Characterization of arm muscle activity levels during cycling at various relative workloads

By

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Abstract

Arm cycling is an effective mode of rehabilitation, exercise, and transportation. Previous studies aimed at examining the neuromuscular control of arm cycling typically use a standard workload (e.g. 25W) as opposed to relative workloads for each participant. This may be problematic given that many measures of neuromuscular excitability are intensity-dependent and a standard workload likely represents different effort levels for each participant. The purpose of this study was to examine and characterize the arm muscles during arm cycling at various relative workloads. While the present thesis is not a detailed examination of the neuromuscular physiology of arm cycling it may be an important step in normalizing the manner in which arm cycling studies are performed, by determining how the muscles respond to increases in relative workloads during arm cycling. With the use of surface electromyography, it is possible to determine an appropriate relative workload. This will allow us to improve current basic research examining the neural control of arm cycling and may also be important for rehabilitative and therapeutic practices for individuals with a neurological injury or impairment.

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"It always seems impossible, until it's done" – Nelson Mandela

Table of Contents

List of Figures

List of Tables

List of Abbreviations

AD – anterior deltoid
BB – biceps brachii
BR – brachioradialis
CNS – central nervous system
CPG – central pattern generator
ECR-extensor carpi radialis
EMG – electromyography
FCR – flexor carpi radialis
iEMG – integrated electromyography
kg – kilograms
ms – milliseconds
mV – millivolts
MVC – maximum voluntary contraction
MU – motor unit
PPO – peak power output
RMS – root mean square
RPM – revolutions per minute
s – seconds
SCI – spinal cord injury
SD – standard deviation
SE – standard error
TB – triceps brachii

Introduction

The first "bicycle" was developed by a German named Baron Karl von Drais in 1817. It was a walking machine and was used as a means for quicker transportation in the royal gardens (Hug and Dorel 2009). In 1855, two French engineers added pedals, and thus the general design of the bicycle was born. Since this creation, millions use the bicycle for recreational or competitive cycling and for daily transportation. In addition to the many practical day-to-day uses of the bicycle, it is also used as a form of rehabilitation and exercise training given its' many health and fitness benefits such as improved cardiovascular fitness. This form of exercise is particularly important for individuals with motor impairments such as spinal cord injury (SCI). Persons with SCI (e.g. paraplegia) typically have very low physical capabilities due to obvious mobility impairments which can be attributed to the paralysis of the lower body, leading to a wheelchair-dependent life (Valent et al. 2008). Because of this, these individuals are at a higher risk of developing obesity, metabolic syndrome, diabetes, and cardiovascular diseases (Valent et al. 2008).

Many rehabilitation programs have introduced hand cycling as a means of exercise and mobility to help mitigate the many secondary health complications associated with SCI as mentioned above (Valent et al. 2008). Individuals with paraplegia are able to hand cycle, which is less strenuous than propulsion and is thereby advantageous for decreasing the risk of developing upper body overuse injuries (Valent et al. 2008). Therefore, arm cycling may be an alternative to training or mobility for individuals who suffer from a SCI. To fully appreciate and appropriately design exercise rehabilitation programs based on arm cycling, however, one must also understand various neuromuscular components of arm cycling such as the muscles being used, as well as their intensity and timing of activation (Hug and Dorel 2009). To address this issue a common means of assessing the neuromuscular system is through the use of electromyography (EMG). To that end, surprisingly little is known about how muscles in the arm are activated as the intensity of arm cycling increases.

Purpose of the Study

To characterize arm muscle activity during arm cycling at various relative workloads.

Research Hypotheses

There are 2 hypotheses in this study:

1) Electromyography activity will increase as the intensity of arm cycling increases.
2) There will be a positive linear relationship between electromyography and arm cycling workload during both the flexion and extension phases of the cycle.

Chapter 2 Review of Literature

2.0 Introduction

Studies utilizing isometric contractions to assess the neuromuscular system typically require participants to contract at an intensity made relative to their maximal voluntary contraction (MVC). During arm cycling, however, an absolute workload is normally used. This could significantly affect various neurophysiological outcome measures due to participants cycling at different relative intensities. To overcome this, it is essential to get participants to complete a cycling MVC which would consist of a 10-second maximal intensity arm ergometry sprint.

Leg cycling has served as a vital tool for investigation of multi-joint actions as an effective rehabilitation and training program for improving muscular function (Elmer et al. 2013). However, arm cycling is also used as a tool for clinical evaluations and exercise rehabilitation (Smith et al. 2008). The hand cycle (opposed to hand propulsion) has evolved into a major form of adapted sport, and it is practiced at a high level by many athletes (Valent et al. 2008). The hand cycle can easily attach to the handrim wheelchair, eliminating the physically demanding transfer to another mobility device (Valent et al. 2008). It has also been reported that hand cycling has a lower energy cost than handrim wheelchair propulsion (Valent et al. 2008).

There are many questions to be answered in regards to arm cycling and its use in rehabilitation settings and as recreational tools. For example, 1) is a standard workload an appropriate guideline for all individuals? and 2) what is the best workload for each individual? It is important to minimize the gap in literature on this topic, as it has potential to benefit individuals with a SCI or other motor impairments Therefore, additional research is required to determine the use or exclusion of the standard workload used. The objective of this chapter is to discuss the current literature examining arm cycling and the various mechanisms involved.

2.1 Assessment of muscle activity using electromyography

Electromyography (EMG) is used to measure the electrical signals produced by a muscle during a motor output (Konrad, 2005). EMG is produced by recording action potentials within the muscle fibres resulting from the depolarization and repolarization processes; after the initial excitation, the action potential travels along the muscle fibre at a velocity of 2-6 m/sec and passes the electrode site where the signal is recorded (Konrad, 2005). EMG has been accepted by the research community as an assessment tool for examining muscle activity and is widely used in sport and applied physiology related research (Hug and Dorel 2009). EMG can be recorded using several different techniques including measures that are invasive and use fine wires or needles inserted into the muscle, or non-invasive using surface electrodes that are placed on the skin overlying the muscle of interest (Cao et al. 2015; Hug and Dorel 2009). Each type of EMG recordings paradigm has its own benefits and drawbacks, meaning that one type of recording is typically chosen over the other based on the research question. Fine wire electrodes are commonly used in studies involving motor unit (MU) recordings. However, there are several issues related to the use of fine wire electrodes including a relatively small recording area which may not be representative of the total muscle mass involved in the exercise (Hug and Dorel 2009). These electrodes may also shift during a muscle contraction due to changes in muscle length which means that recordings would be made from different areas of the muscle and thus different MUs. Surface electrodes record EMG information from a larger surface area, thus allowing the simultaneous recordings of multiple MUs and as opposed to fine wire electrodes are not as susceptible to movement related artifacts. Perhaps most importantly, surface EMG can be used to investigate muscle activity during dynamic motor outputs (Hug and Dorel 2009). There are other factors that influence the surface EMG recordings as well. These include muscle tissue characteristics (e.g. skin and subcutaneous tissue.), physiological cross talk among the muscles (other muscles electrical activity), changes in the length between the muscle belly and electrode site, external noise, motion artifacts (induced by the movements of electrodes or cables) as well as the electrodes and amplifiers (Konrad, 2005; Hug & Dorel, 2009). The use of bipolar electrodes and a ground electrode and/or a proper placement of electrodes on the muscle of interest can help avoid crosstalk (Konrad, 2005).

The most important influence on the magnitude of the recorded signal is the recruitment and firing frequencies of the MUs which control the contraction process and modulation of force output in the muscle being investigated (Konrad, 2005). Human connective tissue and skin layers have a natural low pass filter effect on the original signal that is recorded. This means that some of the original signal produced from the muscle does not get through to the recording electrodes because of the barrier of the skin, subcutaneous fat, and the analyzed firing frequency recorded by the surface EMG may not accurately represent the original firing and amplitude characteristics.

2.1.1 The EMG/force relationship during isometric contractions

Physiological force production is associated with the electrical activity of the muscles involved which, as mentioned, is commonly measured using surface EMG. From a motor control perspective, the force generated during a contraction is mediated by the central nervous system (CNS; brain and spinal cord). Ultimately, however, it is the spinal motoneurone that must convert the electrical activity in the CNS to action potentials which then relay that electrical information to the muscles via the peripheral nerve. The amount of activity in the muscle will thus depend on input from the spinal motoneurone which is controlled by two main factors: the motoneurone recruitment and firing frequency (Cao et al. 2015). These two factors directly affect the EMG signal that is recorded.

Fuglevand et al. (1993) showed that the relationship between EMG and force is linear in the small muscles of the hand (which used for dexterity) but in other muscles they observed a non-linear shape as force and EMG increased. It is known that the relationship between EMG and isometric force production has two shapes. Fuglevand et al. (1993) observed a linear trend in the small muscles of the hand (which used for dexterity). They report that the force range differences in the recruitment of MUs may shape the EMG-force relationship (i.e. linear or non-linear) (Fuglevand et al. 1993). It is suggested that a linear relationship is seen in muscles with MU recruitment confined to a narrow force range (limit of recruitment <50% maximum excitation) and that a non-linear relationship is seen in muscles that recruit MUs over a broad force range (limit of recruitment >70% maximum excitation) (Fuglevand et al. 1993). The shape of action potentials, position of active MUs, instantaneous muscle length, rate of change in length, contraction history and various biomechanical factors will all affect the EMG and force relationship observed (Cao et al. 2015). The relative location of fast and slow muscle fibres within a muscle, as well as the location and distribution of these fibres relative to the electrodes are important to consider. Fast twitch muscle fibres generally have a larger diameter and display a greater range of action potentials compared to slow twitch muscle fibres; therefore, they will generate greater signal amplitudes (Kuriki 2012). The largest MUs, containing the largest diameter of fast twitch muscle fibres are recruited at a high force level following the recruitment of slow twitch muscle fibres, according to Henneman's size principle (Henneman et al 1947).

Woods and Bigland-Ritchie (1983) sought to determine whether the differences in the EMG/force relationship resulted from physiological and/or anatomical differences of motor unit organization. The muscles studied included adductor pollicis and soleus, which are composed primarily of slow twitch muscle fibres, and elbow flexors (biceps brachii) and extensors (long and lateral heads of triceps brachii), which are a mix of slow and fast twitch muscle fibres. The protocol included three or more brief (2-4 seconds) maximum voluntary contractions (MVCs) as well as a series of brief contractions at various submaximal force levels ranging from 10-90% of the MVC. Results from the slow twitch muscles (adductor pollicis and soleus) show EMG and force relationships to have significant linear relationship ($p < .01$), with a mean slope of 1.00 for adductor pollicis and 0.99 for soleus, and mean maximal integrated EMG (correlation coefficients of 0.99). The EMG/force relationship also had a non-linear component from 0-30% of the MVC force, with a linear relationship above this range for both the biceps brachii, and the triceps brahii. Brachioradialis and quadriceps followed a similar pattern (Woods and Bigland-Ritchie 1983)

To determine whether the non-linear relationships in the EMG/force relationship for biceps and triceps brachii were due to the electrode placement, recording configuration or limb position, Woods and Bigland-Ritchie (1983) compared the surface EMG variation with isometric force in the biceps brachii when using: a) monopolar and bipolar recording configuration; b) lateral and medial head electrode placement; and c) supinated and semi-pronated hand positions. In all of these cases, changes in the recording procedure did not substantially alter the EMG/force relationship. They determined that the EMG/force relationships were physiological and not caused by external mechanisms (i.e. electrode placement and lab configuration).

2.1.2 EMG normalization

To compare the muscle activity between different muscles or subjects, the data must be normalized because the data may vary between electrode sites, participants, and even day to day measures of the same muscles (Konrad, 2005). EMG is normalized by expressing the data of interest, EMG in this case, as a percentage of the maximum amount of the EMG recorded during a maximal task, usually an isometric MVC. Normalization allows the data to be standardized for all subjects within the study. This then allows a direct quantitative comparison of EMG findings between subjects and group statistics as well as between testing sessions. Normalized data can be developed and statistically verified (Konrad, 2005). When data is normalized, it provides the researcher with the estimation of neuromuscular effort that is needed for the given task (Konrad, 2005). If the data is not normalized to the reference value, or no reference value was recorded, then the researcher cannot make comparisons between subjects, groups, or testing sessions. This data is not a relative representation of the neuromuscular effort of the individual.

The choice of a normalization method is the first step to the interpretation of the signal. When choosing a reference value, the researcher should ensure that changes in the EMG signal reflect physiological modifications in the neural drive to the muscles (Rouffet and Hautier 2008). During dynamic movements, EMG is typically normalized to an isometric MVC reference, but it is generally recognised that the EMG from an isometric MVC is less reliable than the signal obtained from an isometric submaximal contraction and that it may not represent the maximum activation capacity of the muscle (Burden and Bartlett 1999). Literature examining the knee extensor muscles shows that the largest EMG signal is recorded when the knee extensors are in the mid-range of motion due to the force-length relationship of the muscle, is greater during concentric compared to eccentric contractions due to greater cross-bridge formation and increases with an increase in angular velocity during concentric contractions (Burden & Bartlett, 1999). Burden and Bartlett (1999) reported, however, that electrical activity from the biceps brachii is independent of the angle of elbow flexion during most of the concentric and eccentric contractions performed (20° to 100° of elbow flexion). The EMG signal of an isometric MVC may not represent the maximum activation ability of the muscle at either length other than those at which the MVC was performed.

In addition to changes in muscle length, normalizing the EMG to a maximal value obtained during an isometric contraction may not be appropriate for non-isometric motor outputs. Hautier et al. (2000), for example, recorded above 100% of the recorded isometric MVC for vastus lateralis during a maximal cycling exercise. This could be because there is more neural input from the sensory system during cycling that would enhance the output of the motor system as compared to an isometric contraction. For example, there is more Ia and type II afferent feedback during cycling, which will cause the muscle spindles to detect a change in the length of the muscle and contract more intensely. Muscle spindles are responsible for mediating the stretch reflex, as well as the coordination of movement, perception movement about the joint, and modulation of long-latency or transcortical reflexes. The most common normalization method used to analyze the neural drive during cycling is a series of isometric MVCs either on or off the cycle ergometer (Rouffet and Hautier 2008). However, this is not ideal because it is not obvious that the reference EMG signals during the isometric MVC can be used to represent the maximal neural drive during cycling (Rouffet and Hautier 2008). However, separately testing the muscles used in tasks and positions that are different from the cycling motion is time and energy consuming. Hunter et al. (2002) suggest using the first 5 seconds of a Wingate test as the normalization reference value for dynamic cycling. The Wingate test requires a similar motor output as submaximal cycling, with a difference in both cadence and intensity and may involve muscle input from synergist muscles (Hunter et al. 2002).

Rouffet and Hautier (2008) used an all-out torque-velocity test (lasting 10 seconds) performed on a cycle ergometer as an alternative normalization method to measure EMG amplitude for the reference value to correspond to the neural drive present during a maximal cycling exercise. The researchers reported that the repeatability of the measurements of peak EMG amplitude is comparable when using an isometric MVC and the torque-velocity normalization methods. The differences in the absolute amplitudes between the two methods may be due to modifications in the motivation status which may be responsible for a change in the number of motor units recruited. They also reported that EMG data obtained during the torque-velocity test are in line with those reported during submaximal pedaling exercises. The all-out torque-velocity test is less time and energy consuming and is as repeatable as an isometric MVC to measure peak EMG amplitude. This method of normalization likely reduces the impact of non-physiological factors (such as anatomic factors: thickness of subcutaneous tissues, shape of the volume conductor and detection system factors: skin-electrode contact) on the amplitude of EMG signals which allows quantifying more precisely the activation level of lower limb muscles and the variability of the EMG patterns during submaximal cycling (Rouffet and Hautier 2008)

2.2 Are all motor outputs controlled similarly by the CNS?

A rhythmic and alternating motor output such as locomotion, is initiated by the descending commands that excite spinal motoneurones, which causes the central pattern generator (CPG) to contract muscles and initiate movement (Copithorne et al. 2014; Forman et al. 2014). A CPG is a specialized network of neurons found in the CNS that produces rhythmic, coordinated patterns of movement in the absence of rhythmic external drive or phasic afferent feedback (Mazzocchio et al. 2008). Humans produce rhythmic motor patterns during all forms of locomotor movements (walking, running, cycling, and crawling.) (Zehr 2005). The activation of rhythmic limb movements can be triggered by the descending supraspinal commands that are related to the decision to initiate locomotion which pass on the task to the CPG that control the limbs (Zehr 2005). Sometimes, peripheral feedback may be strong enough to activate the CPG, and as soon as locomotion has begun, peripheral feedback from the moving limbs arrives at the spinal cord to inform the CNS to assist in the output of the CPGs (Zehr 2005).

The 'common core hypothesis' put forth by Zehr (2005) states that "all forms of rhythmic human movement share a similar neural control, which is composed of oscillatory neurons that drive the basic motor pattern". The idea is that it takes a tonic input signal and transforms it to produce a rhythmic output by the reciprocal inhibitory connections; activity in the flexor half-centre inhibits the activity in the extensor half-centre, and vice versa (Zehr 2005). Even though it is not possible to directly evaluate the contribution of CPG output to the rhythmic motor pattern, the probable contributions of CPG activity to the regulation of afferent feedback during the rhythmic movement can be estimated using reflexes (i.e. Hoffman-reflex or nerve stimulation) (Zehr 2005). The reflex modulation during leg movement has been attributed to CPG activity, and the similarity of reflex control during rhythmic arm and leg movements suggests there might be contribution from CPGs as well to control rhythmic arm movement (Zehr 2005). Rhythmic EMG activity and reflex patterns of the arms during rhythmic movement are consistent with the idea of a separate CPG for the control of each individual arm. Zehr (2005) states that this common core hypothesis is applicable for rehabilitation of locomotion after a stroke or spinal cord injury where the focus could be on general rhythmic movement combined with the specific movement training for recovery. It is suggested that arm cycling is CPG mediated, but there are multiple factors to consider during arm cycling that may influence the force outcome.

2.3 Factors affecting EMG patterns during cycling

Power output during cycling can be altered by changing the cadence, workload or both (Hug and Dorel 2009). The EMG activity recorded from various lower limb muscles during a progressive cycling test that was performed until exhaustion showed an increase in EMG activity level with respect to power output (Hug et al. 2003). This could be a result of an increase in central drive to the muscle, which would increase muscle recruitment, and increase the firing frequency of motor unit action potentials. Cadence is an important factor that affects cycling performance and therefore, many investigators have quantified the EMG activity in various muscles, at many different cadences. Ericson (1986) showed an increase in lower limb muscle activity as the cadence increased from 40 to 100 rpm. Neptune et al. (1997) also showed an increase in gluteus maximus (GM), biceps femoris (BF), semimembranosus (SM) and vastus medialis (VM) muscle activity in as cadence increased. On the other hand, some studies have shown no difference, or a decrease in EMG activity with increasing cadence. Sarre et al. (2003) reported no significant cadence effect on VL and VM EMG activity, meanwhile EMG activity for RF was higher at lower cadences. Lucia et al. (2004) reported a decrease in EMG activity for VL and gluteus maximus (GMax) with an increased cadence. These conflicting results could be attributed to the training level of the participants, range of rates tested across all of the studies, as well as the power output levels (Hug and Dorel 2009). Power output is both cadence and workload dependent. Arm cycling has typically been examined using a standard 25W workload, but that may not be the best workload for every individual. Changes in the cadence and/or workload will change the EMG activation that is recorded, and will vary amongst individuals.

2.4 What can we learn from studies on leg cycling?

Leg cycling has been an effective means of rehabilitation and training in a variety of populations such as athletes or individuals suffering from a severe injury such as an SCI (Elmer et al. 2013). There is wide-ranging literature on lower body cycling, with Houtz and Fischer (1959) being the first to record surface EMG during cycling. They recorded from 14 lower limb muscles, and also examined joint range in an effort to evaluate the effectiveness of using leg cycling for various clinical reasons. They had participants cycle with the seat at its lowest position, and then with it elevated 10cm and the workload set at different resistances. Their findings suggest that muscles are activated in a consistent pattern of activation. Saito, Watanabe, & Akima (2015) measured EMG from vastus intermedius (VI) and adductor magnus (AM) while participants cycled at 5 percentages of their maximal power output (20, 40, 60, 80 and 100%). Since VI and AM are deep muscles, the researchers combined ultrasonography and surface EMG to identify the superficial region and detect the activity of the two muscles. VI and AM account for 14% and 18% respectively of the volume of thigh muscles, and together they account for one third of the torque generated at the hip and knee joints (Saito et al., 2015). They found that the VI activates with QF and AM and that AM activates with QF and hamstring muscles while cycling. Ericson (1986) showed that a 120 W workload produced muscle activity of 45% of isometric MVC for VM, 44% for VL, and 32% for soleus (SOL), and that EMG activity is lower for biarticulate muscles, including rectus femoris (RF) and gastrocnemius lateralis (GL) compared to uniarticulate muscles. This 120 W workload is equivalent to approximately 54% of the maximum aerobic power (Ericson 1986). EMG recordings for deeper muscles such as tibialis posterior, adductor magnus, vastus intermedius, can only be done using intramuscular electrodes and because of its invasive nature, few studies have been conducted (Hug and Dorel 2009).

During leg cycling, the knee (VL and RF) and hip extensors (BF) are the most powerful (Bieuzen et al. 2007; Ryan and Gregor 1992). However, VM and gluteus maximus (Gmax) are single joint muscles and therefore are more consistent compared to biarticular muscles, and often are examined (Ryan and Gregor 1992). Synergists such as ST, SM, GM, SOL, and TA are also measured to determine their level of contribution to the task.

2.5 Muscle activity during arm cycling

The manner in which the neuromuscular system controls arm cycling is a relatively new area of research, with relatively little available literature focussed on characterizing arm muscle activation via EMG. This is a very important area because arm cycling has been introduced into many rehabilitation programs as a means of mobility and exercise (Valent et al. 2008). The upper body is comprised of a smaller muscle mass when compared to the lower body, and this causes different responses in both cardiovascular and muscular capacities (Elmer et al. 2013). This is important to consider when comparing to the lower body during cycling.

In the few studies conducted on arm cycling, the muscles of interest are typically the biceps brachii (BB), and triceps brachii (TB) as the primary power sources during flexion and extension, respectively. As well, anterior deltoid (AD) and brachioradialis (BR) are used as synergists. One group reported on abdominal activation during arm cycling and recorded EMG from the abdominal muscles and/or back muscles (Elmer et al. 2013).

2.6 The role of each muscle during cycling

There is minimal literature examining the role of the arm muscles during arm cycling. Zehr and Chua (2000) reported phasic and reciprocal muscle activation patterns between flexor and extensor muscles are seen at the wrist, elbow, and shoulder. This is comparable to the EMG patterns observed during rhythmic lower limb movements such as walking. There are similarities in the muscle activation patterns for upper and lower limb during cycling (Zehr and Chua 2000). It is suggested that during cycling, the shoulder joint acts like the hip, the elbow acts like the knee, and the wrist acts like the ankle (Zehr and Chua 2000).

Based on literature examining the lower limb during cycling, we know that the ankle is not considered a major power-producing joint but it is significant during cycling (Ryan and Gregor 1992). The gastrocnemius (GAST) and SOL muscles are responsible for plantar flexing the ankle joint during cycling, with a phase difference between the two muscles. GAST is activated during the power phase, while SOL peaks prior to GAST (Ryan and Gregor 1992). Uniarticular knee extensors (VM and VL) display consistent patterns, with both muscles being highly synchronized. Although an increase in EMG activity does not always represent an increase in force production, these two muscles have a difference in force production during late recovery and the subsequent power phase (Ryan and Gregor 1992). The RF muscle produced higher activation levels than the vastii muscles and varied during cycle. RF is a biarticular muscle (it crosses both the knee and hip joints) and because of this, it has a decrease in activation levels earlier in the power phase (Ryan and Gregor 1992). It is also shown that the knee extensor muscles share the load during the power phase, with the vastii muscles taking most of the responsibility (Ryan and Gregor 1992).

During the latter part of the recovery phase, the vastii muscles are actively stretched, which may increase the output during the power phase when the muscles are activated then shortened. Ryan and Gregor (1992) reported that the hamstring muscles (SM, ST, and BF) are also active during the power phase, resulting in a coactivation with the knee extensors. They reported SM and BF to be more active during the early part of the power phase, with ST being slightly delayed. As activation levels increase, variability also increases (Ryan and Gregor 1992). BF was reported as showing two separate activation patterns: 1) activation throughout the power phase and early recovery with an increase prior to initiation of power phase, and 2) activation and then relative inactivity during recovery. The difference in activation patterns for BF could be attributed to load sharing amongst the hamstrings (Ryan and Gregor 1992). In this study, gluteus medius (Gmed) was relatively consistent, which supports it's important role in hip extension and contributing to power output during cycling (Ryan and Gregor 1992). Gmed variability was also higher than the vastii muscles. The increased variability for some muscles could be attributed to the fact that uniarticular extensor muscles are the power producers and biarticular muscles are power distributors during cycling. The power producing muscles show less variability in their activation patterns compared to the power distributing muscles. If muscles are required to contribute at a higher percentage maximal, the neural drive to that muscle should be consistent (Ryan and Gregor 1992).

Our main muscles of interest during arm cycling are the biceps brachii (BB) and triceps brachii (TB) as they are the main elbow flexors and extensors, respectively. We are also interested in the anteriod deltoid (AD), brachioradiais (BR), flexor carpi radialis (FCR) and extensor carpi radialis (ECR) as they are contributors during arm cycling. It is known that the BB muscle assists in shoulder flexion, assists with shoulder abduction when the arm is laterally rotated, flexes the elbow joint most effectively when the forearm is supinated, strongly supinates the forearm from a pronated position when the elbow is at least partly flexed, but not when it is extended, and the short head assists in horizontal adduction of the arm across the chest (Simons & Travell, 1999). The TB muscle is the main elbow extensor. However, the long head adducts and extends the arm at the shoulder joint (Simons & Travell, 1999). The AD muscle flexes the arm forward and horizontally adducts the arm across the chest (Simons & Travell, 1999). The BR muscle used to be thought as the primary supinator of the forearm, but is now known as an elbow flexor and upon stimulation, it will bring the forearm to a neutral position whether from supination or pronation (Simons & Travell, 1999). FCR flexes the hand and abducts the hand at the wrist joint (Simons & Travell, 1999). ECR extend and abduct the hand, and ECR longus assists in elbow flexion (Simons & Travell, 1999).

2.8 Conclusion

Pedaling is not a simple movement, and upper body cycling is a very important form of exercise for many populations. It is a new area of research and therefore this is the first of its kind to be conducted. The effect of contraction intensity on muscle activity in the lower limb has been extensively researched. Both linear and non-linear relationships have been observed (Lippold 1952; Woods and Bigland-Ritchie 1983; Fuglevand et al. 1993). Muscle activity of the arm muscles (i.e. biceps brachii and triceps brachii), however, have not been reported during arm cycling. It is necessary to investigate the characteristics of the muscle activity of the arm muscles during arm cycling in order to gain a better understanding of muscle activation patterns. Increasing the number of muscles studied, as well as the various relative workloads, will further develop the knowledge of the neural control system during locomotion.

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Chapter 3

Workload-dependent changes in arm muscle activity levels during arm cycling at different relative workloads

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Running Head: Workload-dependent changes in arm muscle activity during arm cycling

Keywords: electromyography, biceps brachii, triceps brachii, pedalling, arm ergometry

3.0 Abstract

Arm cycling is commonly used in rehabilitation settings for individuals with upper- and/or lower-limb motor impairments. It is thought to induce neural plasticity that may lead to increases in motor function in the affected limb(s). Arm cycling studies typically use absolute workloads for all participants whereas it is standard practice in studies using isometric contractions for participants to contract at relative intensities based on their isometric maximal voluntary contraction. This allows comparison between participants based on relative force outputs. The same does not occur during arm cycling studies. Thus, the objective of this study was to characterize arm muscle activity during arm cycling at different relative workloads. Participants (n=11) completed a 10-second maximal arm ergometry sprint to determine peak power output (PPO) followed by 11 randomized trials of 20-second arm cycling bouts ranging from 5-50% of PPO (5% increments) and a standard 25W workload. Electromyography (EMG) was recorded from the biceps brachii and triceps brachii bilaterally in 11 participants, and from anterior deltoid, brachioradialis, flexor carpi radialis (FCR) and extensor carpi radialis (ECR) of the dominant arm in participants. Results show a linear relationship between iEMG and workload for biceps and triceps brachii during the flexion and extension phases of arm cycling. There were significant main effects for Position ($F_{(1,10)} = 105.363$, $p < 0.001$) and Intensity ($F_{(2.722, 27.224)} = 59.435$, $p < 0.001$) for biceps brachii, with flexion having significantly higher iEMG activation levels. There were no significant main effects for Position ($F_{(1,10)} = 1.362$, $p = 0.270$) but there was a significant main effect for Intensity ($F_{(2.060, 20.603)} = 65.015$, $p < 0.001$) for the triceps brachii, with extension having higher iEMG activation levels. In summary, iEMG amplitudes increase as the workload increases. There is an increase up to 35% in the

biceps brachii, and up to 30% in the triceps brachii, followed by a plateau in iEMG. The increased iEMG is a result of increased recruitment and firing frequency.

3.1 Introduction

Arm cycling is used as a means of exercise in rehabilitation programs for individuals with upper and/or lower limb impairments, following for example, a stroke or a spinal cord injury (SCI). The aim is often to induce neural plasticity that could lead to a regain of neural connections in the affect limb(s). Given the importance of arm cycling to rehabilitation and the knowledge that exercise-induced adaptations including those of neural origin are often intensity-dependent, surprisingly little information is available regarding how different intensities of arm cycling influence the activation of the arm musculature.

It is well known that as muscle contraction intensity increases, so too will the EMG that is recorded. EMG increases represent increased output from the spinal motoneurone pool required to activate the muscle as force output increases. The changes in EMG associated with increased force output have been shown to have both linear (Lippold 1952; Woods and Bigland-Ritchie 1983) and non-linear relationships (Woods and Bigland-Ritchie 1983) in a number of muscles during isometric contractions. The relationship between muscle activity and dynamic muscle contractions is more challenging, however, due to numerous physiological and non-physiological factors. For example, the electrode placement during a dynamic contraction may change throughout the motor output resulting in EMG recordings from different motor units. Regardless, linear relationships between peak velocity and acceleration with the EMG amplitude of various muscles, such as the triceps brachii, elbow flexors and plantar flexors have been demonstrated (Farina 2006). Regardless of the type of contraction utilized to assess motor output and even with the many issues related to data acquisition and interpretation, particularly during dynamic contractions, surface EMG does provide general information regarding the level of muscle activation.

Relatively little, however, is known regarding how EMG changes over various workloads (power outputs) during arm cycling. This is an important distinction for several reasons including the fact that arm cycling is bilateral, involves both shortening and lengthening contractions and is under different neural control than isometric contractions for a given level of EMG output (Forman et al. 2014). There are only three studies that have examined the influence of workload on EMG of the arm musculature during cycling (Bernasconi et al. 2006; Hundza et al. 2012; Spence et al. 2016). While EMG increased in each study with an increase in workload, specific information such as phase-dependence or activation pattern was not provided (Bernasconi et al. 2006; Hundza et al. 2012) and/or there were minimal workloads utilized (Spence et al. 2016).

The purpose of the present study was to characterize arm muscle activity during arm cycling at different workloads. We were particularly interested in characterizing the activity of the biceps and triceps brachii given that these muscles are actively involved in arm cycling and appear to demonstrate strong phase-dependency. We hypothesized that 1) electromyography activity will increase as the intensity of arm cycling increases and that 2) there would be a positive linear relationship between EMG and arm cycling workload during both the flexion and extension phases of arm cycling in the biceps and triceps brachii.

3.2 Methodology

3.2.1 Ethical Approval

The procedures of the experiment were verbally explained to each volunteer prior to the start of the session. Once all questions were answered, written consent was obtained. This study was conducted in accordance with the Helsinki declaration and approved by the Interdisciplinary Committee on Ethics in Human Research at Memorial University of Newfoundland (ICEHR#: 20150140-HK). Procedures were in accordance with the Tri-Council guideline in Canada and potential risks were fully disclosed to participants.

3.2.2 Participants

Eleven apparently healthy individuals (six males and five females, 25.2 ± 4.4 years of age, 73.6 ± 7.8kg, nine right-hand dominant, two left-hand dominant) were recruited for this study. Participants had no known neurological impairments. Prior to the experiment, all participants completed a Physical Activity Readiness Questionnaire (PAR-Q+) to screen for any contraindications to exercise or physical activity and an Edinburgh Handedness Inventory checklist. Participants were required to refrain from any heavy exercise, especially upper body exercise, 24 hours prior to the start of testing.

3.2.3 Experimental Procedure

Participants attended a familiarization session to practice arm cycling sprints that were required during the experimental session to determine peak power output (PPO). This session was followed by an experimental session with a minimum of 24hrs between. During the experimental session participants first completed a 5-minute warm-up using a Monark cycle ergometer (Ergomedic 894 E), with only the 1kg weighted basket as resistance, at a self-selected pace. The ergometer was securely mounted to the top of a table and fitted with hand pedals. Following the warm-up and a 5-minute rest break, participants performed a 10 second maximal arm ergometry sprint using 5% of their body weight as the resistance to determine PPO. Results of this cycling trial were then used to determine the relative intensity for all subsequent trials. Following a minimum 10 minutes post-sprint rest period, participants were moved to a SCIFIT cycle ergometer (model PRO2 Total Body) to perform arm cycling at 11 different intensities, 10 of which were made relative to the PPO and one which was done at 25W. The 25W condition was constant for all participants, given that 25W is a common workload used during arm cycling studies (Bressel et al. 2001; Forman et al. 2014). The remaining 10 trials were randomized and performed at relative intensities ranging from 5-50% of the PPO. For all trials participants cycled at a constant cadence of 60 rpm for 20 seconds.

3.2.4 Experimental Set-up

Participants were seated upright at a comfortable distance from the hand pedals, so that during cycling, there was no reaching or variation in trunk posture (Fig. 1). To further ensure that posture was maintained throughout all trials, each participant was strapped securely to the ergometer seat with straps placed over the shoulders and across the chest. Movement of the shoulders and arms was not impeded. The hand pedals of the ergometer were fixed 180 degrees out of phase and the seat height was adjusted so that the shoulders of each individual were approximately the same height as the centre of arm crank shaft. Participants gripped the ergometer handles with the forearms in a pronated position.

Cycle crank positions were made relative to a clock face (12,3,6, and 9 o'clock, as viewed from the right crank arm) with the "top dead centre" position of the crank arm defined as 12 o'clock and "bottom dead centre" as 6 o'clock. The biceps brachii and triceps brachii were the main muscles of interest, thus the terminology used to describe the cycling movement is based on the position of the dominant elbow joint. Elbow flexion was defined as the movement from 3 to 9 o'clock, while the hand was moving toward the body. Elbow extension was defined as the movement from 9 to 3 o'clock, while the hand was moving away from the body. There were magnets positioned at 3 o'clock and 9 o'clock on the SciFit Bicycle in order to enable crank position to be tracked during cycling. When the crank passed the magnets at the 3 o'clock and the 9 o'clock positions, a 5 volt pulse was sent from the SciFit Bicycle to the data collection software. This pulse was recorded and used to track crank position through-out all cycling trials.

3.2.5 Electromyography Recording

EMG of the biceps brachii, lateral head of the triceps brachii, anterior deltoid, brachioradialis, flexor carpi radialis (FCR), and extensor carpi radialis (ECR) of the dominant arm, and biceps brachii and triceps brachii of the non-dominant arm were recorded using pairs of surface electrodes (Medi-Trace 130 ECG conductive adhesive electrodes). The inter-electrode distance was 2cm and all electrodes were aligned to fiber direction of the target muscles. A ground electrode was placed on the lateral epicondyle. Prior to electrode placement the skin was thoroughly prepared by shaving any hair and the removal of dead epithelial cells (using abrasive paper) followed by sanitization with an isopropyl alcohol swab. Muscle activation data was collected at 2000 Hz using the BIOPAC MP-100 data acquisition system with Acknowledge 4 software and an EMG100C differential amplifier (CMRR 110dB (50/60Hz), input impedance 2MΩ, bandpass filer 10Hz – 500Hz). Data obtained during the experiment were analyzed offline using code written in Visual Basic.

3.2.6 Data analysis

The EMG data were amplitude normalized by dividing the raw EMG during cycling by the muscle specific maximum EMG from the 10-second maximal arm ergometry sprint. The maximum EMG amplitude was determined using a 100ms RMS moving window (as per Burden and Bartlett (1999) to process the raw EMG from each muscle over the duration of the 10-second sprint. The resulting smoothed signal was examined to determine the peak EMG for each muscle, which was then use to amplitude normalize all sub-maximal cycling trials.

The submaximal cycling trials were then analyzed by examining the first 10 seconds of data from each trial. These 10 seconds of data were divided in to sections that represented one complete revolution of the crank handle (from 3 o'clock to 3 o'clock). Each revolution was further broken down in to an elbow flexion phase (from 3 o'clock to 9 o'clock) and an elbow extension phase (from 9 o'clock to 3 o'clock). This was done using the magnet signal described above. Figure 2 provides a sample of the raw data collected from biceps and triceps in addition to the signal from the 3 o'clock magnet. Further details of the windowing method used to partition EMG data are provided in that figure. For most individuals, a total of 10 revolutions were completed during the 10 seconds of cycling. Once the data was windowed, integrated EMG (iEMG) was calculated for the following time periods: the full revolution, the flexion phase and the extension phase. Trapezoid rule was used for these calculations (Winter and Patla 1997).

To assist with the visual presentation of the data, linear envelope, ensemble average EMG was calculated for each arm cycling intensity. This was done using the following steps:

1. Raw, amplitude normalized and windowed EMG was full wave rectified and low pass filtered at 10 Hz using a fourth order dual-pass butterworth filter. Only EMG data from complete revolutions was used for this.
2. The data was then rubberbanded to normalize it to time. One revolution was considered 100% of the whole cycle with the time period from 3-9 being fit to the first 50% (flexion) of the rubberbanded signal and 9-3 to the last 50% (extension).
3. These rubberbanded trials were then averaged across all trials for each intensity. The end result was an average linear envelope for each muscle at each intensity.

3.2.7 Statistics

All statistical analysis was performed using IBM's SPSS Statistics Version 23. Assumptions of sphericity were tested using the Mauchley test, and if violated, the Greenhouse-Geisser estimates of sphericity correction was applied to the degrees of freedom. Separate two-way (position x intensity) repeated-measures ANOVAs were used to assess the iEMG of each muscle during two phases (flexion and extension) and 11 different workloads (25 W and percentages of peak power output). To determine whether the relationship between iEMG and intensity was best described as linear during both phases of arm cycling, a series of twelve repeated-measures one-way ANOVAs were conducted for each muscle examined using Polynomial Contrasts (i.e., linear, quadratic or cubic). Trends were determined by examining the F-values of each of the 3 models as well as the observed power. All statistics were run on group data and a significance level of $p < .05$ was used. All data are reported in text as means ± SD and illustrated in figures as means ± SE.

3.3 Results

3.3.0 iEMG of recorded muscles during arm cycling

The table below summarizes the findings of 6 muscles from the dominant limb. Biceps brachii, anterior deltoid, brachioradialis, and FCR had significant main effects for position, intensity, and an interaction between factors. Triceps brachii and ECR had a significant main effect for intensity. Position and the interaction between the factors were not significant.

3.3.1 Muscle activity patterns of the biceps and triceps brachii

As previously mentioned, the main muscles of interest in our laboratory are the biceps and triceps brachii. As such, we have chosen to show the level and pattern of activation of those two muscles. To show the EMG activity pattern of the biceps brachii throughout arm cycling the ensemble averaged LE EMG was examined (Fig. 3). From this figure it is evident that the biceps brachii is very active during the flexion phase (3 to 9 o'clock) and relatively inactive during the extension phase (9 to 3 o'clock). The triceps brachii appears to be highly active during the extension phase, however, as opposed to the biceps brachii there is more of a biphasic activation pattern, with the muscle also being active during the flexion phase (Fig. 4). It is also clear that as the intensity of cycling increased so too did the EMG activation level in both muscles.

3.3.2 iEMG of the biceps brachii is phase- and intensity-dependent

There were significant main effects for Position ($F_{(1,10)}$ = 105.363, $p < 0.001$) and Intensity ($F_{(2.722, 27.224)}$ = 59.435, $p < 0.001$) (Fig. 5). Flexion had significantly greater iEMG (M = 4.904mV; SD = 1.443mV) than extension (M = 0.942mV; SD = 0.312mV) and as intensity increased, iEMG significantly increased up to 35%; 35-50% did not result in significantly different EMG activation levels. There was a significant interaction effect between both Position and Intensity ($F_{(2.977, 29.775)}$ = 41.737, $p < 0.0001$). The increase in iEMG as intensity increased was significantly different for flexion versus extension; flexion had higher iEMG at all levels of intensity compared to extension.

3.3.3 iEMG of the triceps brachii is intensity-, but not phase-dependent

There were no significant main effects for Position ($F_{(1,10)}$ = 1.362, $p = 0.270$) but there was a significant main effect for Intensity ($F_{(2.060, 20.603)}$ = 65.015, $p < 0.001$) (Fig. 6). Extension had greater iEMG (*M* = 5.793mV; SD = 2.046mV) than flexion (*M* = 4.905mV; SD = 2.507mV). There was no significant interaction effect between Position and Intensity ($F_{(1.516, 15.165)}$ = 2.246, $p = .148$). There was no significant difference between phases as intensity increased. Similar to that of biceps and triceps brachii the main effect of Intensity is apparent for Intensity level changes up to 30% vs. 35% (i.e., levels over 30% Intensity do not differ significantly from each other); the effect of Intensity on iEMG tapers off in both flexion and extension at higher workloads.

3.3.4 There is a linear relationship between iEMG and workload for biceps and triceps brachii during both phases (flexion and extension) of arm cycling

Table 2 shows that the relationship between iEMG and workload is linear for all muscles examined during both flexion and extension phases of arm cycling. We furthered our analysis of the biceps and triceps brachii, our main muscles of interest, by conducting correlation analysis between iEMG and workload for biceps and triceps brachii as seen in Figs 7 and 8, respectively. We also compared the slope of the linear relationships between flexion and extension using a paired t-test to assess if the gain in iEMG was different between phases within a muscle. The slope was significantly different between flexion and extension for the biceps brachii (steeper during flexion than extension; $p < 0.001$) but not triceps brachii ($p = 0.15$) (see Figs. 7 and 8, respectively).

3.4 Discussion

This is the first study to characterize muscle- and phase-dependent activity levels of the arm muscles during arm cycling at different relative power outputs. In this report we show that as arm cycling intensity (power output) increased, there was, as expected, a concomitant increase in EMG activity in each of the muscles examined. Given our labs interest in the neural control of the biceps and triceps brachii musculature during arm cycling (Copithorne et al. 2015; Forman et al. 2014; Forman et al. 2015; 2016a; Forman et al. 2016b; Power and Copithorne 2013; Spence et al. 2016), we were particularly interested in the phase- and workload-dependent changes in those muscles during arm cycling. Interestingly, we show that the biceps brachii demonstrated a strong phase-dependence in EMG activity whereas the triceps brachii did not.

3.4.0 Patterns of activity in the biceps and triceps brachii during arm cycling

The bicep brachii is bi-articular (i.e. crossing two joints – elbow and shoulder) and contributes to elbow flexion. Thus, as expected, the biceps brachii was highly active during elbow flexion and relatively inactive during elbow extension. The long head of the triceps brachii is bi-articular (i.e. shoulder and elbow), while the medial and lateral heads of the triceps brachii are mono-articular (i.e. elbow). Because we recorded from the lateral head of the triceps brachii it was surprising to find that its activity level was not different between the elbow flexion and extension phases of arm cycling given that its' role is to extend the elbow. Our previous work has shown a phase-dependent modulation of triceps brachii activity, however, the difference between the studies in the manner in which the muscle activity was assessed. The present study assessed EMG activity during flexion (3 to 9 o'clock) and extension (9 to 3 o'clock), however our previous work assessed EMG at mid-flexion (6 o'clock) and mid-extension (12 o'clock). These points in time during a full cycle represent very different activity patterns (Figs 4).

The triceps brachii may be active during elbow flexion simply in an attempt to maintain joint stability about the elbow, especially given that this type of motor output is not one that is performed every day. Though we are currently unsure as to why the triceps brachii is active during the flexion phase, recent work showed a similar bi-phasic activation pattern of the triceps brachii that was abolished following arm cycling training in persons with spinal cord injury (i.e. the triceps brachii activity was absent during flexion) (Brousseau 2016), suggesting that a learning response occurs over time and/or practice/training. Thus, the activation of the triceps brachii during the flexion phase of arm cycling could be considered inappropriate and non-functional. This remains to be examined.

3.4.1 EMG increases as power output increases

As expected, muscle activation levels as assessed via iEMG increased as each arm cycling workload increased for all examined muscles (Table 1). With respect to the muscles controlling elbow function (our main muscles of interest), we show a significant effect for intensity (workload) up to 35% of PPO for the biceps brachii, and up to 30% for the triceps brachii, followed by a plateau in both muscles. The increases in iEMG amplitudes reflect the increased recruitment and firing frequency of the motor units from which we recorded. The biceps and triceps brachii are composed of approximately 50% slow and fast twitch fibres (Johnson et al. 1973). Based on the size principle (Adrian and Bronk 1929; Henneman 1957) as the cycling intensity increased additional motor units, including larger faster motor units, would be recruited to assist with force production resulting in an increase in the EMG amplitude.

Only three previous studies have examined the influence of workload on EMG of the arm muscles during arm cycling (Bernasconi et al. 2006; Hundza et al. 2012). Bernasconi et al. (2006) reported an increase in EMG with increased workload in each of the muscles examined (biceps brachii, triceps brachii, anterior deltoid, and infraspinatus) during an arm cycling VO_2 max test. They suggested the increased EMG amplitude as cycling intensity increased was due to the recruitment of additional type 2 muscle fibres during the testing protocol. Their objective, however, was not to give a detailed description of the EMG activation levels as arm cycling intensity increased. For example, they did not examine the EMG activation levels during various phases of arm cycling, nor did they show representative traces of EMG activity. Similar findings were reported by Hundza et al. (2012), (i.e. increased EMG with increased arm cycling workloads). As was the case in the Bernasconi report, however, they did not assess the phase-dependence or pattern of arm muscle EMG, as that was not their intent. Finally, a recent study from our lab examined corticospinal excitability to the biceps and triceps brachii during arm cycling at two different workloads (Spence et al. 2016). In that study, however, only the biceps and triceps brachii EMG were reported at two different workloads (5 and 15% PPO).

3.4.2 EMG-power output relationship

The increase in EMG in the biceps and triceps brachii as workload increases is best described as a linear relationship (Fig. 7) though there appears to be a general plateau once the PPO reaches 35% of maximal activation as also indicated by the lack of statistical difference in EMG above 35% PPO. As already discussed it is clear that the increased EMG is a result of increased recruitment and firing frequency, however there was also a relative plateau in EMG at the higher workloads. It may be that the biceps brachii motoneurone pool were fully recruited and/or firing at their maximal rates. The continued increase in power output may thus not rely on the biceps brachii, keeping in mind that arm cycling is a bilateral motor output that involves multiple muscles. Thus, there are many potential muscle synergies at play to produce a given power output. These findings are in general agreement with previous work using isometric contractions to characterize the EMG force relationship. Studies assessing the relationship between force and EMG have shown that as workload increases, EMG also increases in both linear (Lippold 1952; Woods and Bigland-Ritchie 1983) and non-linear relationships (Woods and Bigland-Ritchie 1983). Lippold (1952) had 30 participants complete contractions at 10 different force outputs in the gastrocnemius muscle in order to examine the EMG force relationship. He reported that the relationship was linear – as force output increased so too did surface EMG. Subsequent work has demonstrated linear relationships between force and EMG during isometric contractions in multiple muscles. Importantly, Moritani et al. (1978) reported a linear relationship between workload and EMG in the right elbow flexor muscles during several submaximal contractions.

3.4.3 Gain of the EMG force relationship during flexion and extension

In the biceps brachii it is also noted that the slope of the line of best fit is steeper during flexion than extension while there was no difference between flexion and extension for the triceps brachii. These results are not unexpected given the findings previously mentioned regarding iEMG amplitudes during the different phases (different for biceps brachii but not triceps brachii). There is one interesting observation however in the triceps brachii data. Though the overall slopes (all PPOs) are not significantly different between the phases it is interesting to note that the amplitude increases at different rates between the phases (i.e. the higher the intensity the greater the difference between phases) (Fig. 8). This may relate to our thought that the triceps brachii is active during flexion at lower intensities as a joint stabilizer and that as the intensity of arm cycling increases the triceps brachii is recruited to produce extension forces to a greater degree to assist with arm crank movement.

3.5 Methodological Considerations

As with most studies, there are some methodological considerations to keep in mind for future studies and when interpreting the results of the present study. One thing to consider with this study is the cadence during the maximal arm ergometry sprint versus the cadence during the cycling trials. During the maximal sprint, participants cycled as fast as they could for 10-seconds against a set resistance. During the cycling trials, however, participants cycled at a set cadence of 60 RPM for each workload. This difference could lead to a much higher level of muscle activity during the cycling sprint, due to the added influence of cadence-dependent changes in descending drive and/or afferent feedback. This is important because the cycling trials are normalized to this maximal sprint EMG, which partially explains the low level of EMG recorded from the muscles during the relative workloads, even at 50% PPO.

Another methodological consideration is the role of each limb during arm cycling. Though we know that there is likely a bilateral difference in force production during arm cycling (Carpes et al. 2010), we are unable to take this factor into account because we do not currently have force transducers positioned on the pedals. Also, the cadence is set at 60 RPM for each cycling trial, therefore the participant is cycling at 1 revolution per minute and it is may be that as power output increases, bilateral differences decrease. Thus, the dominant limb in the present study may have a disproportionate role during arm cycling at the lower power outputs. The dominant limb may have contributed to the cycling task more than the non-dominant limb, thus it is inaccurate to assume both limbs are equally performing the cycling task.

In this study the muscles of interest included the biceps brachii, triceps brachii, anterior deltoid, brachioradialis, flexor carpi radialis, and extensor carpi radialis in an attempt to get a better understanding of several of the joints involved in arm cycling (i.e. the shoulder and elbow). Obviously, however, there are more upper limb muscles involved in arm cycling than those assessed. In addition, muscles such as those in the torso are likely more active as the power output increases in order to stabilize the body. We are currently unable to specify how these considerations may have influenced the present results.

3.6 Conclusion

The main finding in the present study was that there is a linear relationship with iEMG and workload during arm cycling. As workload increased, there was an increase in motor unit recruitment and firing frequency, and thus, and an increase in iEMG. While the biceps brachii is phase and intensity dependent, the triceps brachii is intensity but not phase dependent. This suggests that triceps brachii remains active during both flexion and extension, and thus, has a role throughout the whole revolution of arm cycling. These findings warrant further investigation to determine underlying mechanisms for the triceps brachii during arm cycling.

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3.8 Figure Legends

Figure 1: Participants were seated with their shoulders at approximately the same height as the axis of the crank shaft in the SCIFIT cycle ergometer while cycling at 60 RPM at 11 different workloads (5-50% of PPO and 25W). Positions were made relative to a clock face. EMG was recorded from biceps brachii, triceps brachii, anterior deltoid, brachioradialis, FCR and ECR during the flexion and extension phases as well as the whole revolution. In this example, the participant is grasping the handle at the 6 o'clock position using the right hand.

Figure 2: Sample data collected during a non-maximal cycling trial. The signal from the 3 o'clock magnet is indicated by the straight lines, with the grey box representing 1 complete revolution of the crank. For clarity purposes the 9 o'clock magnet signal is not shown. Following data collection the timing of each 3 o'clock and 9 o'clock pulse were identified. These times were then used to divide the raw data signal into windows that went from 3 o'clock to 9 o'clock and 9 – 3. These time windows were then used to define the period over which iEMG was calculated.

Figure 3: Biceps brachii linear envelope ensemble averaged EMG during 1 full revolution for 25W, 10%, 20%, 30%, 40% and 50% PPO. Amplitudes are expressed as a percentage of maximal EMG. EMG output is the greatest during the flexion phase of the cycle (darker grey shade), and decreases to almost inactive during extension (light grey shade). During the lower intensities (25W-10%) the difference between phases is minimal. However as intensity increases, the difference between phase is more obvious.

Figure 4: Triceps brachii linear envelope ensemble averaged EMG during 1 full revolution for 25W, 10%, 20%, 30%, 40% and 50%. Amplitudes are expressed as a percentage of maximal EMG values. EMG is the greatest during the extension phase of the cycle (light grey shade), but is also very high during flexion (darker grey shade). During the lower intensities (25W-10%) the difference between phases is minimal. However as intensity increases, the difference between phases is more obvious, even though the triceps remains very active during both phases.

Figure 5: Group data (mean ± SE, n=11) of iEMG during flexion (black trace) and extension (grey trace) for the biceps brachii for all workloads (5%-50% and 25W). Amplitudes are expressed as a percentage of maximal EMG. iEMG output is position dependent for biceps brachii. Flexion had significantly higher iEMG compared to extension and as intensity increased; iEMG also significantly increased up to 35%; 35-50% did not result in significantly different EMG activation levels.

Figure 6: Group data (mean ± SE, n=11) of iEMG during flexion (black trace) and extension (grey trace) for the triceps brachii for all workloads (5%-50% and 25W). Amplitudes are expressed as a percentage of maximal EMG. There was no significant difference between phases as intensity increased. iEMG significantly increased up to 30%; 30-50% did not

result in significantly different EMG activation levels the effect of Intensity on iEMG tapers off in both flexion and extension at higher workloads.

Figure 7: Group data (mean, n=11) of iEMG activation levels during flexion (dark grey trace) and extension (light grey trace) for the biceps brachii for all workloads (5%-50% and 25W). Amplitudes are expressed as a percentage of maximal iEMG values. The slope was significantly different between flexion and extension for the biceps brachii (steeper during flexion than extension).

Figure 8: Group data (mean, n=11) of iEMG activation levels during flexion (dark grey trace) and extension (light grey trace) for the triceps brachii for all workloads (5%-50% and 25W). Amplitudes are expressed as a percentage of maximal iEMG values. The slope was not different between flexion and extension for the triceps brachii.

Figure 1: Experimental Set up

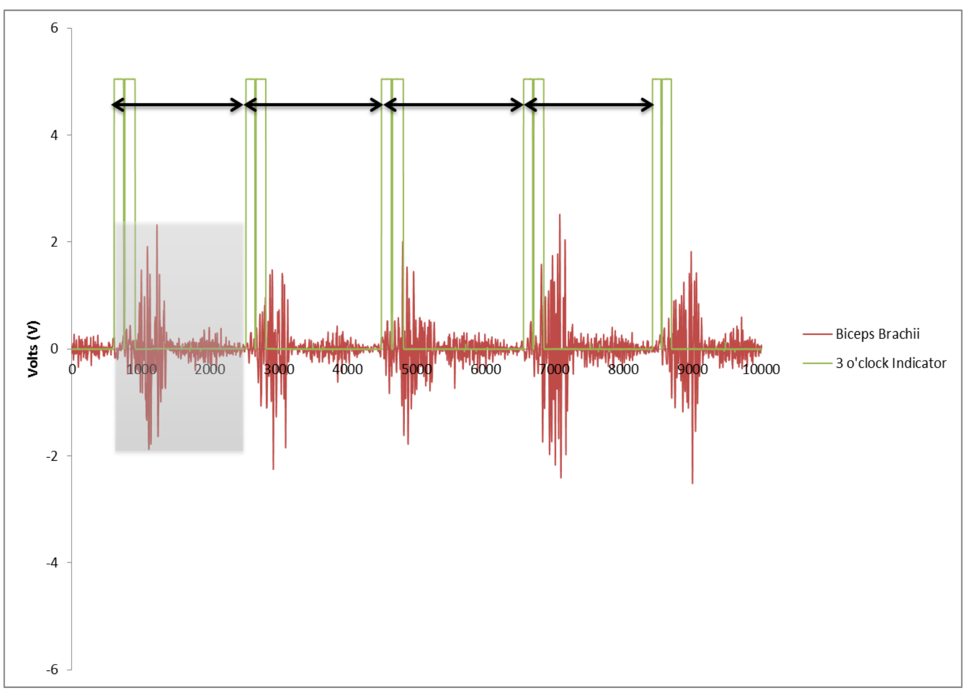

Figure 2: EMG Analysis

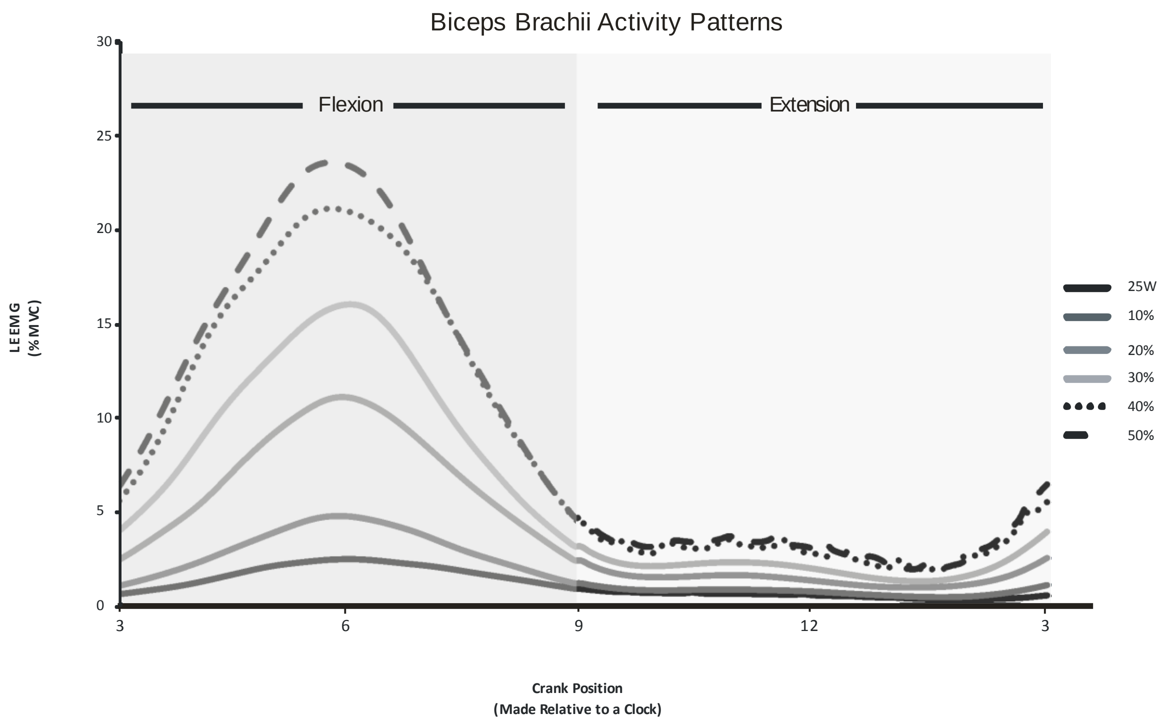

Figure 3: Biceps Brachii Linear Enveloped EMG, Ensemble Average

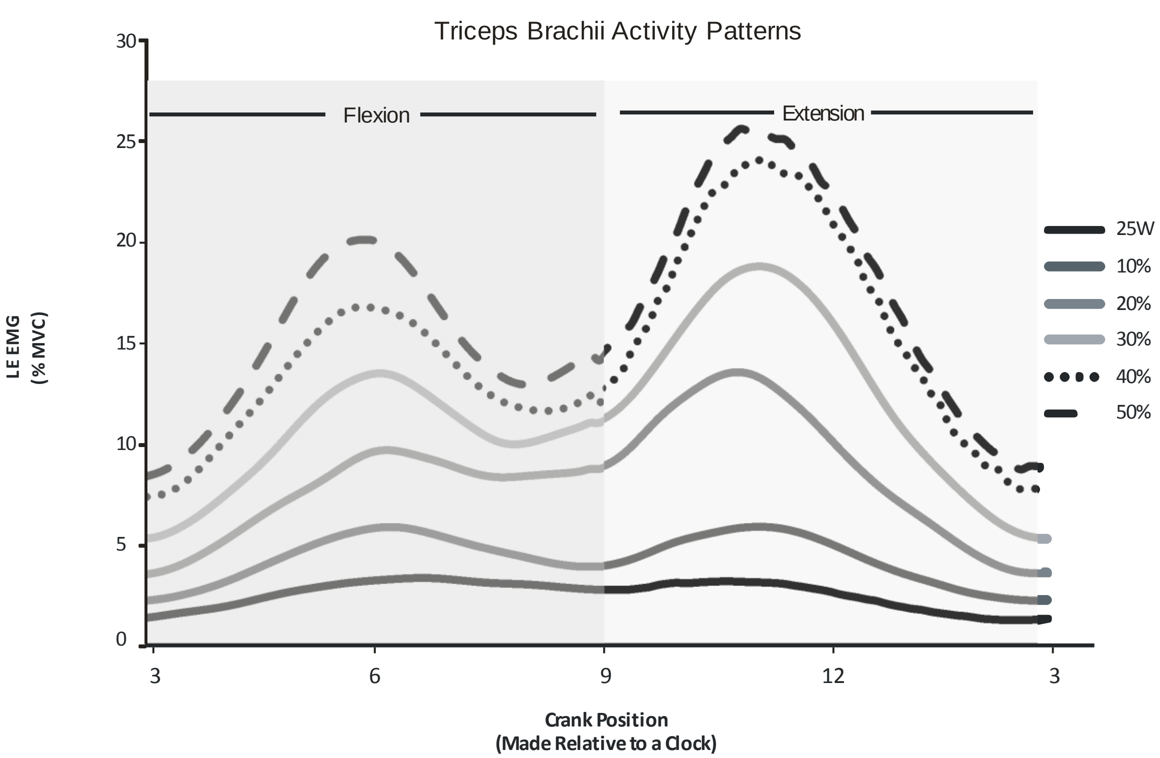

Figure 4: Triceps Brachii Linear Enveloped EMG, Ensemble Average

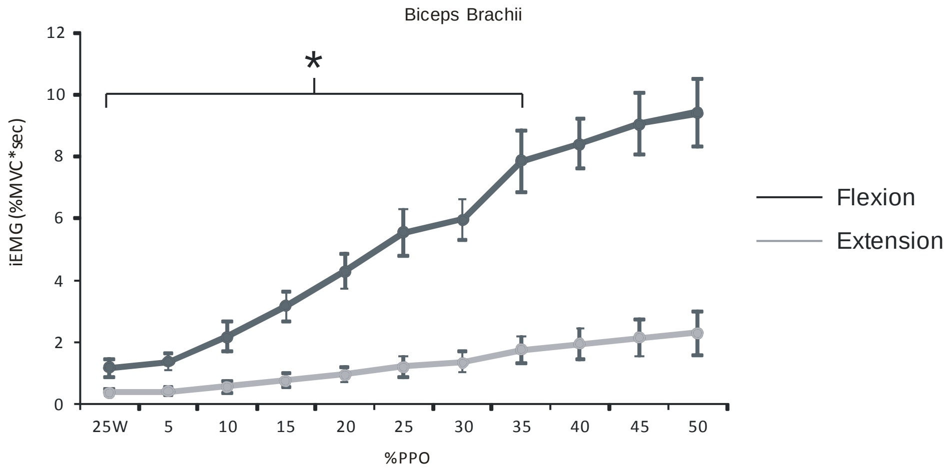

Figure 5: Biceps Brachii Group Data Flexion vs. Extension

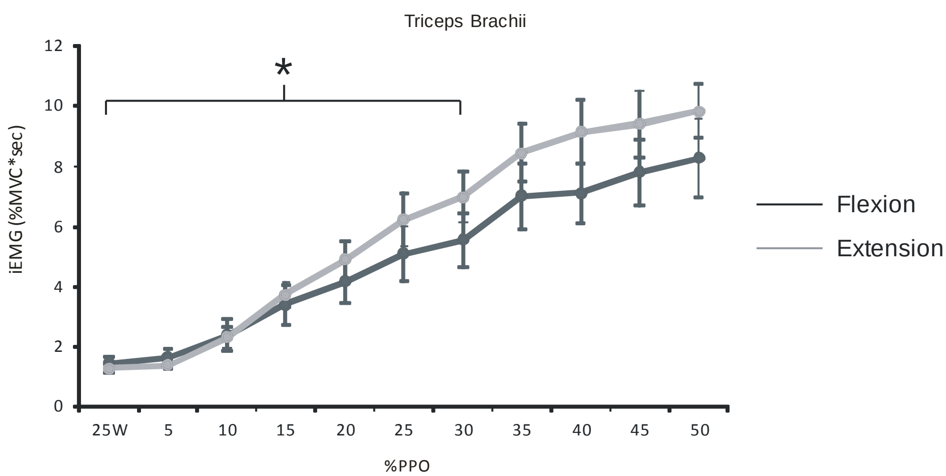

Figure 6: Triceps Brachii Group Data Flexion vs. Extension

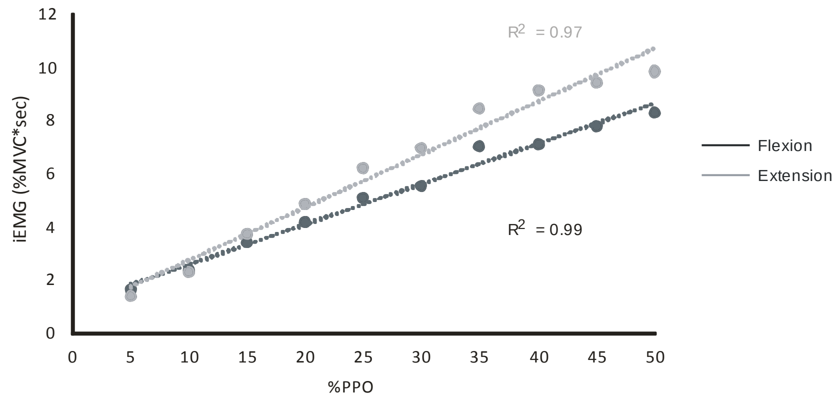

Figure 7: Biceps Brachii Slope

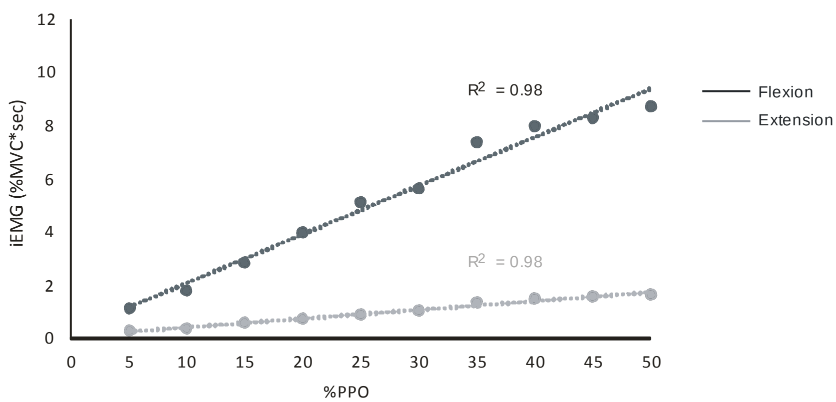

Figure 8: Triceps Brachii Slope

3.9 Table

Table 1. iEMG and workload summary table.

Muscle	Position Main Effect	Intensity Main Effect	Interaction Main Effect
Biceps Brachii	($F_{(1,10)}$ = 105.363, p < .001)	($F_{(2.72,27.22)}$ = 59.435, p < .001)	($F_{(2.98,29.18)}$ = 41.737, p < .001)
Triceps Brachii	*($F_{(1,10)}$ = 1.362, p = .270)*	($F_{(2.06,20.6)}$ = 65.015, < *.001*)	*($F_{(1.51,15.16)}$ = 2.246, p = .148)*
Anterior Deltoid	($F_{(1,4)}$ = 17.067, p = .014)	($F_{(10,40)}$ = 15.110, p = .001)	($F_{(10,40)}$ = 15.039, p = .001)
Brachioradialis	($F_{(1,4)}$ = 37.097, p = .004)	($F_{(10,40)}$ = 37.954, p < .001)	($F_{(10,40)}$ = 28.886, p < .001)
FCR	($F_{(1,4)}$ = 10.484, p = .032)	($F_{(10,40)}$ = 56.171, p < .001)	($F_{(10,40)}$ = 4.772, p = .040)
ECR	*($F_{(1,4)}$ = .217, p = .665)*	($F_{(10,40)}$ = 27.585, p < .001)	*($F_{(10,40)}$ = .509, p = .579)*

Table 2. Relationships between iEMG and workload.

Muscle	Phase	Linear	Quadratic	Cubic
Biceps Brachii	F	$102.766_{(1,10)}$, $p < .001$	$3.138_{(1,10)}$, $p = .107$	$4.669_{(1,10)}$, $p = .056$
Biceps Brachii	E	$83.347_{(1,10)}$, $p < .001$	$.541_{(1,10)}$, $p = .479$	$10.114_{(1,10)}$, $p = .010$*
Triceps Brachii	F	$41.255_{(1,10)}$, $p < .001$	$2.047_{(1,10)}$, $p = .183$	$1.257_{(1,10)}$, $p = .288$
Triceps Brachii	E	$85.090_{(1,10)}$, $p < .001$	$3.653_{(1,10)}$, $p = .085$	$3.764_{(1,10)}$, $p = .081$
Anterior Deltoid	F	$20.378_{(1,4)}$, $p < .05$	$.172_{(1,4)}$, $p = .700$	$3.681_{(1,4)}$, $p = .127$
Anterior Deltoid	E	$28.356_{(1,4)}$, $p < .01$	$.222_{(1,4)}$, $p = .662$	$12.342_{(1,4)}$, $p = .025$
Brachioradialis	F	$85.296_{(1,4)}$, $p = .001$	$7.819_{(1,4)}$, $p = .049$	$1.760_{(1,4)}$, $p = .255$
Brachioradialis	E	$40.878_{(1,4)}$, $p < .01$	$6.775_{(1,4)}$, $p = .060$	$23.252_{(1,4)}$, $p = .009$
FCR	F	$449.438_{(1,4)}$, $p < .001$	$5.663_{(1,4)}$, $p = .076$	$1.584_{(1,4)}$, $p = .277$
FCR	E	$25.641_{(1,4)}$, $p < .01$	$.000_{(1,4)}$, $p = .995$	$.573_{(1,4)}$, $p = .491$
ECR	F	$49.141_{(1,4)}$, $p < .01$	$2.233_{(1,4)}$, $p = .209$	$1.236_{(1,4)}$, $p = .329$
ECR	E	$19.137_{(1,4)}$, $p < .05$	$2.342_{(1,4)}$, $p = .201$	$1.654_{(1,4)}$, $p = .268$

Note: n=11 for biceps and triceps brachii and n = 5 for the remaining muscles. F = flexion phase and E = extension phase.

General Summary

This project was not intended to be a thesis, it began as a "for fun" project to conduct as an introduction into graduate school. The purpose of this study remained the same, even though there were some bumps along the way that slowed us down, deterred us, and really challenged us to think.

Collecting surface EMG from 12 muscles should have been the first sign that this would be a challenging project. During data collection we faced challenges with the EMG signal, the leads, and the amplifiers. After the data had been analyzed, it was apparent that something was not right with the results. After some investigation, it was determined that the amplifiers used to collect the data had been compromised, therefore the data was unusable.

After the data was recollected, and analyzed, our results showed some expected and unexpected findings. The biceps brachii results followed what has previously been found in the literature (rhythmic and alternating: active during flexion, inactive during extension). However, the triceps brachii results were not as expected. During arm cycling, the triceps brachii remains active during the whole revolution. This has potential implications on how the triceps brachii muscle may be researched in the future. This particular finding challenged us to think about what is really happening during a locomotive pattern that requires minimal thinking, such as arm cycling.

This project collected large amounts of data that have not been discussed. For example, the timing of muscle activation during arm cycling may be very interesting as a function of workload. Though it was not our purpose it could lead to future studies examining the onset and offset of each muscle, and more specifically when working together. Even though we collected data to compare isometric vs. dynamic contractions out of interest, it was not our main purpose to discuss the findings and again this could lead to future research to examine the more ideal MVC method when looking at dynamic tasks. We also did not focus on the ‘other’ muscles from which data was collected which included different muscles on the dominant limb as well as the non-dominant limb, and again this could lead to future research examining these various muscles during cycling and their synergist contribution to the task.

This project contributes to the literature and the overall understanding of the biceps brachii and triceps brachii muscles during arm cycling. It has created a foundation to build upon for our research team, and future projects have already started. These include assessing spinal and supraspinal excitability during arm cycling, characterizing EMG during arm cycling with different handgrips, synchronous vs asynchronous cycling, and also various inhibitory process that may or may not take place during arm cycling. Based on the results of this project, I would emphasize that the triceps brachii is as important and should be research as a main muscle of interest when examining the biceps brachii as well. This project, in conjunction with past, present and future studies from our laboratory, could one day influence the way spinal cord injury or other traumatic brain injury rehabilitation protocols are practiced.