tests. EMG data from selected lower limb muscles in addition to accelerometer data to determine heel strike and toe-off were collected using a portable data logger. The vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), gastrocnemius (GS) and gluteus maximus (GM) were analysed due to their ease of measurement via surface electrodes and because of their important role in both running and cycling. Data was processed to provide both the level of activation and time of activation variables for all the abovementioned muscles in both conditions. RMS processing of the EMG signal was used to evaluate the level of activation with the signal being normalised to each subject's maximum and represented as a percentage. Time of activation was calculated from the ensemble average of the six measured strides in each of the three sections of time using a 10% of maximum activation threshold detector. A repeated measures ANOVA with two within subject variables was used to evaluate the statistical significances ($p < 0.05$) between the conditions and across time for each muscle for both the flight and stance phases. The results revealed increases in level and time of activation during stance for the VL and VM and are seemingly related to the change from a non-weight bearing activity (cycling) to a weight bearing activity (running). A greater demand on these muscles is assumed to occur for stability of the knee joint. During flight it is thought that increases in the level and time of activation of the VL and VM are due to the changes associated with the change from a concentric muscle contraction in cycling to stretch shortening cycle in running. The greater magnitude in both level and time of activation of the BF for the run/run condition are considered to be due to a greater level of fatigue of this muscle during 10km running compared with 40km cycling. Change in the time of activation of the RF may be due to the difficulties in flexing the hip in flight and its synergistic role with the VL and VM in extension of the knee.

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CHAPTER ONE

1.0 INTRODUCTION

1.1 Background

Triathlon, as the name suggests, is a multisport event consisting of a swim, cycle and run leg with short transitional times between them. Success is partially determined by the ability of an athlete to overcome difficulties in the change from one discipline to another, and with haste. This requires a large amount of training for each of the three disciplines.

There are many different and versatile events in triathlon, ranging from short distance through to an ironman event. The newly created Olympic distance triathlon consists of a 1500m swim, 40km cycle and a 10km run. Triathlon is a widely populated sport with many countries participating. Triathlon had its debut in the 2000 Olympics, which was contested by 48 female and 52 male competitors (Deutscher, 2000). The International Triathlon Union (ITU) has been successful in maintaining the inclusion of triathlon for the 2004 Olympics in Athens.

It is proposed that the format of the triathlon; swim, bike, and run, is advantageous toward athletes who can run well off the bike under fatigued conditions (Williams, 2000). Traditionally, drafting during the cycle leg of triathlons has been illegal, however, with the push for a higher profile for the sport of triathlon, drafting during the cycle leg of Olympic distance triathlons was recently legalised to increase spectator value. With the advent of drafting, it has been suggested that these races are dominated by the fastest runners and not necessarily those competent in all three disciplines (Williams, 2000).

Anecdotal reports, from triathletes, highlight the transition from cycling to running as the most difficult. This transition involves both a change of equipment and a change from non-weight bearing cycling activity to weight bearing running activity. This transition requires the coordination of the leg muscles to be maintained whilst shifting from a predominantly concentric type of muscle activity in cycling, to the

stretch shortening activity used in running (Millet, Millet, Hofmann, & Candau, 2000; Quigley & Richards, 1996).

To the investigators best knowledge, studies which have examined muscle activation in running after a cycling bout, have been carried out in controlled laboratory situations. Very few studies have studied the effects of cycling on a subsequent running bout (Hauswirth, Bigard, Berthelot, Thomaidis, & Guezennec, 1996). Several studies, (Hue, Le Gallais, Chollet, Boussana, & Prefaut, 1998; Millet et al., 2000; Millet & Vleck, 2000; Quigley & Richards, 1996) have examined the change in biomechanical variables such as stride rate, stride length, and mechanical work in running after cycling.

Witt (1993) examined muscle activity of the lower extremities in running after cycling, in laboratory conditions, and found significant differences in the activation of selected leg muscles when compared to running alone. However, Witt's study involved a run followed by a bout of cycling and then another run all within the one testing session. Therefore, the results may only be significantly different due to the changes in electromyography (EMG) during muscular fatigue, which was not controlled for in Witt's study.

1.2 Significance of the Study

1.2.1 Whilst other studies have examined the effect of cycling on muscle activation in running, they were carried out in controlled laboratory situations and did not control for fatigue, which is known to affect EMG (Witt, 1993). This study examined two conditions, cycle/run (40km cycle, 2km run) and run/run (10km run, 2km run) undertaken on separate occasions to control for this factor.

1.2.2 The level of activation of the rectus femoris, biceps femoris and vastus lateralis is significantly different between over ground and treadmill running (Wank, Frick, & Schmidtbleicher, 1998) therefore, this study was carried out in the field to provide results that were more closely related to actual triathlon competition.

1.3 Purpose of the Study

- 1.3.1 To examine the change in the muscle activation of the lower extremities during the initial 2km of running after a 40km cycle. This was done by making comparisons of lower limb muscle activation under two different conditions, these being a cycle/run and run/run.
- 1.3.2 To measure, and compare between the abovementioned conditions, the extent of activation changes and the level of change over three measured sections of the 2km run. These measured sections consisted of the first six steps once race pace had been reached, the last six steps of the first kilometre and the final six steps of the second kilometre before speed was reduced.
- 1.3.3 To determine whether the level of muscle activation would become similar to the activation of the run/run condition within the 2km period studied.
- 1.3.4 To justify, on biomechanical grounds, the importance and or relevance of training specifically for the transition from bike to run associated with the sport of triathlon.

1.4 Limitations of the Study

1.4.1 Limitations

- 1.4.1.1 It was assumed that a 10km run would create an equal level of fatigue as a 40km cycle.
- 1.4.1.2 Testing was closely controlled to match environmental conditions for each trial and subject.

1.4.2 Delimitations

- 1.4.2.1 All subjects were high-level triathletes with top 10 state rankings.
- 1.4.2.2 The use of each individual's own equipment eliminated factors relating to equipment set up.

1.4.2.3 All subjects were between 20 to 35 years of age.

1.5 Research Questions

- i) What is the effect of a 40km cycle on the level and time of activation of vastus lateralis, vastus medialis, rectus femoris, biceps femoris, gluteus maximus and gastrocnemius in running?
- ii) Does the level and time of muscle activation change within the first 2km of a 10km run?

1.6 Hypotheses

The hypotheses proposed for this study were:

- i) The level and time of activation of the sampled leg muscles during running after cycling when compared to running after an equally fatiguing run at three stages (0km, 1km, 2km) will be significantly different.
- ii) There will be a significant interaction between condition (cycle/run and run/run) and time.

1.7 Definition of Terms

Explanation of the terms used in running and cycling in this thesis are provided below:

1.7.1 Running Terms

- Foot strike: the time at which the foot makes contact with the ground.
- Toe-off: the moment when the toe ceases to have ground contact.
- Stance phase: the portion of the stride beginning at heel strike and ending at toe-off.
- Midstance: the part of the running stride midway between heel strike and toe-off.
- Flight phase: the time during the running stride between toe-off and foot strike, when the foot of the selected leg is not touching the ground. This consists of early swing, middle swing and late swing.
- Early swing: begins with toe-off and ends at contralateral heel strike.

- **Middle swing:** begins at contralateral heel strike and ends at contralateral toe-off in the running stride.
- **Late swing:** begins at contralateral toe-off and ends at foot strike.
- **Contralateral:** the opposite limb to the one being studied or measured.

1.7.2 Cycling Terms

- **Cadence:** pedalling rate.
- **Pedal revolution:** an entire 360° rotation of the pedal crank. This consists of a propulsive and recovery phase.
- **Propulsive phase:** $0 - 180^\circ$ of the pedal revolution beginning at the top. 0° is top dead centre.
- **Recovery phase:** $180 - 360^\circ$ of the pedal revolution. 180° is when the crank is bottom dead centre.
- **Toe clips:** pedals that have straps that can be tightened over the subjects foot.
- **Clipless pedals:** specially designed pedals, that match fittings on the bottom of cyclists shoes, that lock together.
- **Pedal upstroke:** another description for the recovery phase of pedalling.

1.7.3 Triathlon Terms

- **Transition:** for the purpose of this study is considered the time from the cessation of cycling up to the 2km point in the run

CHAPTER TWO

2.0 REVIEW OF LITERATURE

This review of literature will discuss the following factors pertinent to this study:

- 2.1 Muscle Activation in Running
- 2.2 Muscle Activation in Cycling
- 2.3 The Transition Stage in Triathlon
- 2.4 Influence of Fatigue on EMG
- 2.5 Processing of EMG Data

2.1 Muscle Activation in Running

Running involves a fast sequence of contraction and relaxation of the lower limb muscles due to the alternation of foot strike and flight phases (Mero & Komi, 1987). Muscle activation in running is related to the contraction of the muscles and the phase of the stride at which the activation is measured. The leg muscles to be examined in this study and their individual activities throughout a running stride are discussed below.

2.1.1 Rectus Femoris

The rectus femoris (RF) muscle crosses two joints and is used in both hip flexion and knee extension (Nilsson, Thorstensson, & Halbertsma, 1985). Studies by Montgomery, Pink and Perry (1994) and Nilsson et al. (1985) found the RF to have its greatest activation during foot strike, where it works together with vastus lateralis (VL) and vastus medialis (VM) to stabilise the knee throughout the stance phase.

The activity of RF remains high and continues to provide propulsion during toe-off due to its role in knee extension. The RF is still active during early swing however, the level of activation is somewhat lower than during foot strike (Montgomery et al., 1994; Ounpuu, 1994).

Montgomery et al. (1994) and Ounpuu (1994) proposed that the RF contracted eccentrically during early swing thereby controlling the extent to which the lower leg flexes at the knee. The RF displayed increased EMG activity while controlling the hip extension and preparing to initiate hip flexion during middle swing. It was also found to assist the vastii muscles in controlling knee flexion at this stage (Montgomery et al., 1994). A cessation in the activity of RF immediately prior to heel strike in running has been identified by both Mero & Komi (1987) and Montgomery et al. (1994).

Nilsson et al. (1985) in their comparison of muscle activation changes related to the speed of locomotion, showed the RF to be more active in hip flexion as the speed of locomotion increased. This highlights the findings of Mann, Moran and Dougherty (1986) and Montgomery et al. (1994) which concluded that the hip flexors are highly involved in providing forward propulsion in running.

2.1.2 Vastus Lateralis and Vastus Medialis

The VL and VM are both single joint muscles. During running their primary role is in stabilising the knee during foot strike (Montgomery et al., 1994; Nilsson et al., 1985). VL and VM exhibit only one period of high activity during running, that being during the stance phase. However, these muscles do demonstrate some activity when extending the knee in preparation for foot strike during the late swing phase. The activity of VL and VM decreases rapidly during the propulsive action at toe-off (Montgomery et al., 1994).

The VL and VM muscles have a significantly greater activation than the RF during foot strike (Montgomery et al., 1994). Nevertheless, when compared with each other there were no significant differences between the activation of the VL and VM. This highlights the role of these muscles to act as one unit in stabilising the knee (Montgomery et al., 1994).

2.1.3 Gastrocnemius

Activity of the gastrocnemius (GS) begins during late swing and continues throughout the stance phase (Mann et al., 1986; Ounpuu, 1994). Activity in this muscle terminates at the end of stance (Ounpuu, 1994). The GS displays the least

amount of activity directly after toe-off when the heel moves toward the hip (Mero & Komi, 1987).

Mann et al. (1986) and Reber, Perry and Pink (1993) suggested that the GS may play a role in stabilising the ankle at foot strike and during the stance phase, as the centre of gravity moves anterior to the foot. In their study of muscular control of the ankle in running, Reber et al. (1993) recorded peak activity of the GS at 70% and 80% of maximal muscle contraction during the midstance phase of running. This study also showed that increases in muscle activity coincided with increases in running velocity.

Mero and Komi (1987) in their study of sprint running found the GS to be highly active during the propulsive action at toe-off, suggesting the importance of this muscle during the push off in sprint running. Mann et al. (1986) had similar findings in their study of jogging, sprinting and running, where the GS was said to initiate plantar flexion of the foot, which therefore initiates toe-off. Interestingly these findings have been criticised by Reber et al. (1993) since their study showed peak activity of the GS occurring during midstance and then decreasing prior to toe-off, eliminating its role in propulsion.

2.1.4 Biceps Femoris (long head)

Montgomery et al. (1994) found the BF (long head) to have two periods of high activation. These periods of activity were displayed during midstance and late swing. During stance, BF initiates hip extension and in swing the BF works with gluteus maximus (GM) to extend the hip (Montgomery et al., 1994). Pinniger, Steele and Groeller (2000) suggested that during late swing the hamstring muscles are responsible for controlling the forward swing of the thigh, via an eccentric contraction, in readiness for foot strike.

2.1.5 Gluteus Maximus

Gluteus maximus (GM) is a single joint muscle that originates at the iliac crest and inserts into the femur. In running, this muscle acts primarily to provide extension at the hip (Nilsson et al., 1985).

In a study of muscle activity in relation to the speed of locomotion, Nilsson et al. (1985) found activity occurring in GM as the hip was extending prior to foot strike. Montgomery et al. (1994) found the GM to be highly active during foot strike and from their findings suggested a relationship between GM activation and the stabilisation of the hip during foot strike.

Two investigations (Mann et al., 1986; Nilsson et al., 1985) discovered that earlier activation of the GM coincided with an increase in running velocity. This led Mann et al. (1986) to believe that the GM is responsible for slowing rotation of the thigh prior to foot strike. Nilsson et al. (1985) however, suggested a shift in the initiation of activity of GM occurring late in hip flexion.

The activity of GM decreases quickly during the propulsive action during toe-off (Mero & Komi, 1987). Activation remains low until the late swing phase where the activity begins to increase once again (Montgomery et al., 1994).

2.2 Muscle Activation in Cycling

The GM and the hamstring group are used primarily to produce hip extension during pedalling (Clarys, Cabri, & Gregor, 1988). MacIntosh, Neptune and Horton (2000) described the GM as one of the primary power producing muscles in cycling. Gregor, Green and Garhammer (1981) found the highest activation of the GM to occur during 0-45° of the propulsive phase, whereas highest activation of the hamstring was measured during the last 45° (135° to 180°) of the propulsive phase. The VL and VM are active during the propulsive phase to 15° past the horizontal. The activation of the RF begins during the last 90° of the recovery phase when extending the hip. Its activity continues for the first 60° of the propulsive phase, indicating its role in the production of power.

Figures 1 and 2 present the findings of Ericson, Nisell, Arborelius and Ekholm, (1985) for muscle activity of selected lower limb muscles during cycling. Figure 1 shows a peak activity of RF of 12% of maximal voluntary contraction (MVC). In Figure 2, the VM and VL can be seen to have the highest activation with 54% and 50% respectively.

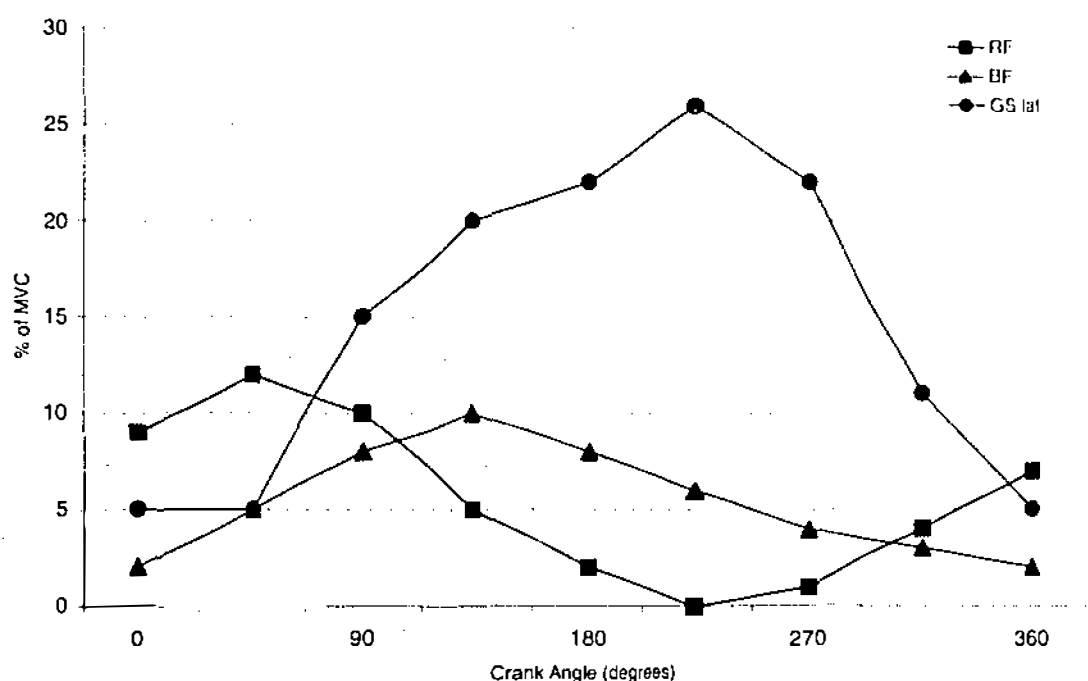


Figure 1. Muscle activation of the biceps femoris (BF), rectus femoris (RF) and gastrocnemius lateralis (GS lat) throughout a cycle revolution (Ericson et al., 1985)

Analysing muscle activation in cycling can be very difficult due to individual alterations in saddle height and saddle position (Clarys et al., 1988). Saddle height has been shown to change muscle activity of the RF and BF, however, this change was not shown to cause a significant difference (Clarys et al., 1988). Ericson et al. (1985) found that increases in saddle height caused the activity of the gastrocnemius medialis to increase but found no significant differences for the other muscles when saddle height was adjusted.

Increases in workload have been found to significantly increase the activation of all muscles (Ericson et al., 1985). Increases in cadence had a similar effect however, RF and BF were not affected by the cadence.

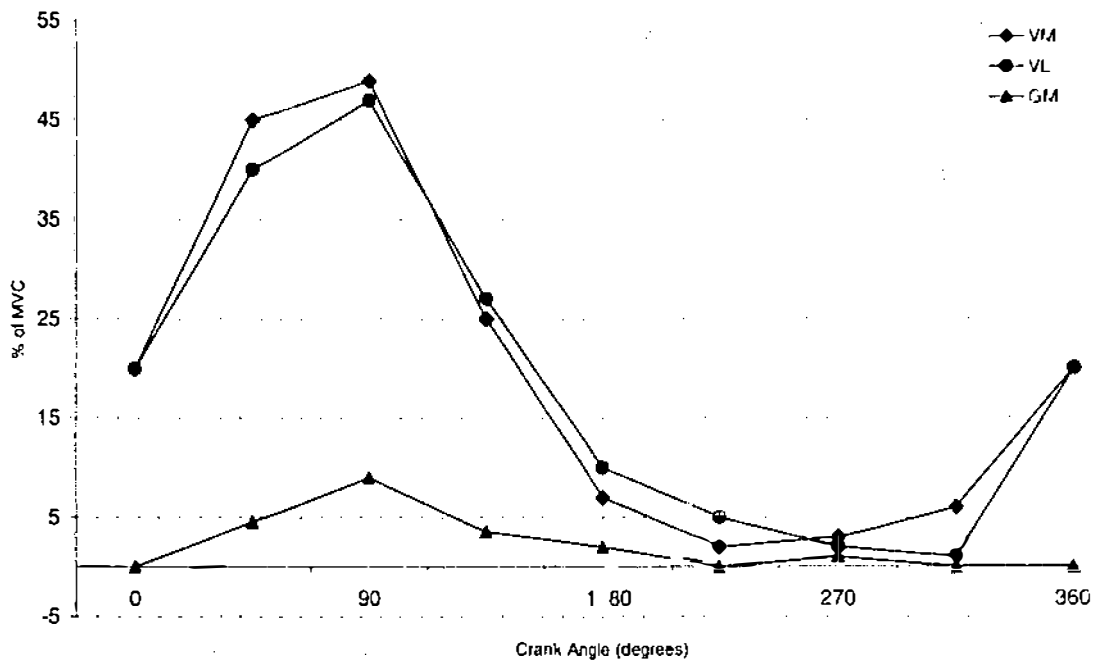


Figure 2. Muscle activation of the vastus medialis (VM), vastus lateralis (VL) and gluteus maximus (GM) muscles throughout a cycle revolution

Competitive cyclists and triathletes usually use toe-clips, or clipless pedals for competition and training. Ericson et al. (1985) found the toe-clips worn by competitive cyclists are often used to assist the pedal upstroke, allowing the cyclists to maximise the use of all lower limb muscles during the entire pedal revolution.

Interestingly, the use of toe-clips causes significant changes in the activation of several muscles. The RF and BF showed increases in activation whilst the VM and VL recorded decreased activation (Ericson et al., 1985). Clipless pedals function in much the same way as toe clips, allowing force to be applied to the pedal on the upstroke therefore, it can be assumed that these would result in the same activation changes.

2.3 The Transition Stage in Triathlon

A paucity of research has been undertaken to examine the biomechanical parameters of running after cycling. Of these studies, stride frequency and stride length are the most commonly considered variables, as they in combination determine running velocity. It should be noted that all these studies have been carried out in laboratory

settings on treadmills. Wank et al. (1998) found a difference in stride length, stride frequency and EMG muscle activation between overground and treadmill running.

Hue et al. (1998) found prior cycling to have no effect on the stride length and stride frequency of a subsequent run. However, Hausswirth et al. (1996) in their study comparing triathlon and marathon running did encounter changes in stride length following a preceding ride. A study examining kinematic variables in running after cycling (Quigley & Richards 1996), found no significant differences and declared specific transition training to be unnecessary.

To the investigator's best knowledge, only two studies have examined the effect of prior cycling on the muscle activity of the legs in running using EMG. Hausswirth, Brisswalter, Vallier, Smith and Lepers (2000) in their study of a prolonged run (2 hours & 15 minutes), a triathlon run (45 minute run after 40km cycle) and an isolated run (45 minutes) measured the EMG activity of the vastus lateralis. Findings from this study indicated that less muscle activation was evident during the triathlon run compared with the prolonged run. Witt (1993) found changes in muscle activation of the leg muscles after cycling. Specifically, activation began earlier in the stride and activation times were extended with the greatest variations in the vastus lateralis, tibialis anterior, and tensor fascia latae. However, the above study did not control for the effect of fatigue on EMG data.

2.4 Influence of Fatigue on EMG

When muscles are fatigued the surface EMG signal changes (Redfern, 1992; Winter, 1984). According to Redfern (1992) there are two commonly cited changes in the EMG signal. Firstly, the amplitude of the signal increases for a given task and secondly, the largest magnitude for a sustained isometric contraction occurs at a lower frequency than the same contraction measured in a non-fatigued state.

The increases in EMG due to fatigue have been attributed to several factors. They are the recruitment of new motor units, an increase in the synchronisation of motor unit firing, and a reduction in conduction velocity (Potvin & Bent, 1997; Winter, 1990).

Winter (1990) found that the maintenance of constant tension in the muscles, after the onset of fatigue, required the recruitment of new motor units to make up for the decreased firing rate of previously recruited units. It was also suggested by Winter (1990), that decreases in the peak twitch tensions of the motor units were compensated by increases in their contraction times. This finding was also highlighted by Pinniger et al. (2000), who found an increase in the duration of EMG activity in a muscle during fatiguing tasks of long duration.

Fatigue involves neuromuscular changes, the failure of force, or a decrease in tension (Tschoepe, Sherwood, & Wallace, 1994) and is dependent upon the task being undertaken (Enoka, 1995). For example, Hautier et al. (2000) found that fatigue of the agonist muscles increased the activation of the antagonist muscles.

In a study investigating neuromuscular fatigue in running after prolonged cycling, Lepers, Hausswirth, Maffiuletti, Brisswalter and Van Hoocke (2000), concluded that the impairment of muscle function was unrelated to the type of contraction. Garside and Doran (2000) and Millet and Vleck (2000), however, found a decrease in the efficiency of stretch-shortening movements like running when fatigued, due to less energy being stored in the muscle structure under fatigued conditions. Furthermore, fatigue has been found to alter leg stiffness, thereby altering stride length and stride frequency in the gait cycle (Garside & Doran, 2000).

The changes in EMG under fatigued conditions as outlined above show the importance of controlling for the effect of fatigue on EMG. If this change in EMG signal is not controlled for, the results could be misinterpreted to indicate changes to activation that may not exist. Therefore, this study will control for fatigue using a second pre-fatigued run condition with which to compare the findings.

2.5 Processing of EMG Data

Normalisation of EMG data is the process of establishing a standard reference from which comparisons among muscles, individuals and trials can be made (LeVeau & Andersson, 1992). Normalisation assists in overcoming changes in the electromyographic signal due to slight changes in electrode placement, temperature, and differences in subcutaneous fat between subjects (LeVeau & Andersson, 1992).

There are many methods to normalise EMG data, the most common of which is a maximal voluntary contraction (MVC). In recent literature, this method has been reviewed to determine its effectiveness in the normalisation of dynamic activities such as gait where the muscles are continually changing in length (LeVeau & Andersson, 1992).

A shortcoming of normalisation is that it retains large intersubject variability (LeVeau & Andersson, 1992; Yang & Winter, 1984). In a study examining walking Yang and Winter (1984) compared a 50% MVC with two different methods of ensemble averaging, they being normalisation to peak amplitude and mean ensemble EMG over each stride. The result revealed smaller intersubject variability when using the ensemble averaging methods of normalisation. Consequently, normalisation to the peak or mean value of the EMG signal has been used in several studies of dynamic activities (Clarys, Publie, & Zinzen, 1994; Lucia, Sanchez, Carvajal, & Chicharro, 1999; MacIntosh et al., 2000).

Other studies have used ensemble averaging, by using the mean linear envelope of the amplitudes for each subject, to quantify the relative intensity of the signal during a gait cycle. Ensemble averages have been used in several studies (Clarys et al., 1994; Hubley-Kozey & Earl, 2000; Nawoczenski & Ludewig, 1999) as a qualitative tool to represent graphically the data relating to significant findings. Bogey, Barnes and Perry (1992) suggested that ensemble averaging of data in gait studies provides the best indication of a subject's "typical EMG profile" due to the easy depiction of the actual amount of EMG activity. Yang and Winter (1984) also found this method to be indicative of an individual's actual EMG pattern and highlighted its ease of calculation without the need for technical software packages and the collection of extra data.

The root mean square (RMS) of the EMG signal is commonly used to determine the amplitude of the EMG signal due to its ability to reveal the full extent of the myoelectric activity (Lucia et al., 1999). MacIntosh et al. (2000) used the average RMS values for individual muscles, normalised to the individual maximum, to represent overall muscle activity and provide comparisons between subjects.

2.6 The Use of Accelerometers in Gait Studies

Conventional methods of gait analysis have used footswitches to determine stride variables such as heel strike and toe-off. According to Willemsen, Bloemhof and Boom (1990) "footswitches ...lack mechanical robustness, while their potential for further improvement ... is limited."

The use of accelerometers in the measurement of high accelerations like those experienced during heel strike has been deemed useful (Whittle, 1991). In a study which examined the use of an accelerometer to detect stance and swing phases in gait Willemsen et al. (1990) compared accelerometer readings against footswitch data to determine the exact points of interest. A clearly visible peak represented heel strike, whilst toe-off was characterised by another smaller peak in acceleration.

Kim, Voloshin, Johnson and Simkin (1993) found the accelerations due to heel strike to be reduced when measured further away from the leg, indicating the effect of placement and location on the acceleration pattern. This is further highlighted by Willemsen et al. (1990) who measured higher values of acceleration at heel strike when accelerometers were placed closer to the ankle, indicating the damping effect.

CHAPTER THREE

3.0 MATERIALS AND METHODS

3.1 Target Population

Ten elite triathletes within the age range of 20 to 35 years of age were invited to participate in this study. Selection was based on their performance at the Western Australian State Olympic distance championships. The maximum time for these subjects was 2 hours and 5 minutes for an Olympic distance triathlon.

The researcher carried out all data collection within a four-month period. All subjects completed testing on the same pre-designed cycle and run course. Informed consent, in accordance with the Edith Cowan University Ethics Department was obtained from subjects prior to testing (Appendix A). Any subjects carrying injuries to the lower limbs were excluded from participating in the study.

3.2 Data Collection

Data collection consisted of two separate tests undertaken in random order, these being; a 40km ride followed by a 2km run (cycle/run) and a 10km run followed by a 2km run (run/run). These tests were carried out at least a week apart. The run/run condition was designed to control for fatigue, as fatigue is known to alter the EMG signal (Oberg, 1995; Redfern, 1992; Winter, 1990). Since the level and influence of fatigue on EMG during different types of prolonged exercise has not yet been recognised, the length of the run in the run/run condition was designed using energy expenditure (Hauswirth et al., 2000) as described below.

To calculate a run distance equivalent to 40km of cycling for the initial run in the run/run condition, it was considered that all subjects weighed between 70kg and 80kg and therefore expend approximately 780 kcal during a 1 hour cycle. Running at a 4km race pace expends approximately 19.1 kcal per minute, this equates to 40 minutes of running equalling 1 hour of race pace cycling (McArdle, Katch, & Katch, 1996). Consequently, the subjects in this study cycle 40km in approximately 1 hour and run 10km in approximately 40 minutes depending on the course therefore, a 10km run was deemed to be equivalent to a 40km cycle.

All testing was carried out on a set course, in the Perth suburb of Shelly, with the subjects using their own bikes. Subjects were instructed to perform a suitable warm up prior to each testing session. The course used for cycling and running was controlled for all subjects. The environmental conditions such as time of day and weather conditions were monitored for similarities, as were the cycle and run times of each test to reduce variability between subjects and trials. It is known that cadence alters the EMG activity of the lower limbs in cycling (Gregor, Broker, & Ryan, 1991). Therefore, to reduce the effects of different cycling cadence between the subjects and its possible effect on muscle activation in running, subjects were instructed to reduce gears and increase cadence to maintain speed 1km prior to transition from cycle to run.

Data was collected via a portable Mega Electronics ME3000 eight channel EMG data logger (Mega Electronics, Finland) with a mass of 590g. EMG and acceleration data were measured at 1000Hz (8-500 Hz bandwidth) over the duration of the 2km run for both the run/run and cycle/run conditions (Figure 3). EMG data were collected from the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), gastrocnemius (GS) and gluteus maximus (GM). These muscles were chosen due to their important contributions to both running and cycling.

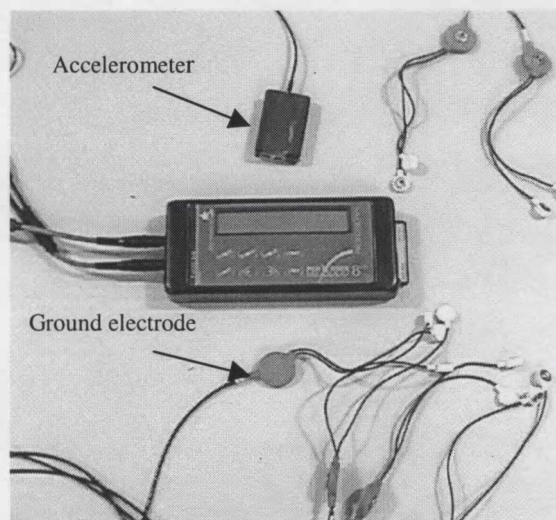


Figure 3. Mega Electronics ME3000 portable EMG data logger with electrodes and accelerometer attached.

During the transition period prior to commencement of each 2km run, the datalogger was set to record continuously. The run was timed and typical triathlon speed was maintained by checking the subject's times to reach 1km and 2km.

To minimise variability between subjects and testing sessions, EMG electrode placement followed the guidelines of Hermens, Freriks, Disselhorst-Klug and Rau, (2000). Electrode positions were measured and marked on the skin with a waterproof marker before subject preparation, which involved shaving and abrading the skin

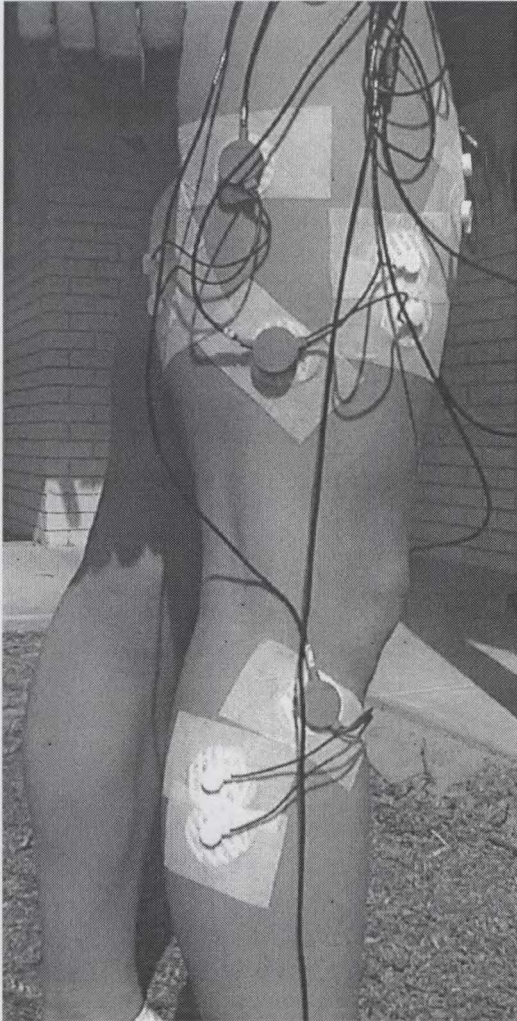


Figure 4. Electrode attachment and configuration

followed by cleaning the area with an alcohol swab. Once the skin was dry, bipolar 3M Red Dot Ag/AgCl surface electrodes with a diameter of 30mm were adhered to the skin (20mm inter-electrode distance) parallel to the muscle fibres over the respective muscle sites on the subject's right side. Electrodes were secured to the skin with Beiersdorf Fixomull™ stretch tape to ensure that they remained attached to the skin during testing.

Pre-amplified cables, one channel per muscle, were connected to each electrode pair and ground electrode as shown in Figure 4. These cables were attached to the datalogger, which was worn in a pouch around the subject's waist throughout each test. Pilot testing confirmed that the equipment did not disturb normal running, particularly with the use of lycra tights worn over the electrodes to minimise movement of the cables and enhance subject comfort.

Typically in gait research a footswitch is used to determine heel strike and toe-off via a digital pulse. Due to the extent of testing and the distances to be run in this study, a more robust method was required for the determination of these variables. Initially, video was considered as a method of determining heel strike and toe-off however,

this would have required additional resources such as camera operators and the design of a synchronisation method between video and EMG. To eliminate these problems an accelerometer was attached with tape above the subject's ankle with its axes aligned to the intended direction of acceleration to record heel strike and toe-off for each stride. The method in which heel strike and toe-off were determined from the accelerometer is described below.

3.3 Determination of Heel Strike and Toe-Off Using an Accelerometer

Prior to testing, the method of using the accelerometer to determine heel strike and toe-off was evaluated. This was done via synchronised data collection at 1000Hz from a Kistler 9287B piezoelectric force plate (Kistler Instrument Corp, Switzerland) and a bi-directional $\pm 1.25G_z$ accelerometer (Mega Electronics, Finland).

True heel strike and toe-off was determined via inspection of the F_z component of the ground reaction force measured from the force plate (F_z is the component of force in the up and down direction). The beginning and end of the F_z signal was regarded as a highly precise measurement with which the reliability and accuracy of the accelerometer in determining heel strike and toe off could be compared (Figure 5).

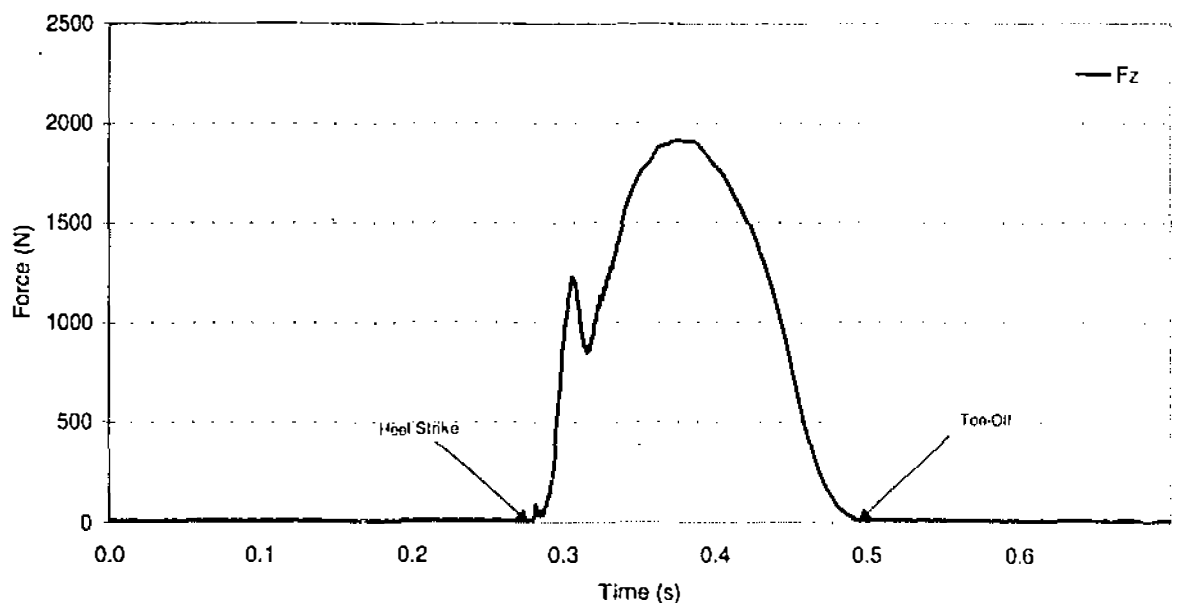


Figure 5. F_z component of ground reaction force from the Kistler force plate showing true heel strike and toe-off.

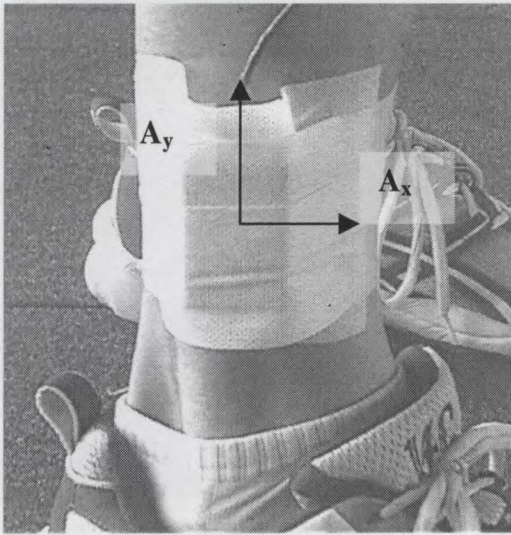


Figure 6. Accelerometer attachment on the ankle with the acceleration in the X (A_x) and Y (A_y) directions shown.

Six male subjects completed five trials with a 15-meter approach. The accelerometer was attached to each subject's right side using Beiersdorf Fixomull™ tape, wrapped firmly over the surface of the accelerometer (Figure 6). To ensure little or no discomfort to the subject, familiarisation trials were carried out prior to testing. Subjects were instructed to contact the force plate with their right foot whilst maintaining speed and stride length, as a change in stride would also change the pattern of the accelerometer. Two attachment locations were examined to determine

the placement site with the most easily depicted and repeatable pattern for heel strike and toe-off, they being the hip and the ankle.

At the completion of these trials, data was downloaded by computer software (Mega Electronics, Finland) designed for use with the datalogger. After downloading, raw data were converted to text file format and imported into Microsoft Excel. F_z and acceleration in X-direction (A_x) and acceleration in Y-Direction (A_y) were then graphed and compared, within and between subjects, to determine whether a repeatable pattern existed to accurately determine heel strike and toe-off from the accelerometer patterns.

Initial attachment of the accelerometer was on the hip, inferior to the iliac crest, close to the centre of gravity with the accelerometers Y-axis aligned with the long axis of the femur. However, due to the difficulty of attaching the accelerometer securely, a clearly distinguishable heel strike and toe-off pattern could not be found.

An example of the A_y pattern with the accelerometer attached at the hip over several gait cycles is shown over time in Figure 7. Due to the large variation between

subjects and lack of specific pattern, the points of heel strike and toe-off were unable to be identified accurately from this pattern. A more detailed comparison of A_y data from the hip and the ground reaction force (F_z) is shown in Figure 8. The precise points of heel strike and toe-off were unable to be identified from the accelerometer output due to its positioning.

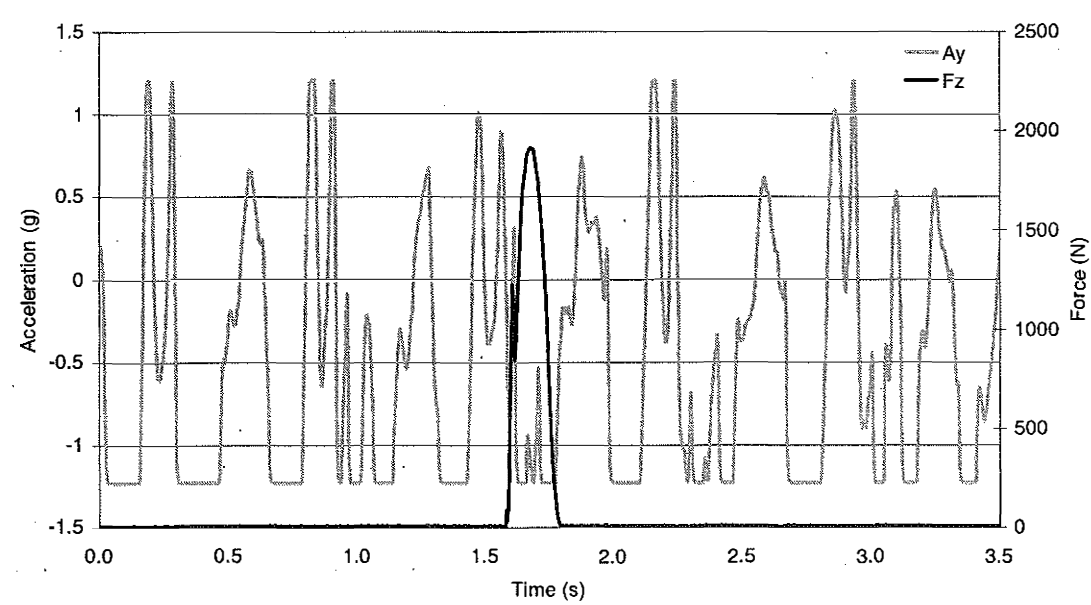


Figure 7. Pattern of acceleration in the Y-direction (A_y) during running with accelerometer attached to the hip.

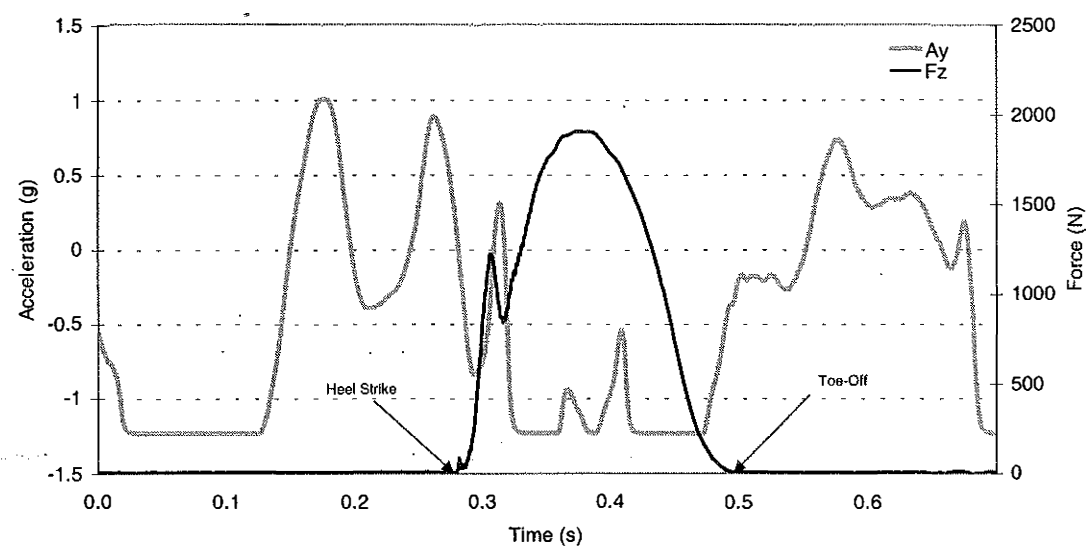


Figure 8. Detailed F_z component of ground reaction force and acceleration in the Y-direction (A_y) data with accelerometer attached to the hip.

Alternatively, the accelerometer was attached to the right ankle above the lateral malleolus with A_y aligned to the long axis of the tibia. The pattern from this attachment was more clearly distinguishable with little variation between subjects (Figure 9). Furthermore, the attachment of the accelerometer was more secure due to the wrapping of tape around the entire limb to reduce any extra movement of the accelerometer.

Figure 9 shows a highly repeatable pattern of heel strike and toe-off when the accelerometer was attached to the ankle. This pattern was similar across all subjects with a major spike representing heel strike and a smaller spike representing toe-off, which is in agreement with the findings of Willemsen et al. (1990). This can be seen in more detail in Figure 10 where the exact points of heel strike and toe-off are shown in relation to the ground reaction force (F_z).

The time lag refers to the elapsed time between actual heel strike and toe-off as measured by the force plate and these same variables estimated from the accelerometer respectively (Figure 10). Preliminary testing revealed the time lag was related to the distance that the accelerometer was positioned away from the impact point, in this case the foot. A "rigid" structure was used to verify time lag as it was expected that once force was registered acceleration would be registered very shortly after.

To verify the existence of a time lag the accelerometer was attached at pre-determined distances away from the base of a 1m long timber post. The post was struck against the force plate with three trials at each distance, 10cm, 30cm, and 40cm from the impact point. The points of force plate and post contact, and contact as registered by the accelerometer clearly verify the existence of a lag time between force plate and accelerometer data (Figure 11). In this figure the time lag is shown not as the first instance of change in the pattern, but at the point of highest positive acceleration. This is in keeping with the heel strike and toe-off measurements that were found to be spikes in the accelerometer pattern (Willemsen et al., 1990).

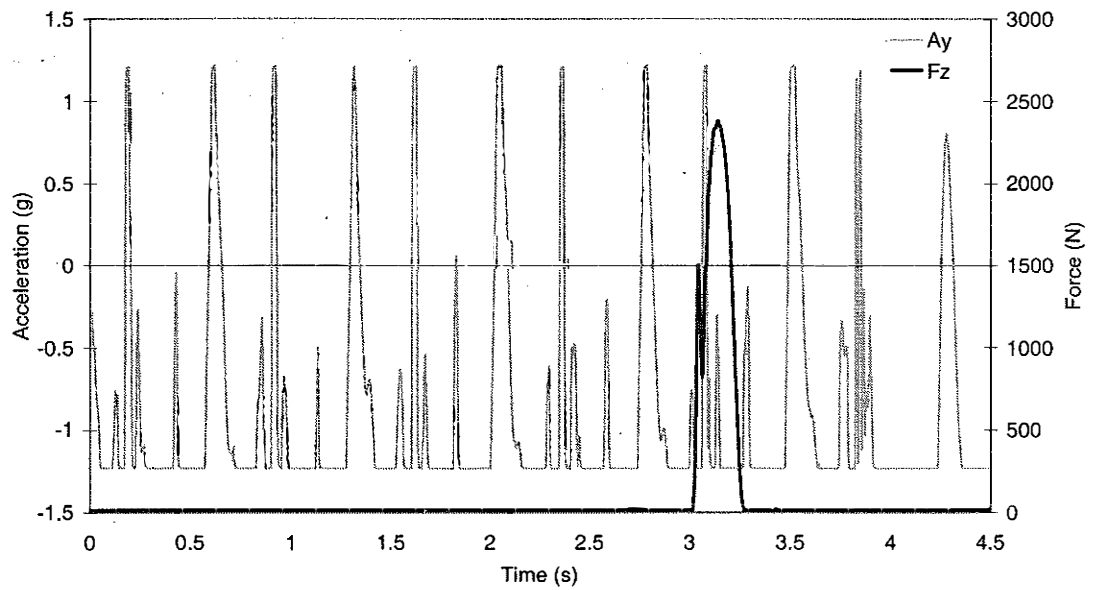


Figure 9. Acceleration pattern in the Y-direction (A_y) during running with accelerometer attached to the ankle.

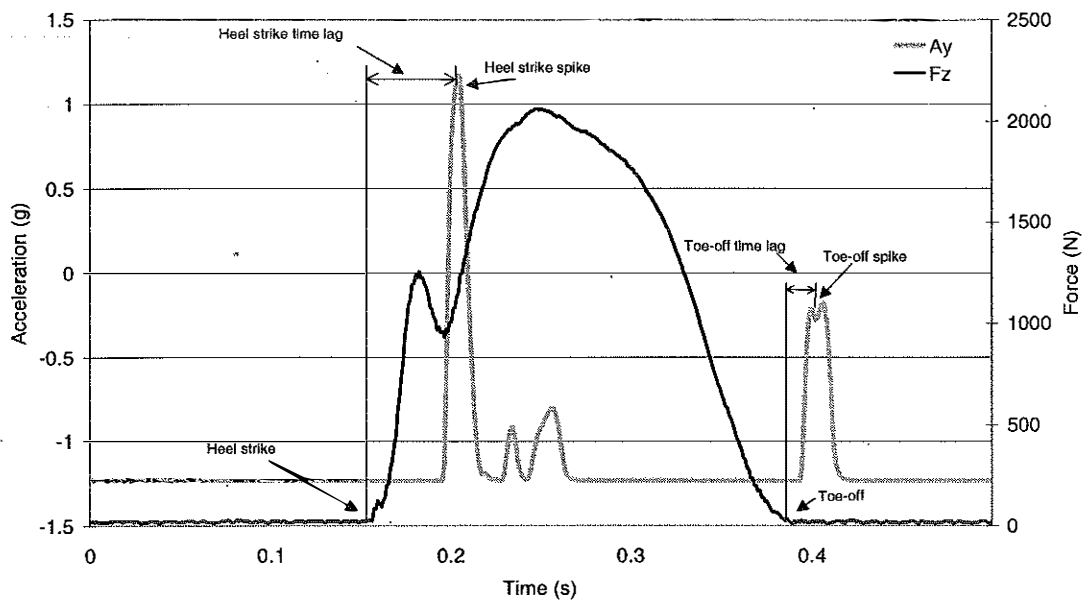


Figure 10. Detailed F_z component of ground reaction force and acceleration in the Y-direction (A_y) data with accelerometer attached to the ankle.

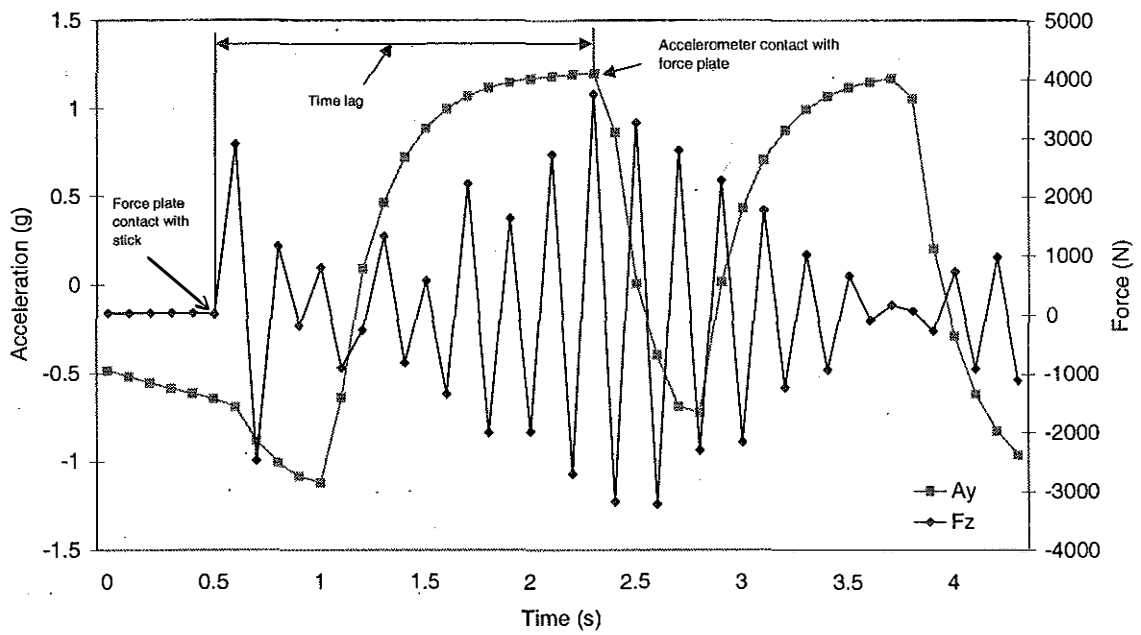


Figure 11. The presence of a time lag between the F_z component of ground reaction force and Y-direction acceleration (A_y) with accelerometer attached 10cm from the point of impact on the timber post.

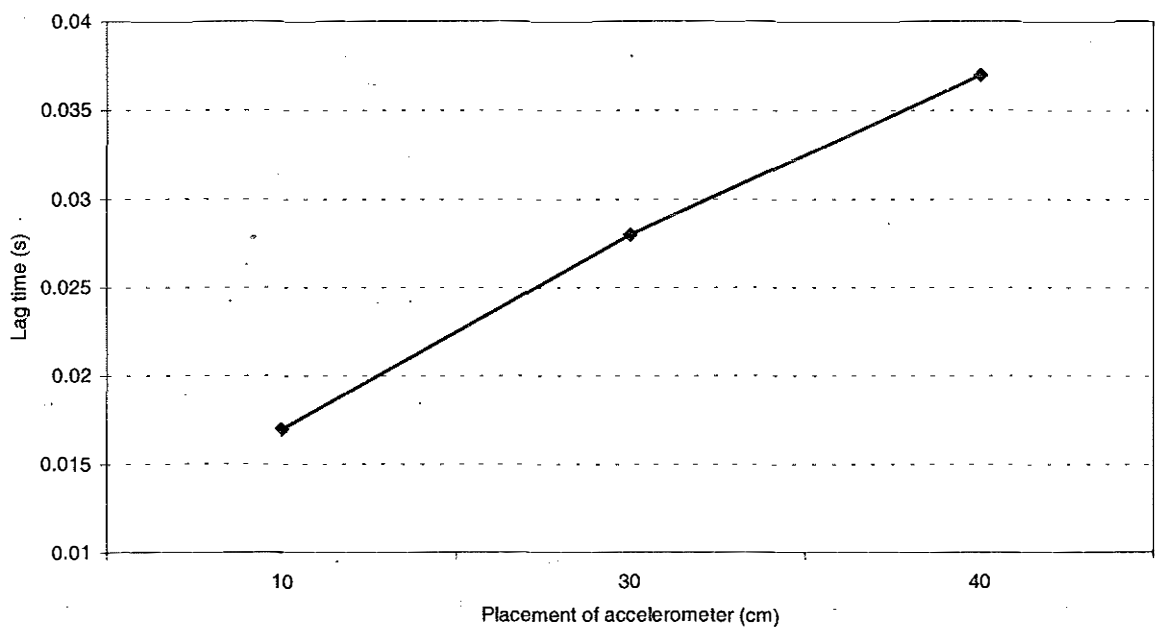


Figure 12. Average lag times for the accelerometer when placed at set distances from the base of the post used in testing.

Figure 12 shows the lag times at each of the three distances of attachment from the base of the post. This clearly indicates that the further away the accelerometer is from the point of impact, the greater the time lag. Again, this highlights the definite existence of a time lag between force and accelerometer measurement.

To calculate the lag time for trials with the accelerometer attached to the ankle of the subjects, the raw data were graphed and the heel strike and toe-off peaks determined. Time lags from the ankle trials were compared for all subjects to ensure consistency. Table 1 shows the lag times for heel strike and toe-off in addition to the summary statistics for each trial.

Table 1
Lag times in seconds between the F_z component of ground reaction force and A_y data for heel strike and toe-off.

Subject	Trial Number									
	1		2		3		4		5	
	HS	TO	HS	TO	HS	TO	HS	TO	HS	TO
1	0.068	0.011	0.071	0.012	0.068	0.011	0.070	0.015	0.064	0.017
2	0.072		0.068		0.068		0.071		0.068	
3	0.058		0.059		0.059		0.066		0.067	
4	0.065	0.020	0.061	0.017	0.061	0.013	0.059	0.014	0.060	0.015
5	0.063	0.013	0.061	0.015	0.061	0.014	0.064	0.016	0.061	0.015
6	0.066		0.073		0.064		0.063		0.063	
Mean	0.065	0.015	0.066	0.015	0.064	0.013	0.066	0.015	0.064	0.016
SD	0.005	0.005	0.006	0.003	0.004	0.002	0.005	0.001	0.003	0.001

Note: HS = Heel Strike, TO = Toe-off, SD = Standard Deviation

The average lag for heel strike taken from the above data, was $65 \pm 5\text{ms}$ and $15 \pm 3\text{ms}$ respectively for heel strike and toe-off (Table 2). These lag times were used to ensure accuracy in the processing of subject data.

It can be seen that there are large variations between the lag times taken during run testing (Table 1) and those measured with the accelerometer attached directly to the

post (Figure 12). This is in part due to the cushioning properties of running shoes worn by the subjects, as when a single barefoot trial was undertaken in pilot testing to determine the effect of footwear on the accelerometer, the lag time was 36ms. Further, the structural complexity of the ankle joint makes it more compliant to the impact than the post, thus increasing the lag time.

As can be seen in Table 1, subjects 2, 3 and 6 did not clearly show toe-off from the accelerometer signal. In the case of this occurring during field testing, an alternative method of identifying toe-off was investigated. Using the acceleration pattern in the X-direction, a model was identified to determine toe-off consistently for every subject, making this a repeatable measure. Figure 13 shows this pattern.

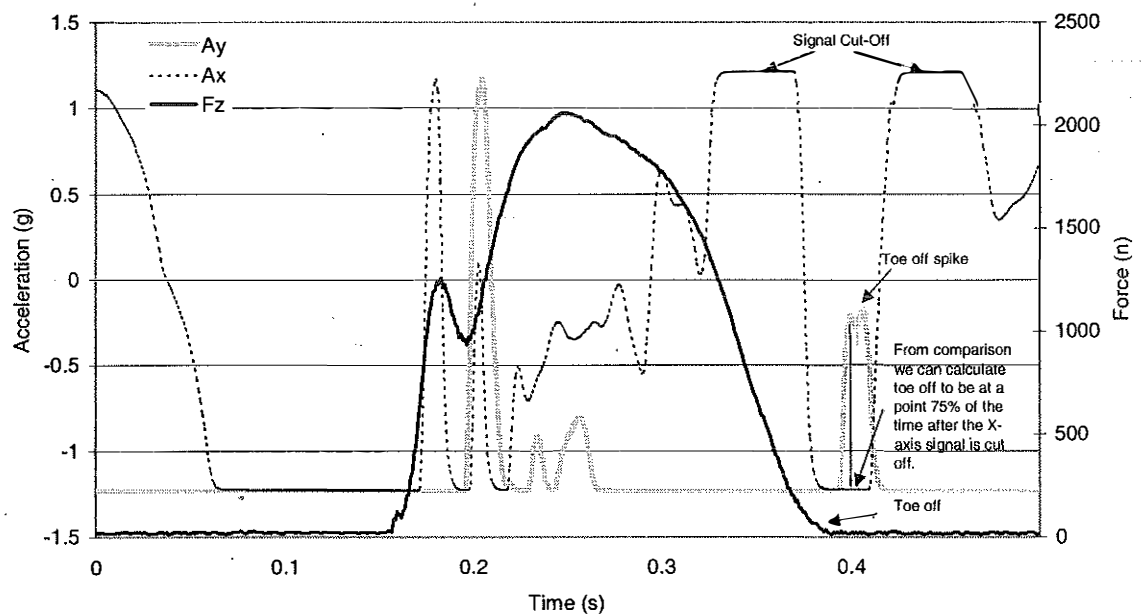


Figure 13. Comparison of acceleration in the X-direction (A_x) and in the Y-direction (A_y) with the F_z component of the ground reaction force for the determination of an alternative method of estimating toe-off.

Figure 13 indicates the point in the A_x pattern where toe-off occurs. The accelerometer used in this study was restricted to a $\pm 1.25G_z$ range of measurement causing some of the accelerometer signal to be cut off as can be seen occurring between 0.3 and 0.4 seconds at $1.25G_z$ and $-1.25G_z$. Therefore, it was necessary to determine the point at which toe-off occurred from the A_x pattern. This was

established by matching up both the time lag and the point of the toe-off spike from A_y , which was calculated to be at a point 75% of the time after signal cut off.

3.4 Data Analysis

Due to the large data files generated during testing (in the order of 70-80MB), data were sectioned into three stages of six strides using the accelerometer measurements to determine heel strike and toe-off. These stages were the first six strides once race pace had been reached (time 1), the last six strides of the first kilometre (time 2) and the last six strides of the second kilometre before slowing occurred (time 3). The data was sectioned via a customised program written in LabVIEW 5.1 (National Instruments, USA) and divided into three text files representing times one, two and three respectively. Data were analysed by another customised LabVIEW software program to calculate two variables, they being; level of activation and time of activation. The stages of processing involved in the calculation of these variables are outlined in Figure 14.

3.4.1 Level of Activation

The raw EMG signals from each of the six channels of muscle activity were root mean square (RMS) processed using a 25 millisecond (ms) moving window (Nawoczinski & Ludewig, 1999). Swing and stance phases of the running stride were identified from the accelerometer data and each stride was time normalised from 0 to 1000 via a cubic spline. RMS values were then normalised to the average maximum value observed for each subject in each condition to allow for comparison of EMG data between subjects (MacIntosh et al., 2000). The average RMS value for each section was computed for all subjects and expressed as a percentage of the individual's maximum RMS value (MacIntosh et al., 2000).

To provide a qualitative analysis of EMG activation, the raw EMG data were full wave rectified and low pass filtered with a Butterworth digital filter ($f_{\text{cutoff}} = 5\text{Hz}$) to form linear envelopes from which ensemble averages were calculated for each muscle over each of the three sections of six strides. The measure of variation between strides for each subject was determined from the time normalised data via the coefficient of multiple correlation (Kadaba et al., 1989).

3.4.2 Time of Activation

The time of activation was determined from the ensemble average by calculating a threshold (10% of maximum amplitude) to determine the onset and offset of muscle activation (Pinniger et al., 2000). Using Microsoft Excel this threshold was applied to the ensemble average to determine the time of activation. This value was calculated for each of the six muscles (RF, BF, VL, VM, GS and GM) and represented as a percentage of the stride phase (flight and stance).

3.5 Statistical Analysis

The independent variables were three levels of time (at the 0km, 1km, 2km stages) and two levels of condition (cycle/run & run/run) for both stance and flight phases of the running stride (Table 2). The dependent variables were level and time of activation for each of the six muscles measured. These variables were compared using a factorial ANOVA with two within-subject variables, as each subject participated in each condition. Comparisons between the conditions were made by examining condition across the independent variable, time. Further comparisons were made between the two separate conditions at each level of time to find significant interactions. Individual comparisons were made for significant F-values using dependent t-tests to determine where the differences occurred. All statistical testing was carried out using the Statistical Package for Social Sciences (SPSS 10.0) software. Differences were considered statistically significant at P values of <0.05.

Table 2
Diagrammatic view of statistical design for stance and flight related variables.

		TIME		
		0km	1km	2km
CONDITION	Cycle / Run			
	Run / Run			

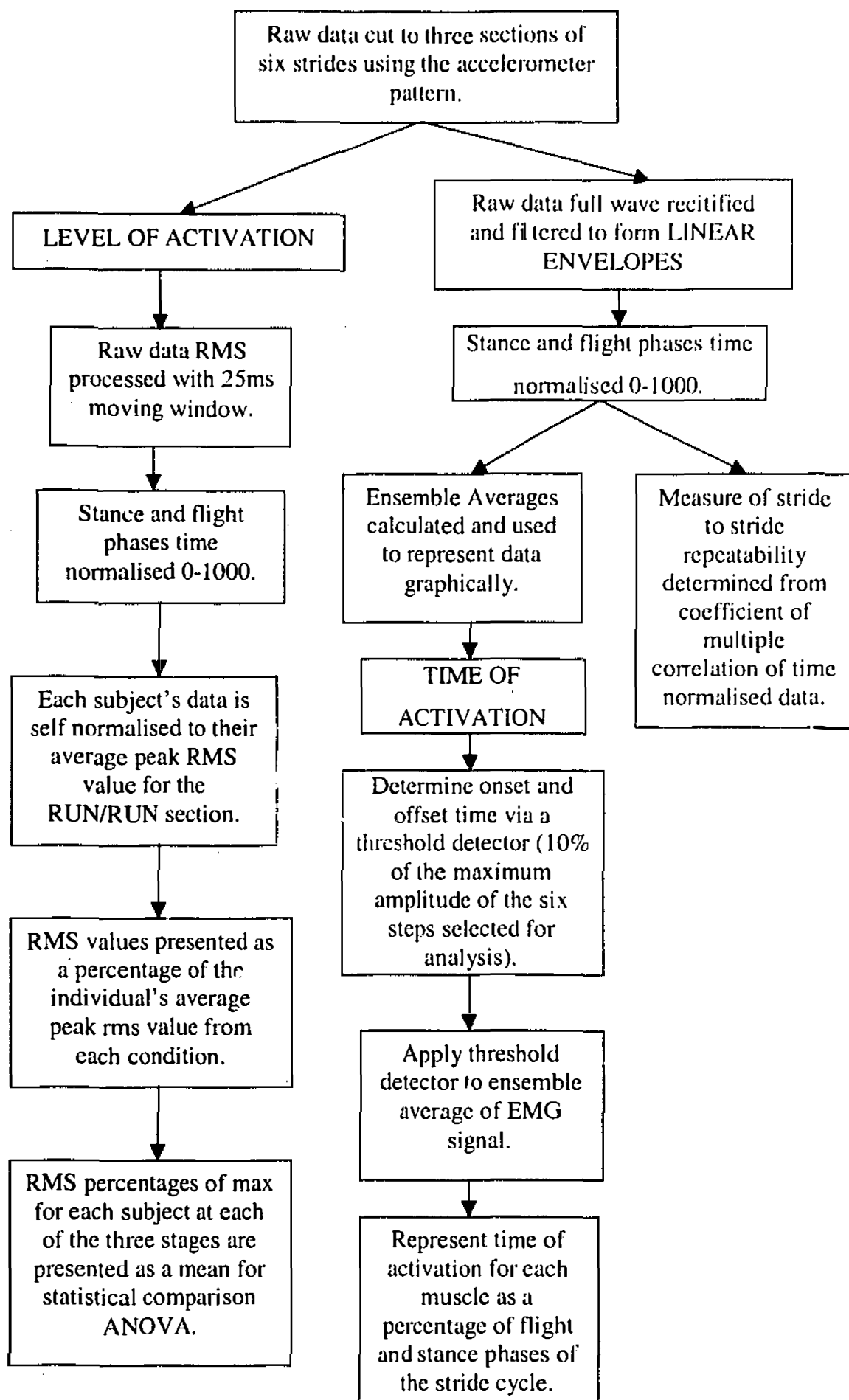


Figure 14. Diagrammatic view of processing methods

CHAPTER FOUR

4.0 RESULTS

This section outlines the results obtained from the analysis of the VL, VM, BF, RF, GS, and GM during flight and stance for measurements taken over 0km, 1km and 2km. The results are presented in three sections as follows; stride to stride repeatability of lower limb EMG, time of activation and level of activation. A $p < 0.05$ level of significance was selected to establish differences between variables. No significant interactions between condition and time were found however, significant differences were found between the run/run and cycle/run conditions for several muscles and significances were also found to occur over times one, two and three. These results will be elaborated upon later in this section.

Data during testing was collected via a data logger. Raw EMG profiles of all six muscles and accelerometer data for the Y-direction (A_y) are shown in Figure 15. Examination of the raw EMG profiles was necessary to ensure data was artefact free. As described previously, the raw accelerometer data (A_y) was used to determine the points of heel strike and toe-off for data separation into flight and stance phases.

4.1 Stride to Stride Repeatability of Lower Limb EMG

The coefficient of multiple correlation (CMC) was used to measure the similarity of the processed EMG waveforms between the six strides from each of the three measured times for all subjects in this study. Similarities of waveforms are depicted by CMC values that are closer to 1.0 (Kadaba et al., 1989). The average CMC values for each muscle in the cycle/run and run/run conditions are shown in Tables 3 and 4. These values indicate a highly repeatable EMG pattern for all subjects across all muscles for both the flight and stance phases of the running stride. Therefore, averaging over several strides to obtain quantified EMG values such as the linear envelope and RMS is a valid approach for this data.

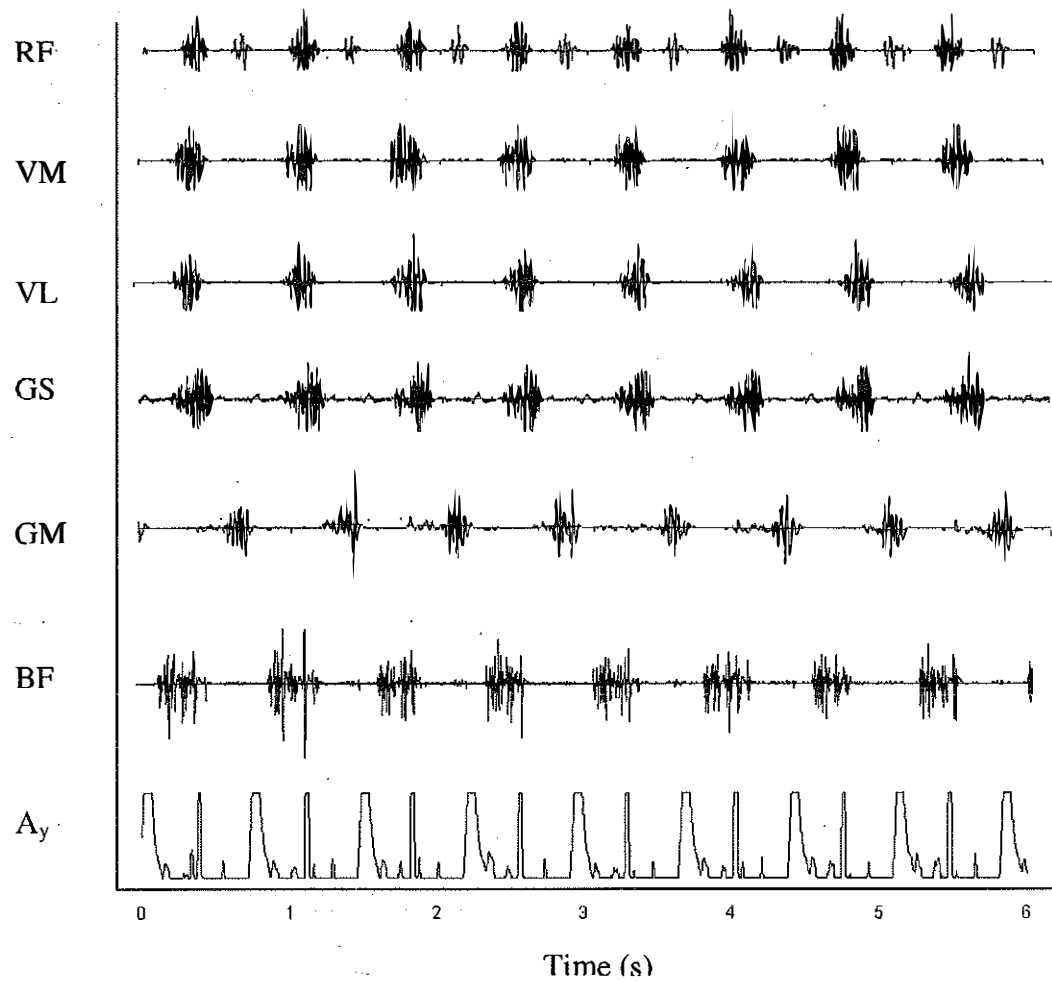


Figure 15. Raw EMG profiles and A_y signal from a representative subject for six muscles, RF (rectus femoris), VM (vastus medialis), VL (vastus lateralis), GS (gastrocnemius), GM (gluteus maximus) and BF (biceps femoris). The narrow spike on the A_y pattern represents heel strike.

Table 3

Mean Coefficient of Multiple Correlation results for the cycle/run condition.

	TIME 1		TIME 2		TIME 3	
	STANCE	FLIGHT	STANCE	FLIGHT	STANCE	FLIGHT
RF	0.964	0.961	0.979	0.956	0.944	0.917
VM	0.949	0.972	0.968	0.968	0.975	0.986
BF	0.863	0.969	0.883	0.974	0.862	0.966
VL	0.975	0.982	0.962	0.986	0.974	0.982
GS	0.946	0.967	0.957	0.937	0.883	0.898
GM	0.932	0.944	0.904	0.933	0.926	0.880

Note: RF = Rectus Femoris, VM = Vastus Medialis, BF = Biceps Femoris, VL = Vastus Lateralis,

GS = Gastrocnemius, GM = Gluteus Maximus

Table 4

Mean Coefficient of Multiple Correlation results for the run/run condition.

	TIME 1		TIME 2		TIME 3	
	STANCE	FLIGHT	STANCE	FLIGHT	STANCE	FLIGHT
RF	0.952	0.967	0.959	0.961	0.937	0.955
VM	0.959	0.984	0.961	0.985	0.967	0.979
BF	0.870	0.980	0.821	0.976	0.787	0.978
VL	0.972	0.986	0.979	0.988	0.982	0.990
GS	0.895	0.879	0.934	0.868	0.877	0.872
GM	0.910	0.954	0.900	0.964	0.885	0.957

Note: RF = Rectus Femoris, VM = Vastus Medialis, BF = Biceps Femoris, VL = Vastus Lateralis,

GS = Gastrocnemius, GM = Gluteus Maximus

4.2 Time of Activation

The time of activation was determined from the ensemble average by calculating a threshold (10% of maximum amplitude) to determine the onset and offset of muscle activation.

The mean \pm SD values for each muscle during the time of activation for the flight phase are shown in Table 5. RF showed the highest time of activation and the greatest variability among the sampled muscles during flight with $71.5 \pm 31.5\%$ for cycle/run and $64.7 \pm 10.8\%$ for run/run. Statistically significant differences ($p < 0.05$) were found for the main effect of condition for both the RF ($F_{1,9} = 6.222$) and VM ($F_{1,9} = 10.064$).

Table 6 shows the mean \pm SD values for the time of activation during stance of each muscle. The highest variance among subjects was in BF which was significant ($F_{1,9} = 6.236$) with $78.3 \pm 13.1\%$ and $90.9 \pm 11.7\%$ for cycle/run and run/run respectively. Furthermore, VL displayed a significant difference across time ($F_{2,18} = 4.708$). Post-hoc statistical analysis via dependent t-tests found the significant difference to have occurred between time two (1km) and time three (2km).

Figure 16 graphically depicts the activation times for each of the measured muscles in both conditions. Average stance and flight phases of all subjects were taken from the accelerometer data for heel strike and toe off to give a clear indication of the time of activation for each phase of the stride. A threshold of 10% of maximum activation was applied to the ensemble average to determine when the muscle was firing. Although not shown in this figure, low levels of activation do occur for the different muscles throughout the entire stride. This is highlighted in Figure 18 where the activation of the VL for an entire stride cycle is represented graphically.

Table 5

Means and standard deviations for time of activation during flight (%).

Variable	Condition	Time			Marginal Mean	F-Condition
		1	2	3		
Biceps Femoris	Cycle/Run	58.1 (13.5)	56.2 (11.1)	58.4 (12.5)	57.6 (12.4)	0.123 F-Interaction 0.562
	Run/Run	59.5 (15.7)	63.6 (17.3)	63.5 (14.7)	62.2 (15.9)	
	Marginal Mean	58.8 (14.6)	59.9 (14.2)	60.9 (13.6)		
	F- Time		0.556			
Gluteus Maximus	Cycle/Run	40.2 (6.8)	37.3 (14.0)	35.0 (7.8)	37.5 (9.7)	0.152 F-Interaction 0.559
	Run/Run	47.7 (21.2)	48.8 (22.7)	41.3 (19.1)	45.9 (21.0)	
	Marginal Mean	43.9 (14.0)	43.1 (18.4)	38.2 (13.5)		
	F- Time		2.353			
Gastrocnemius	Cycle/Run	23.6 (3.0)	22.8 (4.1)	22.3 (2.5)	22.9 (3.2)	0.075 F-Interaction 0.142
	Run/Run	22.8 (7.9)	23.9 (7.6)	24.9 (6.0)	23.9 (7.2)	
	Marginal Mean	23.2 (5.5)	23.4 (5.9)	23.6 (4.3)		
	F- Time		0.788			
Rectus Femoris	Cycle/Run	73.4 (11.8)	67.3 (13.3)	73.7 (15.5)	71.5 (31.5)	6.222* F-Interaction 3.059
	Run/Run	66.6 (7.4)	66.9 (5.1)	60.6 (19.9)	64.7 (10.8)	
	Marginal Mean	70.0 (9.6)	67.1 (9.2)	67.2 (17.7)		
	F- Time		0.462			
Vastus Lateralis	Cycle/Run	33.1 (8.1)	29.1 (2.9)	32.9 (16.9)	31.7 (9.3)	0.335 F-Interaction 0.178
	Run/Run	29.9 (6.7)	29.1 (5.5)	32.3 (11.2)	30.4 (7.8)	
	Marginal Mean	31.5 (7.4)	29.1 (4.2)	32.6 (14.0)		
	F- Time		0.724			
Vastus Medialis	Cycle/Run	28.3 (2.4)	26.3 (1.8)	26.8 (3.4)	27.1 (2.5)	10.064* F-Interaction 1.230
	Run/Run	26.3 (2.1)	25.8 (2.1)	25.4 (1.8)	25.8 (2.0)	
	Marginal Mean	27.3 (2.3)	26.1 (2.0)	26.1 (2.6)		
	F- Time		3.235			

* Denotes significant difference ($p < 0.05$) between cycle/run and run/run.

Table 6
Means and standard deviations for time of activation during stance (%).

Variable	Condition	Time			Marginal Mean	F-Condition
		1	2	3		
Biceps Femoris	Cycle/Run	73.4 (12.6)	77.4 (13.4)	84.0 (13.3)	78.3 (13.1)	6.236*
	Run/Run	91.9 (9.2)	87.7 (16.6)	92.9 (9.3)	90.9 (11.7)	
	Marginal Mean	82.6 (10.9)	82.6 (15.0)	88.5 (11.3)		F-Interaction 0.578
	F- Time		2.279			
Gluteus Maximus	Cycle/Run	57.8 (8.7)	58.4 (9.7)	52.5 (7.5)	56.2 (8.6)	0.154
	Run/Run	64.0 (16.0)	57.0 (6.1)	61.4 (8.6)	60.8 (10.2)	
	Marginal Mean	60.9 (12.4)	57.7 (7.9)	57.0 (8.1)		F-Interaction 2.433
	F- Time		0.088			
Gastrocnemius	Cycle/Run	81.9 (3.9)	83.9 (6.7)	82.5 (4.4)	82.8 (5.0)	0.744
	Run/Run	83.6 (7.3)	83.1 (8.4)	85.7 (6.6)	84.2 (7.4)	
	Marginal Mean	82.8 (5.6)	83.5 (7.6)	84.1 (5.5)		F-Interaction 0.001
	F- Time		0.857			
Rectus Femoris	Cycle/Run	61.8 (6.6)	60.5 (8.1)	66.0 (8.9)	62.8 (7.9)	0.032
	Run/Run	63.3 (11.5)	62.6 (9.8)	67.0 (8.6)	64.3 (10.0)	
	Marginal Mean	62.6 (9.1)	61.6 (9.0)	66.5 (8.8)		F-Interaction 0.447
	F- Time		3.361			
Vastus Lateralis	Cycle/Run	60.0 (6.2)	58.9 (4.8)	60.0 (6.5)	60.1 (8.5)	2.510
	Run/Run	59.0 (12.6)	55.0 (8.9)	58.7 (8.0)	57.6 (9.9)	
	Marginal Mean	59.5 (9.4)	57.0 (6.9)	59.4 (7.3)		F-Interaction 0.446
	F- Time		4.708**			
Vastus Medialis	Cycle/Run	60.3 (6.2)	58.9 (4.8)	60.0 (6.5)	59.7 (5.9)	3.391
	Run/Run	58.4 (7.4)	56.1 (7.3)	57.7 (6.4)	57.4 (7.0)	
	Marginal Mean	59.4 (6.8)	57.5 (6.1)	58.8 (6.5)		F-Interaction 0.137
	F- Time		2.145			

* Denotes significant difference ($p < 0.05$) between cycle/run and run/run.

** Denotes significant difference ($p < 0.05$) across times 1, 2 and 3.

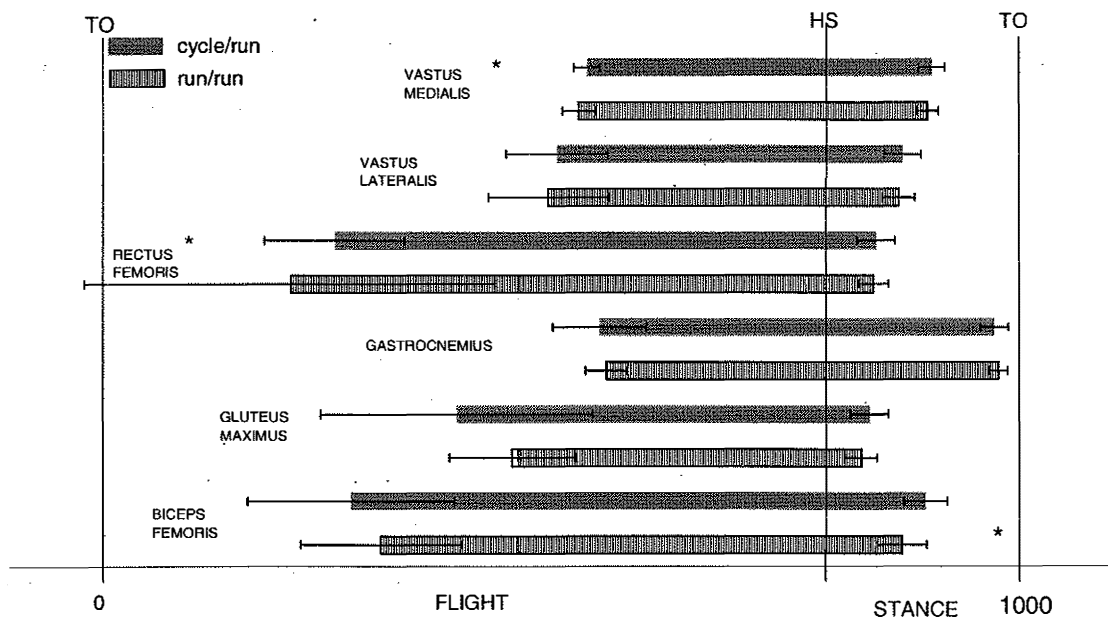


Figure 16. Normalised time of EMG activity for all six muscles for the cycle/run and the run/run expressed as percent of cycle (flight occupied 77% and stance occupied 23%) with standard deviations shown. TO is toe-off and HS is heel strike. Significant differences ($p < 0.05$) between conditions are indicated by *.

4.3 Level of Activation

Statistical comparisons of RMS values for muscle activation levels in flight (Table 7) found significant differences between condition for the VL ($F_{1,9} = 7.980$) with $16.7 \pm 2.4\%$ for cycle/run and $15.7 \pm 1.8\%$ for run/run. VM displayed a significant difference across time ($F_{2,18} = 4.785$) and post-hoc analysis via a dependent t-test found the difference occurred between time one (0km) and time three (2km) indicating a reduction in the level of activation of this muscle during flight as the run progressed.

The statistical analysis of RMS values for muscle activation levels during stance is shown in Table 8. Two significant differences were found to occur between conditions. The first of these was in the BF ($F_{1,9} = 6.354$) with values of $29.6 \pm 8.5\%$ for the cycle run condition and $33.5 \pm 6.1\%$ for run/run. The second was VL ($F_{1,9} = 7.334$) with $21.7 \pm 4.3\%$ for cycle/run and $19.6 \pm 3.1\%$ for run/run.

The differences found between conditions for the BF are represented via ensemble averages in Figure 17. Higher activation levels were recorded for the BF in the 2km run for the run/run condition. A difference in the level of activation between conditions was also found for the VL and is represented graphically in Figure 18. This clearly indicates a lower level of activation in the run/run condition for the VL.

Table 7

Means and standard deviations for level of activation during flight (%).

Variable	Condition	Time			Marginal Mean	F-Condition
		1	2	3		
Biceps Femoris	Cycle/Run	26.1 (4.2)	25.4 (4.7)	25.1 (5.3)	25.5 (4.7)	1.657
	Run/Run	23.8 (3.3)	23.7 (4.3)	23.1 (3.4)	23.5 (3.7)	
	Marginal Mean	25.0 (3.8)	24.6 (4.5)	24.1 (4.4)		F-Interaction 0.148
	F- Time		0.663			
Gluteus Maximus	Cycle/Run	19.1 (3.7)	18.6 (4.0)	17.7 (4.3)	18.4 (4.0)	0.111
	Run/Run	18.4 (7.3)	17.6 (5.1)	18.1 (6.3)	18.0 (6.2)	
	Marginal Mean	18.8 (5.5)	18.1 (4.6)	17.9 (5.3)		F-Interaction 0.233
	F- Time		0.381			
Gastrocnemius	Cycle/Run	14.9 (1.8)	12.3 (2.7)	13.2 (2.9)	13.5 (2.5)	0.648
	Run/Run	12.6 (2.6)	12.8 (2.7)	14.4 (4.0)	13.3 (3.1)	
	Marginal Mean	13.8 (2.2)	12.6 (2.7)	13.8 (3.5)		F-Interaction 2.170
	F- Time		2.275			
Rectus Femoris	Cycle/Run	23.3 (4.5)	22.0 (4.1)	21.8 (4.0)	22.3 (4.2)	1.767
	Run/Run	21.7 (2.8)	21.5 (2.2)	20.7 (3.7)	21.3 (2.9)	
	Marginal Mean	22.5 (3.7)	21.7 (3.2)	21.3 (3.9)		F-Interaction 0.261
	F- Time		1.646			
Vastus Lateralis	Cycle/Run	17.3 (2.5)	16.2 (1.8)	16.7 (2.8)	16.7 (2.4)	7.980*
	Run/Run	16.0 (1.7)	15.6 (1.9)	15.5 (2.0)	15.7 (1.8)	
	Marginal Mean	16.7 (2.1)	15.9 (1.9)	16.1 (2.4)		F-Interaction 0.385
	F- Time		1.373			
Vastus Medialis	Cycle/Run	14.9 (2.0)	14.9 (2.7)	14.3 (2.3)	14.7 (2.3)	0.651
	Run/Run	14.7 (2.2)	14.0 (1.6)	13.4 (1.2)	14.1 (1.7)	
	Marginal Mean	14.8 (2.1)	14.5 (2.2)	13.9 (1.8)		F-Interaction 0.275
	F- Time		4.785**			

* Denotes significant interaction between condition ($p < 0.05$).** Denotes significant interaction across time ($p < 0.05$).

Table 8

Means and standard deviations for level of activation during stance (%).

Variable	Condition	Time			Marginal Mean	F-Condition
		1	2	3		
Biceps Femoris	Cycle/Run	27.2 (7.7)	29.2 (9.4)	32.5 (8.4)	29.6 (8.5)	6.354*
	Run/Run	33.4 (7.5)	34.0 (6.3)	33.2 (4.5)	33.5 (6.1)	
	Marginal Mean	30.3 (7.6)	31.6 (7.9)	32.9 (6.5)		F-Interaction
	F- Time		2.008			1.080
Gluteus Maximus	Cycle/Run	23.1 (5.6)	21.9 (3.3)	21.1 (5.4)	22.0 (4.8)	0.587
	Run/Run	24.0 (6.6)	22.4 (6.8)	23.5 (3.2)	23.3 (5.6)	
	Marginal Mean	23.6 (6.1)	22.2 (5.1)	22.3 (4.3)		F-Interaction
	F- Time		0.216			1.196
Gastrocnemius	Cycle/Run	33.9 (6.7)	34.1 (6.5)	35.0 (6.6)	34.3 (6.6)	0.190
	Run/Run	36.5 (6.5)	35.7 (5.5)	38.6 (7.3)	37.0 (6.4)	
	Marginal Mean	35.2 (6.6)	34.9 (6.0)	36.8 (7.0)		F-Interaction
	F- Time		2.763			0.397
Rectus Femoris	Cycle/Run	26.8 (6.7)	25.9 (5.2)	27.8 (8.1)	26.8 (6.7)	1.454
	Run/Run	25.0 (5.6)	24.0 (3.1)	25.7 (5.4)	24.9 (4.7)	
	Marginal Mean	25.9 (6.2)	25.0 (4.2)	26.8 (6.8)		F-Interaction
	F- Time		0.799			0.007
Vastus Lateralis	Cycle/Run	22.4 (5.3)	21.3 (3.3)	21.3 (4.2)	21.7 (4.3)	7.334*
	Run/Run	18.7 (3.0)	19.5 (3.5)	20.5 (2.7)	19.6 (3.1)	
	Marginal Mean	20.6 (4.2)	20.4 (3.4)	20.9 (3.5)		F-Interaction
	F- Time		0.149			1.452
Vastus Medialis	Cycle/Run	22.1 (4.3)	21.1 (3.3)	20.4 (4.2)	21.2 (3.9)	1.432
	Run/Run	19.8 (4.7)	19.8 (4.2)	19.5 (3.2)	19.7 (4.1)	
	Marginal Mean	21.0 (4.5)	20.5 (3.8)	20.0 (3.7)		F-Interaction
	F- Time		1.208			1.303

* Denotes significant interaction between condition ($p < 0.05$).

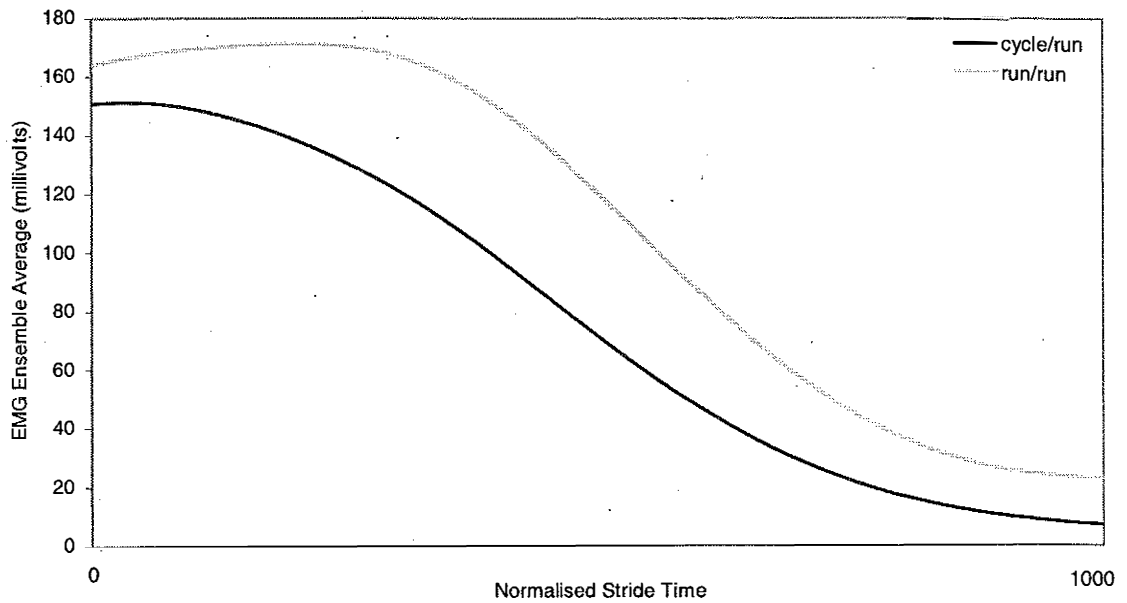


Figure 17. Ensemble average of EMG for the Biceps Femoris in stance for cycle/run and run/run conditions at time 3 (2km). Heel strike occurs at 0 whilst toe-off is at 1000.

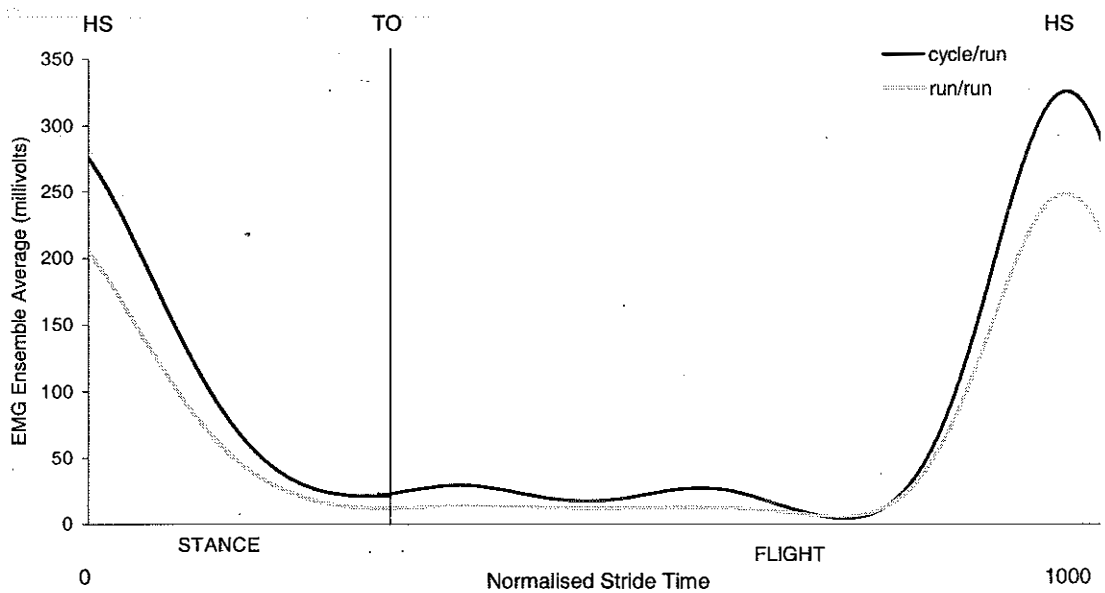


Figure 18. Ensemble average of EMG of the Vastus Lateralis showing muscle activation throughout an entire stride cycle for the conditions cycle/run and run/run with stance and flight phases shown.

CHAPTER FIVE

5.0 DISCUSSION

This study sought to collect data specific to triathlon by testing outdoors on a pre-designed and measured course to examine the effect of cycling on lower limb muscle activation in running. The variables quantified for examination were the time of muscle activation and the level of muscle activation. The repeatability of the EMG waveforms for each of the lower limb muscles measured over the 2km run from both the run/run and cycle/run condition was also examined. To the author's knowledge, limited research on muscle activation in running after cycling has been documented therefore, this discussion will draw from information in the literature for running and cycling individually in addition to kinematic findings on the transition from cycle to run.

Possible factors influencing the results are outlined prior to the discussion of the results obtained. The stride-to-stride repeatability is discussed followed by the findings for time of activation and level of activation for selected lower limb muscles. Finally, the practical implications for training of the triathlete are outlined.

5.1 Factors Influencing the Results

The results in this study may have been influenced by environmental conditions, the difficulty of controlling for fatigue when undertaking different types of activities, and a problem consistent with all quantitative research, equipment.

5.1.1 Environmental Conditions

Although this study simulated a triathlon by testing in the field, there was an absence of competitors, spectators and general triathlon competition atmosphere that may have altered the performance of the subjects.

Wind speed was monitored at the transition area and remained under 10km/h for all tests. Due to the size and distance of the course wind speeds may have varied at different points around the course and affected the subjects accordingly. Climate

changes within tests were inevitable due to the time taken for each test, approximately 2.5 hours for the cycle/run and 1.5 hours for the run/run.

5.1.2 Level of Fatigue in Running and Cycling

The surface EMG signal changes when muscles are fatigued (Redfern, 1992) and for this reason the cycle/run condition was compared with another condition run/run. Unfortunately, the levels of muscle activation and type of muscle contraction of the VL, VM, RF, BF, GS and GM are different in running than in cycling and therefore, it is impossible to expect that the extent and type (central or peripheral) of fatigue of these muscles after 40km of cycling can be equalled by 10km of running. For the purpose of this study, the results were considered assuming equal fatigue from the two conditions however, it is acknowledged that the levels of fatigue may have been different.

5.1.3 Equipment Issues

During the testing period two channels of the datalogger ceased to operate and testing was delayed until new channel cables were sent from the equipment manufacturer in Finland. On examination of the raw data, artefacts were found to exist in these channels for several subjects, which led to low sample numbers of unaffected data for the GS and GM muscles. A lack of significant differences for these muscles may have been a consequence of this. Due to these problems and the lack of any significant observations being available for GS and GM they will not be discussed in the following section.

The abovementioned equipment problems also delayed the completion of testing, which ended up being done over three to four months instead of the intended two-month period. This led to slight variations in the weather conditions for some subjects.

5.2 The Effect of Cycling on the Activation of the Lower Limb Musculature During Running

Previous studies examining the biomechanics of the cycle to run transition have concentrated on the variables of stride length, stride frequency, kinematic data, and EMG affecting the running stride (Hauswirth, Bigard, & Guezennec, 1997; Hauswirth et al., 2000; Lepers, Millet, Maffiuletti, Hauswirth, & Brisswalter, 2001; Quigley & Richards, 1996). This discussion will examine the significant differences found for the level and time of muscle activation in this study and discuss these in relation to previous kinematic and EMG findings.

The differences examined for this study are those measured in the first 2km of running after transition where the greatest variations in the kinematics of running after cycling have been recorded (Gottschall & Palmer, 2000).

5.2.1 Stride to Stride Repeatability

Examination of the CMC values obtained in each of the conditions revealed that the repeatability of the EMG waveforms for the selected lower limb muscles were very high for both conditions, cycle/run (stance 0.862-0.979, flight 0.880-0.986) and run/run (stance 0.821-0.982, flight 0.868-0.990). Previous examinations of CMC values in gait have been reported by Kadaba et al. (1989). They reported within-day CMC values of 0.851 for GM, 0.883 for VL, 0.856 for RF, 0.871 for VM and 0.837 for BF. The CMC values found in this study compare quite favourably with all EMG repeatability measures found in gait, and on average the results were higher than those obtained by Kadaba et al. (1989). The lower levels of repeatability found in the current investigation were from the GM and GS and this is expected due to the previously mentioned equipment problems.

5.2.2 Time of Activation

In agreement with the original hypothesis, that the time of activation of the sampled leg muscles during running after cycling would be significantly different, differences in the time of activation occurred between the cycle/run and run/run conditions for the BF, RF and VM. Further, the time of muscle activation of the VL changed within

the 2km test period highlighting a change in muscle activation during the initial transition phase.

5.2.2.1 Biceps Femoris

The time of activation of the BF during stance was significantly greater in the run/run condition compared to the cycle/run condition. Elliot & Roberts (1980) found an increased support time in running when fatigued, which may in turn increase the time of activation of the BF to extend the hip during the stance phase of running in the 2km run following the 10km run.

5.2.2.2 Vastus Lateralis

The activation time of the VL increased between times two (1km) and three (2km) during stance, highlighting an increased demand on this muscle in providing stability of the knee joint throughout the 2km run period. The VL plays a role in increasing joint stiffness by co-activating with the BF during stance (Kyrolainen, Belli, & Komi, 2001). The increased time of activation demonstrated by the VL may occur to provide improved stiffness as a precaution in stabilising the knee joint due to the change from non-weight bearing cycling to weight bearing running activity.

An alternative reason for the increased activation time of the VL may be to compensate for a more fatigued VM since VM has a slightly higher maximum activation level in cycling. VL and VM work synergistically in stance to stabilise the knee joint (Montgomery et al., 1994). In this study however, only the activation time of the VL in stance was significantly different between conditions. Clements, Yates and Curran (1999) in a study of triathlon injuries found the highest incidence of injury to occur on the lateral side of the knee in the running stage of triathlon. This may be a further implication associated with a longer time of activation of the VL.

5.2.2.3 Vastus Medialis

During flight VM was found to have a significantly greater time of activation during the cycle/run condition compared with the run/run condition. During the flight phase of running, the VM functions to extend the leg in late swing in preparation for foot

strike whilst exhibiting a maximum activation of 50% (Montgomery et al., 1994). This muscle is also highly activated in cycling with a maximum activation level of 54% (Ericson et al., 1985).

There are two possibilities for the extended time of activation of VM during flight. Firstly, the change from predominantly concentric contractions of the VM in cycling, where it is never fully extended, may impede its function in fully extending the knee at the end of flight during the run. Secondly, Hausswirth et al. (1997) found a greater degree of knee flexion to occur in the flight phase of running after cycling. As a result, a longer contraction of the VM may be required to extend the leg in preparation for foot strike. Although Quigley and Richards (1996) findings were not significant their results revealed slight increases in knee extension velocity in running after cycling, which may further relate to the increased time of activation of VM. Alternatively however, the small moment arm of the VM may mean that the increase in knee extension velocity may also be related to the decreased eccentric activation of the BF.

5.2.2.4 Rectus Femoris

RF showed a significantly greater time of activation during flight in the cycle/run condition when compared to the run/run condition. During flight, the RF has its highest activation of 22% of maximum as it flexes the hip (Montgomery et al., 1994). RF has a low peak activity of 12% during cycling where it functions as a knee extensor (Ericson et al., 1985). When running velocity is increased, the main function of the RF changes from knee extension to hip flexion where it is required to produce torque to bring the leg forward during swing (Nilsson et al., 1985). Since these subjects were running at close to race pace, it could be considered that the RF was acting more in hip flexion than in knee extension.

Hausswirth et al. (1997) found a greater knee angle during the flight phase in running indicating greater knee flexion. If this is due to an increased flexion velocity of the knee joint then the RF would need to contract eccentrically to control hip extension whilst contracting concentrically to flex the hip thereby requiring a longer contraction time. Alternatively, increases in the time of activation of the RF may be

due to an increased role in both flexing the hip and extending the knee during flight due to its synergistic role with the VL and VM.

5.2.3 Level of Activation

It was hypothesised that there would be changes in the level of activation of the sampled leg muscles during running after cycling. In this study significant differences in the level of activation were found between the conditions, run/run and cycle/run for the BF and VL. VM displayed a significant change in the level of activation over the 2km test period.

5.2.3.1 Biceps Femoris

A significantly higher level of activation during stance in the run/run condition was established for BF when compared to the cycle/run condition. In running, BF has previously been found to display a maximum level of activation of 52% during stance where its role is to initiate hip extension (Montgomery et al., 1994). The activation of the BF in cycling, in stark contrast to running, has been found to be quite low with a maximum activation level of 12% (Ericson et al., 1985). It is thought that the low activation levels experienced by the BF during cycling would allow this muscle to remain relatively fresh when moving into the run leg where its functional demands would be expected to be somewhat higher.

A greater forward lean of the trunk has been found in runners when fatigued (Elliot & Roberts, 1980) in addition to running after cycling (Hausswirth et al., 1997). A greater forward trunk lean has been found to increase the activation of the BF during the stance phase of running (Warrak et al., 1998). Therefore, it is thought that the higher demands placed on this muscle in the 10km run when compared with a 40km cycle, would have exhausted this muscle to a greater extent prior to the 2km run. Consequently, its level of activation would need to be higher to provide extension of the hip during stance. Pinniger et al. (2000) attributed a longer duration and higher level of muscle activation in the hamstrings to occur in a fatigued state, due to a decreased capacity of the muscles to continue to produce the necessary contractions to maintain function. It should be noted however, that under fatigued conditions the amplitude of the EMG signal increases for a given task (Redfern, 1992).

5.2.3.2 Vastus Lateralis

Activation level of the VL during stance was found to be significantly greater in the cycle/run condition in this study. In running it has been found that the VL contracts to stabilise the knee joint upon heel strike. According to Montgomery et al. (1994) it is during this initial impact that the VL has its maximum activation of 78%. The VL also plays a major role in cycling by providing propulsion with maximum activation levels of 50% (Ericson et al., 1985).

The increased level of activation of the VL in running after cycling, as shown in this study, highlights the demands associated with knee stability when changing from a non-weight bearing cycling activity to a weight bearing running activity. Kyrolainen et al. (2001) found an increase in the stiffness of the knee joint when running under fatigued conditions, which is thought to provide better force potentiation in push off due to increased “tendomuscular elasticity”. However, this would have affected the knee musculature under both conditions in this study and thus is not recognised as a major factor in the increased activation of the VL in stance.

The VL was also found to have a significantly greater level of activation during flight for the cycle/run condition. During flight the VL is responsible for extending the knee prior to foot strike (Nilsson et al., 1985). Maximum activation of the VL during flight in running (32%) has been found to occur in late swing as the knee is extending prior to foot strike (Montgomery et al., 1994). Extension of the knee may be made more difficult when changing from the concentric muscle activation in cycling, where the knee is never fully extended, to the stretch shortening cycle in running where its role in flight is to extend the knee. A study of the effects of cycling on running mechanics (Quigley & Richards, 1996) recorded faster levels of knee extension in running after cycling, implying a greater role of the knee musculature during the flight phase and hence an increase in activation of the VL.

5.2.3.3 Vastus Medialis

During the flight phase of running, the level of activation of the VM was found to reduce significantly between times one (0km) and three (2km). This highlights the initial difficulties with extension of the leg due to changes in muscle contraction types during the transition from cycling to running. Further, an adaptation of the

muscle to the changes in muscle contraction type, as time progressed, is highlighted by the subsequent decrease in the activation level of the VM during flight.

5.3 Training Implications

The findings from this investigation highlight changes associated with individual muscle function when changing from cycling to running. Millet, Millet and Candau, (2001) and Quigley and Richards (1996) found no specific reasons for triathletes to train specifically for the cycle to run transition however, it should be noted that testing in both these instances was undertaken in laboratory situations using cycle ergometers and treadmills.

The current investigation is to the author's best knowledge, the first of its kind to be undertaken entirely in the field and the findings of this study highlight a definite need for specific cycle to run transition training to be undertaken for Olympic distance triathlons. It is interesting to note that the subjects in this study were elite triathletes and it was previously found unnecessary for athletes at this level to train specifically (Millet et al., 2001).

Interestingly, Clements et al. (1999) found that 65% of the triathletes interviewed in their study had knee injuries from running, most of which occurred on the lateral side of the knee. Several other studies of triathlon injuries (Vleck & Garbutt, 1998; Williams et al., 1988) also found a high incidence of knee injury amongst triathletes. Unfortunately, these studies did not consider the transitional changes in relation to the aetiology of injury however, they found the highest incidences of knee injury tend to occur during running.

Manninen and Kallinen (1996) found that the change from a non-weight bearing activity such as cycling to weight bearing activity such as running tended to cause the most complaints and suggested that the change from a predominantly concentric to stretch shortening muscle contraction may be the cause of this. Therefore, the high levels of activation found in the VL, VM and RF in this study may be implicated in the common occurrence of knee injuries in triathletes.

5.4 Conclusions

Two hypotheses were proposed for this study. Firstly, it was hypothesised that there would be significant differences in the level and time of activation of the sampled leg muscles during running after cycling. This was found to be the case for the BF, VL, VM, and RF. Secondly, it was hypothesised that there would be a significant interaction between the conditions of cycle/run and run/run unfortunately this was not confirmed in this study. The following conclusions, within the limitations of the study, can be drawn from this investigation:

- BF is not highly activated in cycling and remains relatively fresh allowing a smooth transition from cycling to running.
- The change from concentric muscle activation in cycling to stretch shortening muscle activation in running affects the ability of the VL and VM to extend the knee in the flight phase of running.
- A reduction in the level of VM activation during the 2km run suggests an adaptation to the transition between cycling and running over time, where extension of the knee becomes progressively easier.
- The change from non-weight bearing cycling activity to weight bearing running activity compromises the normal function of the VL and VM highlighting difficulties in stabilising the knee joint.
- Due to biomechanical changes of the lower limbs, in running after cycling, it is necessary to train specifically for the cycle to run transition for an Olympic distance triathlon. Specific training may improve performance and reduce the risk of injury.

5.5 Recommendations for Further Research

To further investigate the effect of cycling on muscle activation in running more research is needed. Questions arising from this investigation and specific areas of importance include:

- Research into the effect of cycling on lower limb muscle activation in novice triathletes to determine the effects of experience on the transition from cycle to run.
- Explore changes in lower limb muscle activation during the transition from cycle to run in triathlons of varying length.
- Examine the effects of training specificity for the cycle to run transition and its effect on lower limb muscle activation in running after cycling.
- Investigate ways to equally fatigue different lower limb muscles in different activities so that studies may control for fatigue correctly.

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APPENDIX A

DOCUMENT OF INFORMED CONSENT

The effect of cycling on muscle activation in the running leg of an Olympic distance triathlon

Background Information

Anecdotal reports suggest that the transition from cycling to running in triathlon is the most difficult. A loss of coordination or the feeling of "jelly in the legs" is often described by athletes in this situation. Several factors have been implicated in this feeling, including the change from non-weight bearing activity, such as cycling, to weight bearing activity, such as running both of which have different types of muscle contraction.

The ability to run well off the bike in triathlon may well determine the success of the athlete to finish in a good position. In some Olympic distance triathlon events drafting has been legalised, this has increased the importance of the run leg in triathlon, making the transition from cycle to run of utmost importance.

A limited amount of research has been done regarding the muscle activation in running after cycling. Furthermore, what research has been undertaken has been done in laboratory situations. Specific measurements under realistic conditions will help to examine the true effects of this transitional period.

Purpose of the Proposed Research

The purpose of this research project is to determine the effect of cycling on muscle activation in running in an Olympic distance triathlon. The extent to which activation changes and the level of change over the three measured sections of the 2km run will be examined. To the investigators best knowledge studies of this kind have mostly been performed in a laboratory situation using a cycle ergometer and treadmill. This study will provide a more realistic measure that will better relate to the specifics of triathlon and the training associated with the sport.

Selection

Based on your performance at the recently completed state Olympic distance championships you have been selected as an appropriate candidate to undertake the study outlined below.

Methods and Procedures

You will be required to perform two separate tests, they being a 40km ride followed by a 2km run (cycle/run test) and then a 12km run (run/run test) a week later. The order in which you will perform these may vary. Each test will be conducted at your appropriate race pace for an Olympic distance triathlon.

Prior to testing, your skin will be marked with permanent pen for electrode placement, lightly abraded to remove dead skin cells, and wiped with alcohol. Surface electrodes will be attached (via tape) to the gluteus maximus (buttock), biceps femoris (hamstrings), rectus femoris, vastus lateralis, vastus medialis

(quadriceps) and gastrocnemius (calf). Data collection will begin at the commencement of the 2km run in the cycle/run test, and at the beginning of the 11th kilometre of the 12km run. As part of testing it is necessary for each subject to carry a portable EMG machine, weighing approximately 1kg, in a pouch with a strap around their waist.

From the collected data, three stages of six steps will be analysed for all muscles across each of the subjects. The three stages will allow analysis of the muscle activation changes within the 2km run period, this will be done for both tests and then the two tests will be compared via computer analysis. There are no risks over and above those normally encountered during a typical training session or race.

Time frame

The cycle/run session will take approximately two hours and the run session approximately one hour. Testing will commence in late June and continue in July and August until completed. It is hoped that most testing will be completed within the months of June and July.

Confidentiality of all subjects is assured. Individual feedback will be given to those athletes who request it. You have the right to refuse to participate in this study without prejudice, however if you decide to take part completion of both tests would be highly appreciated.

If you have any questions regarding this study feel free to ring Tamika Heiden on [REDACTED]

I wish / do not wish to participate in this study (please cross out as appropriate)

Athlete Name _____

Signature _____

Date _____

Investigator _____

Signature _____
