Changes in Biceps and Triceps Brachii Electromyography During Wingate Anaerobic Tests at Different Intensities

By

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Abstract

Introduction The Wingate Anaerobic Test (WAnT) is a cycling protocol used to assess maximal power, with metrics like velocity, power, and muscle activity (EMG) helping to evaluate performance and motor activation. EMG can reveal differences in motor drive between limbs, including bilateral deficits. There is currently a gap in the research on the different activation strategies employed by the biceps and triceps brachii during a WAnT.

Objectives This study had two main objectives: 1) quantify the influence that arm dominance, unilateral or bilateral arm cycling, fatigue, WAnT resistance intensity, or crank position have on EMG amplitude of the biceps and triceps brachii; 2) assess whether there is evidence to support a bilateral EMG deficit during arm cycling.

Methods In this quasi-randomized study, 12 participants performed a series of 30s WAnTs with a resistance of 3, 4, and 5% of body mass (BM) unilaterally with their dominant and non-dominant, and bilaterally with both arms with 24-48h between each session (1 familiarization, 3 experimental). Surface EMG from the biceps and triceps brachii was recorded during each WAnT. Metrics for arm dominance, laterality, intensity, crank position, and fatigue were used to determine what had the greatest effect on the normalized mean EMG amplitude for the biceps and triceps brachii.

Results Crank position (6 o'clock/12 o'clock), ($p<0.001$, $p<0.001$), intensity (3%, 4%, 5%) ($p<0.001$, $p<0.001$), and arm (dominant/non dominant) ($p<0.001$, $p<0.001$) had a significant effect on EMG for the biceps and triceps brachii, respectively. WAnT section (beginning, middle, end) was only significant for the biceps brachii ($p=0.022$). There were significant intensity * arm ($p<0.001$, $p<0.001$), and crank position * arm ($p<0.001$, $p<0.001$) interactions, for biceps and triceps brachii, respectively, whereas intensity *

crank position (p=0.019) and crank position * WAnT section (p=<0.001) were significant for the biceps brachii.

Conclusions This study found that average muscle activity during a fatiguing arm cycling task varied with crank position, resistance, and whether the task was performed unilaterally or bilaterally. This variance of EMG activity for each of the aforementioned aspects, specifically when comparing between bilateral and unilateral trials suggests that there are different recruitment strategies used, depending on which arm is used for a given task. Future research could explore how these differences in individual motor unit recruitment, including bilateral deficits or facilitations, are influenced by factors like sex, training status, and task intensity, and assess their relationship to performance measures.

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List of Abbreviations

BLD – Bilateral Deficit

BLF – Bilateral Facilitation

BM – Body Mass

CNS – Central Nervous System

EMG – Electromyography

FI – Fatigue Index

fMRI – Functional Magnetic Resonance Imaging

H – Hours

HD-EMG – High-density Electromyography

iEMG – Integrated Electromyography

INaP – Persistent Inwards Sodium Current

ITT – Interpolated Twitch Technique

M1 – Primary Motor Cortex

MAV – Mean Absolute Value

MPO – Mean Power Output

mV – millivolts

MVDC – Maximal Voluntary Dynamic Contractions

MVIC – Maximal Voluntary Isometric Contraction

N – Newtons

nEMG – Needle Electrode Electromyography

Nm/s – Newton Metres per Second

PICs – Persistent Inwards Current

PM – Premotor Cortex

RMS – Root Mean Squared

PPO – Peak Power Output

RTD – Rate of Torque Development

S - Seconds

sEMG – Surface EMG

SIT- Sprint Interval Training

SLWAnT – Single Leg Wingate

SNR – Signal to Noise Ratio

Sub-MVIC – Submaximal Voluntary Isometric Contraction

VA – Voluntary Activation

W – Watts

W/kg – Watts per Kilogram

WAnT – Wingate

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Chapter 1: Literature Review & Introduction

1.1 Introduction

A Wingate anaerobic test (WAnT) is a common way of using either arm (Grant et al., 2014) or leg cycling (Green et al., 2001) as a means to assess a participant's maximal power (Grant et al., 2014; Green et al., 2001). A WAnT cycle ergometer set up consists of a set of pedals attached to crack shaft on a flywheel (Bar-Or, 1987). The cogs of the flywheel then drive a larger wheel via a chain (Bar-Or, 1987). Braking force is applied to the larger wheel either via an electromagnetic current or physical weight, adding resistance to the wheel (Bar-Or, 1987). A WAnT protocol typically consists of a short warm up period, allowing the participant to reach the desired cadence, followed by a percentage of the participants body mass being applied to the wheel as a load of resistance with the participant cycling at maximum effort for a set duration of time (Bar-Or, 1987; Silveira-Rodrigues et al., 2021). Often, a percentage of body mass applied as the load is a variable factor of a WAnT (Bar-Or, 1987; Patton et al., 1985; Silveira-Rodrigues). An additional parameter that can vary within WAnT protocols is the duration that the participant is asked to cycle at maximum effort (Bar-Or, 1987), though 30s is the most common (Bar-Or, 1987; Castañeda-Babarro, 2021).

Some common metrics obtained from WAnT are velocity (Parisi & Allen, 1994), power (Patton et al., 1985; Zupan et al., 2009), force development (Obmiński et al., 2015; Parten et al., 2023), fatigue index (Dupont et al., 2007; Maud & Shultz, 1989) and at times, electromyography (EMG) (Chtourou et al., 2011; Souissi et al., 2012). Velocity, or cadence is almost always built into the WAnT cycle ergometer and its software (Parisi & Allen, 1994). While EMG is not included in all WAnT protocols, WAnTs are highly compatible

with EMG (Okano et al., 2017). The main reason for this is EMG allows for a consistent measure of motor activation during the trial (Rana, 2006), and its reliability is further enhanced when the participant is secured in such a way to limit excess movement (Samuel et al., 2017).

Depending on the research question, EMG can be a very important metric to include when performing a WAnT. Specifically, it provides information on nervous system output (Farina et al., 2004). EMG values are a representation of the average motor neuron activation over a given area within a given sample window (Suzuki et al., 2002) which allows researchers to estimate neural drive (Farina et al., 2010; Karimimehr et al., 2017). Activation can differ across limbs and muscles, depending on the task, if the said task is performed unilaterally with the dominant or non-dominant limb, or if it is performed bilaterally (Niu et al., 2011; Saeterbakken et al., 2015). The use of EMG during a task such as a WAnT allows the researcher to quantify the difference in activation or drive and contrast it with other recorded metrics (Farina et al., 2004).

Lastly, in order for the EMG data obtained to be valid, along with any claims made from the said data, it has to be normalized (Singh & Singh, 2020). Normalization is required because each participants' individual outputs will be different (Galán-Rioja et al., 2020; Naughton et al., 1992) and can vary depending on the day (Pedersen et al., 2016; Souissi et al., 2012). So, in order to make generalizations about a population, a controlled average must be made (Kern et al., 2016). A specific benefit to using WAnTs for studies such as this, is that all data received is automatically partially normalized, since applied force is relative to the body mass (BM) of each participant (3%, 4%, or 5% of BM) reducing the amount of work the researcher has to do for each trial, since every participant has a load

automatically tailored to them, rather than using a series of static loads that would not change from participant to participant..

1.2 Wingate performance of the legs and the arms

General performance metrics of a WAnT include peak and mean power, measured in watts (W) (Carlson & Naughton, 1994), rate of torque development, measured in Newton meters per second (Nm/s) (Morris et al., 2010), fatigue index (FI), expressed as a percentage (Lunde & O'kroy, 2016), and RPM, expressed as the number of rotations the cycle ergometer wheel performs within a minute (Hager et al., 2011). When comparing absolute power outputs, not normalized for body mass, males demonstrate greater power outputs than females (Thompson et al., 2015). Specifically, in the Thompson et al. (2015) study that took sex differences into account, they found that the peak power output (PPO) was approximately 55.1% greater in males, than in females (849.21 ± 127.41W vs 547.74 ± 94.56W). When normalizing for weight, male and female participants have similar peak power outputs 50.4 ± 5.6 W/kg vs 50.5 ± 6.2 W/kg, respectively (Pérez-Gómez et al., 2008). However, when comparing the relative mean power outputs, men (26.6 ± 3.4 W/kg) do have higher results than women (21.9 ± 3.2 W/kg) (Pérez-Gómez et al., 2008).

Aside from sex differences in WAnT performance, there is also a difference in WAnT performance between trained athletes and untrained individuals (Arslan, 2005; Delahunt et al., 2013). When specifically comparing untrained, sedentary individuals that were trained over a 6-week period, performance increased by approximately 18.3% for PPO (970 ± 176 W (pre training) *vs* 1148 ± 159 W (post training)), and approximately 10.0% for mean power output (MPO) (399 ± 55 W *vs* 439 ± 53 W) (Delahunt et al., 2013).

This indicates that a more trained person, such as an athlete, will be able to output more power during a WAnT than an untrained individual.

The last major performance difference that needs to be made is between WAnT arm cycling and leg cycling. Expectedly, PPO and MPO are greater in leg cycling than in arm cycling (Weber et al., 2006). Weber et al. (2006) found that the power output for arm cycling was less than it was for leg cycling. Specifically, PPO was, on average, was equal 70% of a participant's leg cycling PPO, whereas their MPO for arm cycling was only 59% of their leg cycling MPO (Weber et al., 2006). This difference is PPO and MPO can largely be attributed to the increased muscle mass of the legs when compared to the arms (Hosler et al., 1982; Lynch et al., 1999).

All the aforementioned metrics do change as resistance is added to the WAnT. As resistance is increased, peak power will increase up to a certain point, after which, peak power will decrease (Carlson & Naughton, 1994; Patton et al., 1985). This suggests that there is an optimal resistance level for a given WAnT (Jaafar et al., 2014). Rate of torque development shows a clear decrease as resistance is increased (Morris et al., 2010). Additionally, rate of torque development can also act as an indicator of fatigue for a participant, where the more the rate of torque development decreases, the more a participant is becoming fatigued (Morris et al., 2010). Fatigue index (FI) is the other primary measure of fatigue during a WAnT (Lunde & O'kroy, 2016) and it will increase as the resistance increases (Lunde & O'kroy, 2016). Another metric that can be used to measure fatigue is RPM of the cycle ergometer wheel, which decreases as the resistance is increased during a WAnT and as a participant fatigues, even when the applied load remains constant (Hager et al., 2011).

Regarding a leg cycling WAnT, the results are affected by serval factors. One of the factors that can affect the performance of a leg cycling WAnT is whether the test is performed unilaterally or bilaterally (Hebsertreit et al., 1999). When a single leg Wingate (SLWAnT) is performed, the total amount for power and torque is higher in the bilateral WAnT, when compared to either leg performing a SLWAnT (Dunstheimer et al., 1999; Hebsertreit et al., 1999). However, when summing both legs performing SLWAnTs together and contrasting them with the bilateral WAnT, the sum of both is greater than the bilateral WAnT (Dunstheimer et al., 1999). Another factor that affects the performance of a leg cycling WAnT is exercising prior to the WAnT (Grant et al., 2014). Interestingly, the exercise does not have to be localized to the legs, to see a reduction in leg cycling performance (Grant et al., 2014). In the study performed by Grant et al. (2014), they were able to demonstrate that fatiguing arm exercise prior to a WAnT was able to meaningfully decrease the performance of the subsequent WAnT, even though the legs remained unfatigued prior to the WAnT.

A pharmacologically way to increase leg cycling WAnT performance is to increase a participant's intake of inorganic nitrates (Jodra et al., 2020). In the study performed by Jorda et al. (2020), they were able to demonstrate that increasing intake of inorganic nitrates led to an increase in peak power during a WAnT and lowered the time to reach peak power. Additionally, they demonstrated that the treatment led to a decrease in perceived muscle exertion, post-exercise (Jorda et al., 2020). Physiologically, the performance of a leg cycling WAnT can be influenced by prior training (Hager et al., 2011; Wun et al., 2020; Zinner et al., 2016). One method of training that can affect the performance of a leg cycling WAnT is previous experience with a WAnT (Wun et al.,

2020). Over time and practice with a leg cycling WAnT, an increase in peak aerobic power can be observed, which leads to improved WAnT performance (Wun et al., 2020). High-cycle resistance training and leg press training also increase leg power output and subsequently increase WAnT performance (Hager et al., 2011). Similarly, sprint interval training (SIT) also increases the aerobic performance of the legs which, predictably, leads to an increase in leg cycling WAnT performance (Zinner et al., 2016).

Arm cycling WAnT performance is similarly affected by a multitude of factors (Grant et al., 2014; Lovell et al., 2011; Zinner et al., 2016). Similar to leg cycling WAnTs, arm cycling WAnTs can also be affected by SIT. The implementation of SIT leads to an increase in performance in arm cycling WAnT similar to leg cycling WAnTs (Zinner et al., 2016). Additionally, asynchronous vs synchronous cycling affects the performance of an arm cycling WAnT (Lovelle et al., 2011). More specifically, having the cycling crank set up in an asynchronous manner (where one pedal is 180° to the other) results in higher WAnT performance when compared to positioning them in a synchronous (where one pedal is 0° to the other) manner (Lovell al., 2011).

1.2.1 Why use arm cycling as opposed to leg cycling for a WAnT

The current body of research supports the idea that CPGs can produce rhythmic motion in both the arms and legs, seen during locomotion and cycling (Calancie et al., 1994; Power et al., 2018; Zehr et a., 2004). Evidence supporting the notion that locomotor patterns are housed in the spinal cord, and not the brain can be seen with involuntary stepping movements overserved in a patient with a complete spinal cord injury (Calancie et al., 1994). Arm cycling has also been directly used to assess reflex modulation in human participants, with convincing evidence suggesting that CPGs are at least partially at play

(Zehr et al., 2004). Though both leg and arm cycling are two valid models of rhythmic output, arm cycling does have the distinct advantage of high head and arm stability since it allows for the use of a back rest to support and strap the trunk in place as well as to stabilize and strap the head of the participant into position (Power et al., 2018). When a participant is secured to the cycle ergometer in such a way, it removes balance from being another variable (Power et al., 2018). Stimulation techniques to assess cortical spinal excitability, such as CMEPs [] or M-wave stimulation are also more convenient protocols using the arms, since the stimulation is required to travel a shorter distance(Power et al., 2018).

Additionally, there are a few circumstances where it may be advantageous to use arm cycling over leg cycling. Once of such situations is accessibility. Arm cycling is generally more accessible than leg cycling, allowing participants who are wheelchair users (assuming they still have use of their arms) to be included in such studies or training regiments (Hill et al., 2019; Krops et al., 2017). Similarly, an arm cycling WAnT would be a much better choice in a rehabilitation setting where the participant has a disability affecting their lower extremities, such as a spinal cord injury, cerebral palsy, or lower limb amputation (Krops et al., 2017). Lastly, an arm cycling WAnT is the ideal choice when the research question or training regimen is focused on the upper limbs, such as training a sport such as water polo (Vrdoljak et al., 2022), or if the research is predominantly focused on the biceps or triceps brachii (Forman et al., 2019; Nippard et al., 2018; Nippard et al., 2019).

1.2.2 Neuromuscular Fatigue During a WAnT

The WAnT is a highly fatiguing task, which not only fatigues the muscle(s) involved, but also fatigues the nervous system output to the muscles (Hunter et al., 2003; Pearcey et al., 2015). Neuromuscular fatigue can be described as a combination of peripheral and central fatigue (Collins et al., 2018). Peripheral fatigue is shown as a non-linear decrease in power or torque output over the course of the WAnT (Hunter et al., 2003; Pearcey et al., 2016). Looking at the central portion of neuromuscular fatigue, during a WAnT, within the biceps brachii specifically, supraspinal excitability decreased, whereas spinal excitability increased as the participants fatigued (Pearcey et al., 2016).

The primary methods employed to assess central fatigue are the use of the Interpolated Twitch Technique (ITT) either via transcranial magnetic stimulation (TMS), or via peripheral nerve stimulation (Collins et al., 2018). ITT is necessary to estimate the voluntary activation (VA) of the muscle and whether the changes in VA are caused by supraspinal output, central drive proximal to the motor neuron axons, or a combination of the two (Collins et al., 2018). When considering sprints, such as the WAnT, there is an observable peripheral fatigue that develops early and persists throughout the task and throughout repeated tasks (Collins et al., 2018). When looking at repeated sprit trials, the neuromuscular fatigue also varies in origin depending on if the assessment is taking place early or late in the repeated sprint trials (Collins et al., 2018). Within the beginning half of the repeated sprint trials neuromuscular fatigue was peripheral in nature, whereas towards the end of the trials, neuromuscular fatigue is both peripheral and central in nature (Collins et al., 2018).

[add in section about peripheral changes in muscle activity with fatigue (seen with EMG)]

1.3 What is a Bilateral Deficit?

The bilateral deficit (BLD) is a decrease in the observed contraction metric (often force) when comparing the sum of unilateral measurements to the sum of the bilateral measurements (Kuruganti et al., 2007; Ruiz-Cárdenas et al., 2014; Škarabot et al., 2016). It is often expressed as the bilateral index (BI), using the following equation BI = (100 * ∑ of the average bilateral force / ∑ of unilateral force) – 100 (Fountaine, 2018). The exact mechanism behind BLD is currently unknown, however, there are some leading ideas on what contributes to it. One of the proposed ideas contributing to BLD is higher-order neural inhibition (Škarabot et al., 2016). This idea suggests that the cortex limits the amount that a muscle can activate during a bilateral contraction (Post et al., 2007). More specifically, current research indicates that the command to reduce motor output during bilateral contractions may come from the M1 motor cortex (Post et al., 2007). Another proposed factor to BLD is variation in shortening velocity and displacement of the force-velocity curve during different contraction modes (Škarabot et al., 2016). Though, this explanation predominantly applies to the BLD observed during ballistic type movements (Škarabot et al., 2016). Regardless of the contributing factors to BLD, it is well preserved being observable across upper and lower limbs (Janzen et al., 2006; Taniguchi, 1998) and during dynamic and isometric contractions (Jakobi & Chilibeckl, 2001; Škarabot et al., 2016). With that in mind, BLD is more pronounced in step contractions when compared to ramp contractions (Koh et al., 1993), and dynamic contractions when compared to isometric contractions (Škarabot et al., 2016).

BLD can also be increased or decreased depending on training and exercise. For example, bilateral resistance training tends to decrease BLD, whereas unilateral resistance

training tends to increase BLD (Ruiz-Cárdenas et al., 2014). Additionally, rapid contractions generally produce a larger BLD when compared to slow contractions (Owings & Grabiner, 1998). BLD is also suggested to be affected by age, with greater BLDs observed in older participants, when compared to younger participants (Owings & Grabiner, 1998), with a potential explanation being the atrophy of high-threshold motor units as people age (Beurskens et al., 2015; Owings & Grabiner, 1998).

Traditionally, BLD is measured with force output (in Newtons, N) (Kuruganti et al., 2007; Yamauchi et al., 2009), rate of torque development (RTD) (in Newton meters per second, Nm/s) (Šarabon et al., 2020), or power (in watts, W) (Nakachi et al., 2019) of the limbs in question. However, BLD can also be quantified though the use of EMG (Carr et al., 2020; Dieën et al., 2003; Post et al., 2007). When using EMG as the method of comparing BLD, the metric compared is the depolarization of the target muscle, measured as voltage across the skin (in millivolts, mV) (Post et al., 2007). To get usable data, the data must first be rectified (all negative values are converted to positive values) and normalized, at the minimum (Ward et al., 2013). However, when using EMG to assess BLD, it is common to implement more signal processing techniques, such as Carr et al. (2020) did when they used a fourth-order Butterworth band-pass filter between 20 and 450 Hz and smoothed the data with a 50ms zero-shift root mean squared (RMS). The primary advantage to using EMG to assess bilateral deficit, is that since EMG is interpreted as a measurement of activity sent from the CNS (Brown et al., 1999), and can be used to give insight on the amount of cortical drive that is being expressed within a contraction (Griffin et al., 2008; Myers & O'Malley, 2003). EMG is not a flawless tool in interpreting activity from the CNS (Chowdhury et al., 2013; Türker, 1993). Specifically, EMG signal is

suceppitble to external noise from power sources (Leske & Dalal, 2019), motion artifacts (Oo & Phukpattaranont, 2022), crosstalk contamination, (Germer et al., 2021), and other physiological noise, such as electrocardiography artifacts (Miljković et al., 2017), all of which reduce the signal to noise ratio (SNR) (Daley & Kuna, (2009).

1.4 Movement - From the Central Nervous System to the Muscles

The general flow of information within the circuits is as follows; upper motor neurons of the descending systems (the motor cortex and brain stem) integrate information from the basal ganglia and the cerebellum (Purves et al., 2001). Information then travels to the local circuit neurons and motor neuron pools within the spinal cord and brain stem. Information from the local circuit neurons is then sent to the motor neuron pool as well (Purves et al., 2001). All the information from the motor pool then leads to a change in the skeletal muscle, either contraction or relaxation (Purves et al., 2001).

Once the signal has reached the target muscle, motor units must be recruited in order for that muscle to generate force (Petajan, 1991). Within motor recruitment, there is an order that is typically followed. Motor units are recruited in order of their size, with the smaller units (slow, type I) being recruited first, with the size of the recruited motor units increasing as the contraction persists (Clamann, 1993; Gordon et al., 2004). Similarly, motor units are de-recruited in the reverse order (Gordon et al., 2004). This phenomenon is called the Henneman Size Principle (Vilensky & Gilman 1998). Specific support for the Henneman Size Principle has been shown with the use of EMG, demonstrating that amplitude, duration, and the area of the motor unit action potentials increase with recruitment order (Akaboshi et al., 2000; Ertaş et al., 1995). Additionally, H-reflex studies

have also added support for Henneman's size principle, demonstrating that smaller motor units are recruited first in both voluntary and reflexive movements (Zhu et al., 2018).

While the Henneman Size Principle often holds true, it is not an absolute. Recruitment patterns can vary depending on the activity and fatigue level of the participant (Hodson-Tole & Wakeling (2008). An example of when the Henneman size principle may not be observed is during rapid, explosive type movements (Sale, 1987). During explosive movements, there is a need to develop high amounts of force rapidly (Sale, 1987), and as such, there is an immediate demand for the larger (fast, type II) motor units to be active immediately (Dideriksen et al., 2019; Sale, 1987). Fatigue also can impact motor unit recruitment by decreasing the threshold at which higher threshold motor units are recruited (Adam, & Luca, 2003; Mcmanus et al., 2015). Additionally, fatigue can increase the de-recruitment threshold of active motor units (Enoka et al., 1989; Stock & Mota, 2017).

It is also worthwhile noting that EMG can be used to determine global output of the nervous system to the muscles (Farina et al., 2010; Farina et al., 2014). This is primarily through estimation of neural drive (Farina et al., 2010). Surface EMG allows for the summation of the action potentials discharged by the motor units, which in turn were sent that information to discharge from higher order portions of the CNS (Farina et al., 2004; Farina et al., 2010). Generally speaking, the greater the amplitude of the EMG, the motor units within the targeted muscle are firing at a greater frequency and or in great numbers, which allows the inference of greater neural drive (Farina et al., 2004; Farina et al 2010; Szyszka-Sommerfeld et al., 2020).

1.5 Electromyography and How it is Analyzed

Electromyography (EMG) is a common diagnostic tool used in neurophysiology (Stålberg et al., 2019), and kinesiology research (Clarys & Cabri, 1993). EMG is most often used when a researcher is interested in looking at the neural drive to the muscle in question (Hug et al., 2015). EMG can gather this information by using one or more electrodes to derive muscle activity from the surface of the skin (surface electrode, sEMG), or within the muscle (needle electrode, nEMG) in reference to a ground electrode (Mayo Foundation for Medical Education and Research, 2019). The result is a reading measured in millivolts (Khanam et al., 2015) and serves as a method to interpret neuromuscular activity (Raez et al., 2006). In fact, EMG readings can act to infer neuromuscular drive from the CNS (Dideriksen et al., 2011; Farina, 2006), with EMG amplitude increasing as a participant's neuromuscular drive increases (Dideriksen et al., 2011).

While the general underpinnings of remain constant, there are two main ways to acquire this data, either via surface recording or intramuscular recording (Gohel & Mehendale, 2020). Should a researcher wish to record from a single muscle fibre, then a needle electrode is often the best choice, as when inserted into the muscle, it will only record from that singular muscle fibre (Gohel & Mehendale, 2020). The major downside to this process is that it is invasive, as well as it requires the assistance of a physician to surgically implant the electrode, increasing cost, and complexity to the trial (Yamashita et al., 2022). Additionally, since, needle electrodes only record from one muscle fiber at a time, it is not the best option to use when the research question necessitates looking at either a large number of muscle fibres, a group of muscles for a given action (Papathanasiou & Zamba-Papanicolaou, 2013; Tanhehco et al., 2003), or if the particular action would

exert a high amount of strain on the needle electrode, leading to a fatigue fracture of the electrode within the participant (Wang et al., 2023). The far more common option is to use surface electrodes for EMG signal acquisition (Gohel & Mehendale, 2020) which can either be applied as a high-density array or a series of single electrodes placed over the target muscle belly (Gohel & Mehendale, 2020). Surface electrodes have the distinct advantages of being noninvasive and being able to record from a multitude of muscle fibres all at once (Gohel & Mehendale, 2020). There is a further variant of surface electrodes, known as high-density surface electrodes. These are used in high-density electromyography (HD-EMG) and consist of an array of small electrodes which are placed onto the target muscle area (Zhao et al., 2023). HD-EMG has the advantage of being able to be decomposed to look at individual motor unit recruitment (Clarke et al., 2020; Glasser et al., 2013). Irrespective of which electrode type is chosen, the EMG signal must be rectified and processed in order make use of the acquired data (Raez et al., 2006). Signal processing is often conducted behind the scenes by the software in which the signal was recorded. There are multiple ways in which the EMG recording can be processed, such as wavelet analysis, time frequency approach, artificial intelligence modelling, higher order statistics, fuzzy logic, Choi-Williams method, and Wigner-Ville distribution (Raez et al., 2006).

Similar to other bio-signaling techniques, a number of factors can affect the quality of EMG readings. One such example being crosstalk, where the EMG signal in a non-active muscle is contaminated by the myoelectric signal of a nearby active muscle (Farina et al., 2006). Additionally, the other more common ways that EMG data can be contaminated is with though motion artifacts and power line noise (Boyer et al., 2023). A

motion artifact is seen when there is a sharp, brief spike in activity much higher than what the muscle is capable of producing, often caused by changes in the skin-electrode impedance level, or from disturbing the elected cables (Boyer et al., 2023). Power line interference, on the other hand is the noise picked up by the electrode from the electrical grid itself (Boyer et al., 2023), which is often filtered out to improve the SNR of the recording (Boyer et al., 2023).

When acquiring EMG signal, it must be processed. This typically involves rectifying the waveforms, using a filter to remove background noise, such as a Butterworth low-pass filter, and then normalizing the values to a set reference point, often to the peak value, either dynamic or isometric (Alizedah et al., 2023; Daley, & Kuna, 2009). When acquired, the EMG signal for rhythmic motion should appear wave-like, which allows for separate calculations of muscle activity and allows each cycle to be broken down into phases (Zehr et al., 2007)

Once EMG data has been acquired, there are multiple ways to interpret it. One common way to do this is to use the mean EMG amplitude. Mean EMG amplitude is used to assess the average muscle activation of the muscle in question during a contraction and is directly related to the number of motor units firing and the rate at which they are firing (increasing either will increase the mean EMG amplitude) (Renshaw et al., 2010; Suzuki et al., 2002). It is worth noting that calculating EMG amplitude with the mean absolute value (MAV), while the calculation is performed over a given duration of time, the resulting function only reports in voltage, not voltage and time. This may be noteworthy, depending on how MAV is applied. Even still, the use of MAV does produce a robust SNR (Clancy & Hogan, 1999). An alternative way to analyze EMG amplitude is by taking its

RMS (Clancy & Hogan, 1997; Clancy & Hogan, 1999; Wang et al., 2019). A particular advantage RMS has over MAV, is that it does take the time domain of the EMG signal into account (Farfán et al., 2010). However, it does have the disadvantage of having a lower SNR ratio, when compared to MAV (Clancy & Hogan, 1999; Phinyomark et al., 2013). Integrated electromyography (iEMG) is very similar to the RMS approach; however, it differs by utilizing the rectified amplitude values and sums them over time, rather than measuring the square root of the average over time (Arabadzhiev et al., 2009; Mushtaq & Chawla, 2016). Regardless of which method is used to interpret the data, the raw EMG values should be normalized in order to provide an accurate way of comparing between participants, days, or muscles, depending on the study, prior to performing any statistical analyses on the data (Burden, 2020; Chalard et al., 2020; Yang & Winter, 1983). Additionally, while mean values are often used for EMG because they more accurately provide insight into what the muscle activity is over time. However, there are other ways to average the initial EMG values. A Fast Fourier transformation can be used on raw data to look at mean and median power (Alizedah et al., 2021). The use of FFT transforms the EMG data from the time domain, into the frequency domain, which allows the frequency components of the signal to be analyzed (Zawawi, 2015) Analyzing EMG in this fashion is preferrable when signal conduction velocity is of interest and is better able to show fatigue (Alizedah et al., 2021).

1.5.1 Fatigue Development and Electromyography

Depending on the muscle of interest, and how the fatiguing task is performed, and the position of the limb, EMG will either increase (Alizadeh et al. 2024; Chaytor et al., 2020; Dimitrova and Dimitrov, 2003), decrease (González-Izal et al., 2012), or remain the

same (Pearcey et al., 2016). When a submaximal fatiguing task is employed, EMG amplitude is more likely to increase (Alizadeh et al. 2024), whereas when a maximal fatiguing task is employed, surface EMG is more likely to decrease over time (Alizadeh et al. 2024; González-Izal et al., 2012).

The method of assessing fatigue and collecting EMG can affect the delta EMG as well. For example, when EMG is collected and assessed during the particular fatiguing trial, a very evident change in EMG can be observed (Alizadeh et al. 2024). However, if a study was to perform a fatiguing task, and then measure EMG with another task, such as an isometric flexion task, a change in EMG might not be visible (Pearcy et al., 2016). This can be directly applied to bilateral WAnT studies. During a WAnT, muscle activity and EMG data are subject to change (Alizadeh et al., 2024; Robergs et al., 2015) due to various muscle contributions to the task/movement. Alizadeh et al. (2024) were able to demonstrate that EMG amplitudes decreased or increased as the WAnT progressed, depending on which muscle was being assessed. For example, the EMG decreased as the WAnT progressed, when looking at the biceps brachii, however EMG increased as the WAnT progressed when the muscle in question was the latissimus dorsi (Alizadeh et al. 2024). This indicates that EMG activity during a WAnT is variable, depending on the muscle observed.

1.6 Data Normalization for Electromyography

Normalizing data is the process of transforming raw, experimental data based on a set of reference data with the goal of improving the quality of the data (Singh & Singh, 2020). The primary benefit to normalizing data is that it allows the researcher to make comparisons across participants and between muscles within an idividual over the course of a study (Cotton-Barratt et al., 2020; Knutson et al., 1994). Critically, within the scope

of EMG research, normalizing data allows for some measure against the inherent variability of EMG signals (Lehman & McGill, 1999; Tabard-Fougère et al., 2018). When it comes to obtaining good reference data, the performed task should be highly consistent and repeatable (Halaki & Gi, 2012).

There are a few different techniques employed when normalizing EMG data. The most common of these is taking the raw EMG amplitude measurement and expressing it as a percentage of the participants peak maximal voluntary isometric contraction (MVIC) (Knutson et al., 1994). An isometric contraction is a type of contraction where the participant exerts a force with the target muscle, however the limb is held firmly in place, so that there is no movement (Dunleavy, 2019). Using MVICs are often the simplest way to normalize EMG data because it only requires the perform series of MVICs immediately prior to performing the experimental task, and a simple calculation to generate the normalized percentage value. While normalizing dynamic contractions to static contractions is commonplace, and does provide a baseline to compare the experimental data against (Chalard et al., 2020; Hunter et al., 2002; Rota et al., 2013), it is technically comparing two different types of movement; static and dynamic. Additionally, peak MVIC occurs at a single instance in time, so it does not provide insight into the MVIC EMG across the contraction and cannot be used to normalize EMG area under the curve.

Keeping with isometric normalization, another way to normalize EMG data is to use a submaximal voluntary isometric contraction (sub-MVIC) (Yang & Winter, 1983). Just as the name implies, the participant is asked to perform a series of isometric contractions below what their maximal output would be (Yang & Winter, 1983). The main potential advantage that sub-MVICs have over MVICs is that sub-MVICs are less variable

over different days (Dankaerts et al., 2004; Yang & Winter, 1983), and across individuals (Biviá-Roig et al., 2019). While using sub-MVICs does certainly have its advantages over MVICs, it does also have a substantial drawback when performing fatiguing trials (Fuglevand et al., 1993). Specifically, a sustained submaximal contraction will impair neuromuscular propagation (Fuglevand et al., 1993). Additionally, muscle activation patterns can change over repeated submaximal isometric contraction sessions, which can add variability into the reference data (Hunter & Enoka, 2003).

While using peak isometric EMG to normalize a dynamic movement is common, it is not the only way to normalize data from a dynamic movement. Maximal isokinetic contractions can also be implemented as a source of reference data. An isokinetic contraction is a type of contraction where the movement of the contraction about a joint is maintained at a constant velocity (Baltzopoulos & Brodie, 1989). The potential advantage that maximal isokinetic contractions have over MVICs, is that they are less likely to overestimate the percentage MVC, and adding a time component to the reference data, making them a viable option for normalizing EMG during dynamic tasks (Calver et al., 2023). Like all normalization methods, maximal voluntary isokinetic contractions do have their drawbacks when selected to obtain reference data. The primary disadvantage to them is that they often show increased variability between and within participants (Burden et al., 2003; Chalard et al., 2020).

Like maximal voluntary isokinetic contractions, a maximal voluntary dynamic contraction (MVDC) (Hodder & Keir, 2013; Warnock et al., 2019) can also be used to normalize EMG data. This procedure is like the MVIC, with the participant performing a movement with a fixed load at a fixed pace to ensure the target muscle is activated

maximally (Wang et al., 2023). How it differs from an maximal voluntary isokinetic contraction is that a MVDC does not necessarily have to move about one joint, and can at times be a separate task to what the experimental task is, such as chin up being used to normalize upper limb EMG data, so long as the target muscles are still active (Rota et al., 2013).When an MVDC is the normalization task of choice, its EMG data contains mean and peak EMG amplitude, and similar to maximal voluntary isokinetic contraction, it also contains a time component which can be used to normalized data slightly differently vs using isometric data (Rozand et al., 2017). Similarly to using maximal voluntary isokinetic contractions for normalizing EMG, MVDC data also has higher variability when used as reference data, as compared to MVIC data (Burden & Bartlett, 1999; Chuang & Acker, 2019).

1.7 Conclusion

The notion that EMG values are influenced by several factors are well established (Neptune & Herzog, 2000; Niu et al., 2011; Saeterbakken et al., 2015; Souissi et al., 2012). Similarly, EMG has been used with great success during WAnTs (Chtourou et al., 2011; Souissi et al., 2012) and BLDs have been observed in the upper body (Taniguchi, 1998; Ye et al., 2019). However, to this date, only one study has investigated whether a BLD is observable during an upper body WAnT (Antolinez et al., 2024), though they did not assess EMG during this study. Currently, there is a gap in knowledge on which factors have the greatest effect on EMG amplitude during an upper body WAnT, and whether EMG can be used to observe a BLD during arm cycling.

1.8 Research Objectives

This study has two main objectives: 1) quantify the influence that arm dominance, unilateral or bilateral arm cycling, fatigue, WAnT resistance intensity, or crank position have on EMG amplitude of the biceps and triceps brachii; 2) create EMG profiles to assess whether there is a difference in activation of the biceps and triceps brachii between unilateral and bilateral trials, which could support a bilateral EMG deficit during arm cycling, and to break down at the difference in change in activation by clock phase and beginning, middle, and end of the WaNT.

1.9 Hypotheses

Based on previous arm dominance, cycling unilaterally or bilaterally, WAnT resistance, and crank position research all are expected to affect EMG amplitude (Chaytor et al., 2020; Niu et al., 2011; Pearcey et al., 2016; Saeterbakken et al., 2015; Yu et al., 2022). Specifically, the EMG amplitude for the biceps brachii and triceps breachii with be greatest at 6 and 12 o'clock, respectively (Nippard et al., 2019), higher forces output by the muscles result in higher EMG amplitudes (Bilodeau et al., 2003; Ricard, 2005), so greater BW% trials are expected to lead to greater EMG amplitudes for the biceps and triceps brachii for unilateral and bilateral WaNTs, Additionally, dominant bicpes and triceps brachii are expected to show greater EMG amplitudes when compared to the non-dominant biceps and triceps brachii (Lad et al., 2023). However, results vary on if fatigue contributes to a change in EMG amplitude (Hunter et al., 2003; Park, E., & Meek, 1993), so it is uncertain how or if it will influence the EMG data of this study. Additionally, based on the study done by Antolinez et al. (2024), the EMG amplitudes the triceps and biceps

brachii for the non-dominant unilateral and dominant unilateral WaNTs should be greater than the amplitudes for the triceps and biceps brachii during the bilateral WaNTs.

Chapter 2: Research Study

2.1 Methodologies

2.1.1 Participants

Eighteen (9 male and 9 female) recreationally active, healthy adults were recruited for the study. Of the participants recruited, only 12 completed all familiarization and experimental WaNT sessions. Prior to the study, participants did not have experience with upper body Wingates (WAnT). Exclusion criteria included having a previous history of upper limb injury in the last six months or pain that would otherwise prevent the participant from engaging in vigorous exercise. After the participants provided informed consent, they completed a Physical Activity Readiness Questionnaire (PAR-Q+) to ensure the task would be safe for them to perform. Hand dominance was assessed using the Edinburgh Handedness Inventory. Prior to data collection and analysis, participants were informed of all potential risks and benefits of the study and given the opportunity to raise any concerns or questions. All participants gave written informed consent. This study was approved by the Interdisciplinary Committee on Ethics in Human Research at Memorial University (No. 20250289) following the Tri-Council Policy Statement for Ethical Conduct on Research Involving Humans (TCPS2) in Canada, with full disclosure of potential risks to participants.

2.1.2 Arm Cycle Ergometer

Experimental data was recorded while the participant was seated on a computer operated, electromagnetically braked cycle ergometer, modified for arm cycling (Dynafit Pro; Racemate, Seattle, WA). All participants were seated in a padded armless chair, strapped with a secured belt, and positioned at a comfortable distance from the

crankshaft, such that the crank position on the opposite of one arm was close to full extension. The chair was adjusted so that the participants shoulders were approximately in line with the ergometer's axis of rotation. The participants also had their feet strapped to the floor to minimize movement. Ideal chair position was recorded for each participant and was used for each of their experimental sessions. During arm cycling, the biceps brachii are most active during the flexion phase, which is represented by a 6 o'clock crank position and the triceps brachii are most active during extension phase, which is represented by a 12 o'clock crank position (Nippard et al., 2019) (See Figure 1 for details).

2.1.3 Maximal Voluntary Contractions

For every session, after the participant was correctly positioned in the cycle ergometer, they performed a series of maximal voluntary isometric contractions (MVICs). All MVICs were performed on the cycle ergometer with the wheel locked in place, and the dominant arm in the target position (either 6 o'clock or 12 o'clock). The participants performed 3 MVICs at each position, lasting 5 seconds each, and the averaged MVIC at each position was calculated. During each MVIC, EMG data was recorded from the biceps brachii and the triceps brachii. For each participant, averaged MVIC EMG output was used to normalize the mean EMG data recorded during the WAnTs performed in the same session.

2.1.4 Arm Cycling WAnT

Each arm cycling WAnT consisted of a 10s warmup period at 60 rpm, followed by a 3s countdown (visual and auditory) prior to an electromagnetic brake applying a resistance of either 3%, 4%, or 5% of the participants body mass (BM) to the wheel of the

cycle ergometer. As the braking force was applied each participant cycled with maximal effort for the full 30s duration. Cadence feedback was provided visually to the participants via a real-time on-screen display of their RPM and verbal encouragement was provided for the duration of the WAnT. During each WAnT, EMG data was recorded unilaterally or bilaterally from the biceps brachii and triceps brachii, which was later broken down into their 6 o'clock and 12 o'clock data for each rotational cycle.

2.1.5 Electromyography Recording

EMG data was recorded from the biceps brachii, and the lateral head of the triceps brachii, based on SENIAM guidelines (Hermens et al., 2000) for the dominant and non-dominant arms using a CED 1902 and CED Power 1401 (Cambridge Electronic Designs, Cambridge, United Kingdom) amplifier data acquisition system and Spike 2 software (Cambridge Electronic Designs, Cambridge, United Kingdom) and sampled at 10 Hz. Each arm had disposable Ag-AgCl surface EMG electrodes (10 mm diameter; MediTrace^{TM} 130 Foam Electrodes, Massachusetts, USA) placed in a bipolar arrangement over the mid-point of the muscle belly, in line with muscle fibres, with a ground electrode placed on the lateral epicondyle. In preparation for electrode placement, the target skin was shaved, abraded with Nuprep Skin Preparation Gel (Weaver and Company, Aurora, Colorado, USA), and cleaned with an isopropyl alcohol wipe.

2.1.6 Experimental Sessions

Each participant completed a total of 12 WaNTs, 3 in the familiarization session, and 9 across three experimental sessions (3 per session) with a minimum of 24 hours between each session. Participants completed a total of 4 sessions, a familiarization session followed by 3 experimental sessions, with 24 to 48 hours between sessions. During the

familiarization session, participants completed a condensed version of all the tasks they would be asked to perform during the experimental sessions, consisting of elbow flexion and extension MVCs, 3 10s WAnTs at 3%, 4%, and 5% BM. Following the familiarization session, the participants randomly completed the remaining 3 experimental sessions consisting of isometric elbow flexion and extension MVICs at 12 o'clock and 6 o'clock respectively followed by 3 30s upper body WAnT at either 3%, 4%, or 5% of the participant's BM. Within each of the experimental sessions 2 parameters were randomized; The order in which the 3%, 4%, and 5% BW sessions would be (such as 3% BM in experimental session 1 4% BM in experimental session 2, and 5% BM in experimental session 3), and the order in which they performed the bilateral, unilateral dominant, and unilateral non-dominant WaNTs, Participants were prepped for biceps and triceps EMG for both arms and EMG was recorded from these muscles during all MVICs and upper body WAnTs in the experimental sessions. See Figure 1 for the experimental protocol.

2.1.7 Data Analysis

From the recorded EMG data, absolute values were plotted every 1ms and rectified and smoothed in Spike 2 with a time constant of 0.1s. Mean EMG was calculated bilaterally for each phase of interest for the triceps brachii and biceps brachii for all the WAnTs. Resulting data files were exported to Microsoft Excel for Mac 16.88 (Microsoft Corporation, Redmond Washington) for further analysis. Average aggregate activation curves of the triceps brachii and biceps brachii at every 10ms of the first 3 complete cycles, middle 3 complete cycles, and final 3 compete cycles of each WAnT were generated by averaging absolute rectified EMG for the duration of each cycle. The resulting data was

smoothed and visualized using Microsoft Excel for Mac 16.88 (Microsoft Corporation, Redmond Washington) as 12-line charts.

EMG data for the triceps brachii and biceps brachii in the dominant and non-dominant arms was recorded throughout each WAnT. Each WAnT was segmented into 3 major sections, with each section consisting of 3 complete and continuous rotational cycles. The 3 major sections were 1) beginning (e.g. the first three full rotations during the WAnT), 2) middle (e.g. the middle three rotations during the WAnT), and 3) end (e.g. the final three rotations during the WAnT). Each cycle was further broken down into 6 o'clock and 12 o'clock phases. Mean EMG for the triceps brachii and biceps brachii was averaged for each phase during each rotation and averaged for each section. Rectified mean EMG was visualized in Excel for each participant for each of the experimental conditions for both the triceps brachii and biceps brachii. Rectified mean EMG was then normalized for each participant relative to their rectified mean MVIC EMG data collected at the beginning of each session. Biceps brachii EMG WAnT data was normalized to the 6 o'clock MVIC EMG data. Triceps brachii EMG WAnT data was normalized to the 12 o'clock MVIC EMG data. Normalized data was represented as a percentage of activation, relative to MVIC activation. Normalized EMG was further averaged to find the mean EMG activation across the study sample for the unilateral dominant, unilateral non-dominant, and bilateral triceps brachii and biceps brachii at the beginning 6 o'clock and 12 o'clock phase, the middle 6 o'clock and 12 o'clock phase, and end 6 o'clock and 12 o'clock phase for 3%, 4%, and 5% WAnTs. Standard deviation was calculated for each reported average.

The percentage change in activation for the triceps brachii and biceps brachii at 6 o'clock and 12 o'clock relative to their activation at the beginning phase of the WAnT was

obtained by calculating the ratio of average EMG activity of the middle and end phases relative to the beginning phase. This was calculated at each WAnT intensity for the bilateral dominant, unilateral dominant, bilateral non-dominant, and unilateral non-dominant arms. The resulting ratios were visualized using Microsoft Excel for Mac 16.88 (Microsoft Corporation, Redmond Washington) as 4 smoothed line graphs of change in activation across time, as the arm completes each cycle.

The percentage change in muscle activation for the triceps brachii and biceps brachii was used to assess the difference in activation between bilateral and unilateral trials by calculating the ratio of the muscle's bilateral activation, relative to its unilateral activation. This calculation was repeated for both the dominant and non-dominant arms at 6 o'clock and 12 o'clock across the beginning, middle, and end segments of the 3%, 4% and 5% WAnTs. The formula used was ((unilateral target muscle/bilateral target muscle)*100)-100. The results were visualized as two horizontal bar charts with standard error bars, using Microsoft Excel for Mac 16.88 (Microsoft Corporation, Redmond Washington).

2.1.8 Statistical Analysis

All statistical analysis was performed using SPSS 29.0.2.0 (SPSS for Mac, IMB Corporation, Armonk, NY). The percentage change in activation between the biceps and triceps brachii Statistical analyses were performed on normalized EMG data. A four-way ANOVA was used to evaluate the factors of ARM (unilateral dominant: UD, unilateral non-dominant; UND), bilateral dominant; BLD and bilateral non-dominant; BLND), INTENSITY (3%, 4%, and 5%), CRANK POSITION (6 and 12 o'clock) and WAnT section (beginning, middle and end) on EMG activity of the biceps and triceps brachii.

Statistical significance for all tests was set at $p \leq 0.05$. If the ANOVA outcome was significant, pairwise comparisons were assessed post hoc using a Bonferroni correction. Partial eta-squared ($\eta^2 p$) measures indicating the magnitude of changes associated with significant main effects were provided and reported as small (<0.01), moderate (≥0.06), or large (≥0.14) (Cohen, 1992). The text and tables report data as mean ± SD.

2.2 Results

All rectified and smoothed (0.1ms) cycling and isometric, and normalized data for biceps and triceps brachii EMG can be seen in Tables 1-3. Table 4 is a summary of the main effects, post hoc tests and percentage differences (if there was a significant difference). Table 5 is a summary of p-values for significant interactions. Figure 2 shows average rectified and smoothed (0.1ms) EMG profiles for the biceps and triceps brachii over one revolution (360°) and the duration of EMG for ARM, INTENSITY and WAnT SECTION.

2.2.1 Main Effects

CRANK POSITION had a significant effect on EMG amplitude for the biceps brachii ($F_{(1, 2526)} = 816.091$, $p = <0.001$, partial $\eta^2 = 0.244$) and triceps brachii ($F_{(1, 2544)} = 216.091$, $p = <0.001$, partial $\eta^2 = 0.078$), respectively during the WAnT. The post hoc test indicated that the normalized mean EMG amplitude for the biceps brachii was 165.9% larger at 6 o'clock compared to 12 o'clock and 54.1% larger at 12 o'clock compared to 6 o'clock for the triceps brachii.

INTENSITY had a significant effect on EMG amplitude for the biceps ($F_{(2, 2526)} = 15.305$, $p = <0.001$, partial $\eta^2 = 0.012$) and triceps ($F_{(2, 2544)} = 16.407$, $p = <0.001$, partial $\eta^2 = 0.013$) brachii, respectively during the WAnT. The post hoc test indicated that the

normalized mean EMG amplitude for both biceps and triceps brachii were 16.4% (p = <0.001) and 21.4% (p = <0.001) larger at 4%BM compared to 3%BM, respectively. Mean EMG amplitude for both biceps and triceps brachii were 22.4% (p = <0.001) and 13.9% (p = <0.001) larger at 5%BM compared to 3%BM, respectively. The mean EMG amplitude for the biceps (p = 0.452) and triceps (p = 0.206) brachii were not different between 4%BM and 5%BM.

ARM had a significant effect on EMG amplitude for the biceps ($F_{(3, 2526)}$ = 18.012, p = <0.001, partial η^2 = 0.021) and triceps ($F_{(3, 2544)}$ = 496.541, p = <0.001, partial η^2 = 0.369) brachii, respectively during the WAnT. The post hoc test indicated that the normalized mean EMG amplitude for the biceps brachii was 17.4% (p = <0.001) larger for unilateral non-dominant arm compared to bilateral non-dominant arm and 33.2% (p= <0.001) larger for unilateral dominant arm compared to bilateral dominant arm. Normalized mean biceps brachii EMG for the unilateral dominant arm compared to the unilateral non-dominant arm (p = 0.213) and the bilateral non-dominant arm compared to the bilateral non-dominant arm (p = 0.722) were not significantly different. The post hoc indicated that the normalized mean EMG amplitude for the triceps brachii were 359.5% (p = <0.001) larger for the bilateral dominant arm compared to the bilateral non-dominant arm. The normalized mean EMG amplitude was 90.0% (p = <0.001) larger for the unilateral non-dominant arm compared to the bilateral non-dominant arm. The normalized mean EMG amplitude was 141.9% (p = <0.001) larger for the unilateral dominant arm compared to bilateral dominant arm. Normalized mean triceps brachii EMG for the unilateral dominant arm compared to the unilateral non-dominant arm (p = 0.183) was not significantly different.

WAnT SECTION had a significant effect on EMG amplitude for the biceps ($F_{(2,2478)} = 3.843$, $p = 0.022$, partial $\eta^2 = 0.003$) but the triceps ($F_{(2, 2496)} = 2.445$, $p = 0.087$, partial $\eta^2 = 0.002$) brachii. The post hoc test indicated that that normalized mean biceps brachii EMG amplitude was 11.3% ($p = 0.016$) larger for the middle section when compared to the beginning section. Comparisons made between mean normalized EMG amplitude for the beginning section compared to the end section ($p = 0.225$) and the middle compared to the end section ($p = 0.511$) were not significant.

2.2.2 Main Interaction Effects

There was no observed significant interaction for CRANK POSITION * INTENSITY * ARM for the biceps brachii ($F_{(6, 2526)} = 1.323$, $p = 0.243$, partial $\eta^2 = 0.003$) nor the triceps brachii ($F_{(6, 2544)} = 0.945$, $p = 0.461$, partial $\eta^2 = 0.002$) EMG.

There was a significant interaction for CRANK POSITION * ARM for the biceps ($F_{(3, 2526)} = 27.840$, $p = <0.001$, partial $\eta^2 = 0.032$) and triceps ($F_{(3, 2544)} = 14.466$, $p = <0.001$, partial $\eta^2 = 0.017$) brachii EMG amplitude. At 6 o'clock, mean normalized EMG of the arms decreased during a bilateral WAnT compared to a unilateral WAnT by 29.0% ($p = <0.001$) and 48.9 % ($p = <0.001$) for the biceps and triceps brachii, respectively. Mean normalized EMG of the dominant arm decreased compared to non-dominant arm by 8.11% ($p = 0.004$) and 128.0% ($p = <0.001$) for the biceps and triceps brachii, respectively. At 12 o'clock, mean normalized EMG increased during a bilateral WAnT compared to unilateral WAnT by 31.9% ($p = <0.001$) for triceps brachii. Mean normalized EMG for the dominant arm decreased compared to non-dominant by 11.9% ($p = <0.001$) and 98% ($p = <0.001$) for the biceps and triceps brachii, respectively. Furthermore, mean normalized EMG for unilateral arms, bilateral arms, the non-dominant arm, and the dominant arm increased at

6 o'clock compared to 12 o'clock by 222.3% (p = <0.001), 113.9% (p = <0.001), 140.4% (p = <0.001), and 195.0% (p = <0.001), respectively for the biceps brachii and decreased by 39.4% (p = <0.001), 31.8% (p = <0.001), 41.4% (p = <0.001), and 32.1% (p = <0.001) respectively for the triceps brachii.

There was a significant interaction for CRANK POSITION * INTENSITY for the biceps brachii ($F_{(2, 2526)}$ = 3.967, p = 0.019, partial η^2 = 0.003) but not in the triceps brachii ($F_{(2, 2544)}$ = 1.369, p = 0.254, partial η^2 = 0.001). Normalized mean biceps brachii EMG increased at 6 o'clock compared to 12 o'clock by 164.6% (p = <0.001), 180.5% (p = <0.001), and 154.0% (p = <0.001) during the 3, 4, and 5% BM WAnT respectively. Furthermore, normalized mean biceps brachii EMG at 6 o'clock decreased from 3 to 4% BM WAnT by 12.6% (p = <0.001). Normalized mean biceps brachii EMG at 12 o'clock decreased from 3 to 4% BM WAnT by 17.6% (p = 0.001). Figure 3 shows delta changes (percentage) in mean normalized EMG amplitude between unilateral WAnTs and bilateral WAnTs for the biceps and triceps brachii in the dominant and non-dominant arms.

There was a significant interaction for INTENSITY * ARM for the biceps brachii ($F_{(6, 2526)}$ = 7.402, p = <0.001, partial η^2 = 0.017) and triceps brachii ($F_{(6, 2544)}$ = 5.026, p = <0.001, partial η^2 = 0.012). Table 5 shows all interaction combinations and whether these interactions were significant. Normalized mean biceps brachii EMG for unilateral and non-dominant decreased from 3% to 4%BM WAnT by 7.55% (p = <0.001) and 20.0% (p = <0.001), respectively and unilateral from 4% to 5%BM WAnT by 18.9% (p = <0.001). Normalized mean biceps brachii EMG for unilateral, non-dominant and dominant all decreased from 3% to 5%BM WAnT by 12.9% (p = <0.001), 21.0% (p = <0.001), and 25.0% (p = <0.001), respectively. Normalized mean EMG for the biceps brachii during

bilateral WAnT decreased from 3% to 4% BM WAnT by 22.2% (p = <0.001) and increased from 4% to 5% BM WAnT by 19.8% (p = <0.001).

For the triceps brachii, mean normalized EMG for the unilateral arm, bilateral arm, non-dominant arm, and dominant arm decreased from 3% to 4% BM WAnT by 16.0% (p = <0.001), 18.7% (p = <0.001), 15.5% (p = <0.001) ,and 18.6% (p = <0.001), respectively and decreased from 3% to 5% by 4.6% (p = <0.001), 16.15% (p = <0.001), 1.12% (p = <0.001), and 15.35% (p = <0.001), respectively. Mean normalized EMG for the unilateral arm and bilateral arm increased from 4% to 5% BM WAnT by 13.7% (p = <0.001), 2.82% (p = <0.001), respectively. Furthermore, normalized EMG for the biceps brachii during bilateral compared to unilateral WAnTs at 3% and 4% BM decreased by 21.6% (p = <0.001) and 34% (p = <0.001). Normalized biceps brachii EMG for the dominant arm compared to the non-dominant arm increased at 4% BM WAnTs by 9.33% (p = 0.008).

Mean normalized EMG for the triceps brachii during bilateral WAnTs increased compared to unilateral WAnTs and normalized EMG for the dominant arm increased compared to non-dominant arm at 3%, 4%, 5% BM WAnTs by 45.9% (p = <0.001), 124.8% (p = <0.001), and 41.3% (p = <0.001) and 116.3% (p = <0.001), 27.8% (p = <0.001) and 87.5% (p = <0.001), respectively.

No significant interaction was observed for WAnT SECTION * INTENSITY * CRANK POSITION * ARM for the biceps brachii ($F_{(12, 2478)}$ = 0.582, p = 0.859, partial η^2 = 0.003) nor the triceps brachii F(12, 2496) = 0.142, p = 1.000, partial η^2 = 0.001. Furthermore, within the biceps brachii and triceps brachii, no significant interaction was shown for INTENSITY * CRANK POSITION * WAnT SECTION ($F_{(4, 2478)}$ = 0.201, p = 0.938, partial η^2 = 0.000); ($F_{(4, 2496)}$ = 0.351, p = 0.844, partial η^2 = 0.001), INTENSITY *

ARM * WAnT SECTION ($F_{(12, 2478)} = 0.210$, $p = 0.998$, partial $\eta^2 = 0.001$); ($F_{(12, 2496)} = 0.221$, $p = 0.998$, partial $\eta^2 = 0.001$), OR CRANK POSITION * ARM * WAnT SECTION ($F_{(6, 2478)} = 0.492$, $p = 0.815$, partial $\eta^2 = 0.001$); ($F_{(6, 2496)} = 0.301$, $p = 0.937$, partial $\eta^2 = 0.001$).

There was a significant interaction observed for WAnT SECTION * CRANK POSITION ($F_{(2, 2478)} = 6.681$, $p = <0.001$, partial $\eta^2 = 0.005$), but not the WAnT SECTION * INTENSITY ($F_{(4, 2478)} = 0.411$, $p = 0.801$, partial $\eta^2 = 0.001$) or WAnT SECTION * ARM ($F_{(6, 2478)} = 0.141$, $p = 0.991$, partial $\eta^2 = 0.000$) for the biceps brachii EMG amplitude output. Comparing mean normalized EMG for WAnT SECTION * CRANK POSITION, the biceps brachii EMG during the beginning, middle, and end increased at 6 o'clock compared to 12 o'clock by 131.5% ($p = <0.001$), 165.8 ($p = <0.001$), and 206.4% ($p = <0.001$), respectively. Going from the beginning section to the middle section the biceps brachii mean normalized EMG for the 6 o'clock position increased by 15.95% ($p = <0.001$). Going from the middle section to the end section normalized mean EMG for the 12 o'clock positions decreased by 13.8% ($p = <0.001$). Going from the beginning to the end section normalized mean EMG values for the 6 o'clock and 12 o'clock position increased by and 15.15% ($p = <0.001$) and decreased by 13.05% ($p = <0.001$), respectively. See Figure 4 for average change in normalized EMG amplitude at each crank position for the middle and end segments, expressed as a percentage of the beginning 6 and 12 o'clock positions at each WAnT intensity.

There was no significant interactions were observed for WAnT SECTION * INTENSITY ($F_{(4, 2496)} = 0.345$, $p = 0.848$, partial $\eta^2 = 0.001$), WAnT SECTION * CRANK

POSITION ($F_{(2, 2496)} = 0.476$, $p = 0.621$, partial $\eta^2 = 0.000$), or WAnT SECTION * ARM ($F_{(6, 2496)} = 0.437$, $p = 0.855$, partial $\eta^2 = 0.001$) for triceps brachii EMG amplitude output.

2.3 Discussion

The most important findings of this study were that during arm cycling: 1) EMG muscle activity looks fundamentally different depending on the muscle in question, the crank position of the arm, the intensity at which the muscles have to work at, and if the arms are moving bilaterally or unilaterally, 2) with few exceptions, the duration of a rotation, expressed as time, and correlated with fatigue (as a participant fatigued, each roation took longer to complete) time did not meaningfully affect the mean EMG amplitude when comparing beginning, middle, and end to each other, and 3) from the tested resistances there is a maximum resistance at which arms can cycle against to obtain the greatest EMG amplitude and going beyond this resistance will not lead to a further increase in EMG amplitude. These findings show that EMG amplitude during 30s maximal arm cycling is highly dependent on phase, resistance, and arm. In a single sentence, within this study, the aspects most impactful on mean EMG of the biceps and triceps brachii were arm dominance, and if the task was performed unilaterally or bilaterally, crank position, and fatigue (indicated by time) did not affect mean EMG.

2.3.1 EMG amplitude during WAnT is highly influenced by intensity, crank position, and arm, but not WAnT section.

In the current study, EMG amplitude was highly affected by the intensity of a given WAnT, whether the arm was at the 6 or 12 o'clock position, and whether the participant was cycling unilaterally with their dominant or non-dominant arm. For the biceps brachii, crank position had the greatest effect (partial $\eta^2 = 0.244$) on EMG amplitude. Furthermore,

biceps brachii peak activation during maximal elbow flexion occurred at full flexion (6 o'clock) and then deactivated and reached its lowest point of activation during maximal elbow extension (12 o'clock) (Yu et al., 2022; Date et al., 2021).

Conversely, the single most impactful factor for the triceps brachii EMG during the WAnT was ARM (partial $\eta^2 = 0.369$). Similar to the study by Krzysztofik et al. (2021), this study also showed that the triceps brachii were more active in the dominant arm, when compared to the non-dominant arm. Interestingly, the findings in this study disagree with those of Kuhtz-Buschbeck and Keller (2019) and Oyama and Sako (2015), where they saw similar levels of activity between dominant and non-dominant arms and greater levels of co-contraction in the non-dominant arms compared to the dominant arm, respectively. This discrepancy in findings may have been due to the experimental activity selected, suggesting that motor activation strategies employed by the dominant and non-dominant arms may be different for locomotor tasks, when compared to non-locomotor tasks. When looking at activation of the triceps brachii across unilateral and bilateral conditions, there was less consistency whether unilateral or bilateral activation led to an increase in EMG activity. Unilateral non-dominant activity was greater than bilateral non-dominant activity by 90%, which disagrees with the findings of Nadzalan et al., (2019) where they found no significant difference between bilateral and unilateral triceps brachii activation. However, when comparing bilateral dominant to unilateral dominant arm, there was an increase in the bilateral arms by 141.9%. This falls in line with the increase in activation observed during bilateral tasks by Serrau et al. (2011).

2.3.2 The biceps and triceps brachii are recruited differently within the WAnT.

Within this study, the triceps and biceps were shown to be recruited differently during the WAnT. The activation of the triceps brachii exhibits different phase dependent patterns than the biceps brachii (Chaytor et al., 2020). This is clearly shown in Figures 2 and 4. While the biceps brachii were shown to have significant fluctuations in their EMG amplitude, the triceps brachii did not demonstrate the same EMG activation peaks and troughs. Rather, when the triceps brachii were active, they maintained a much flatter activation peak, which was most notable during unilateral WAnTs. The biceps brachii is a flexor muscle (Tiwana et al., 2018), and the triceps brachii is an antagonistic extensor muscle to the biceps (C. W. Moore & Rice, 2017). The difference in EMG activation curves demonstrate that the triceps brachii was not recruiting and de-recruiting completely complimentary to the biceps brachii. More specifically, when the triceps brachii was active, they reached their peak at the expected 12 o'clock position, however, during peak elbow flexion, they did not fully turn off. Instead, once the triceps turned on, they remained on, even when the biceps brachii were at peak activation. The flatter nature of the triceps brachii EMG amplitudes for most conditions, is consistent with the idea that the triceps brachii acts as a stabilizing muscle (Moore et al., 1985) during arm-cycling during submaximal (Chaytor et al., 2020), and as we show here, maximal intensity.

A potential physiological reason for the differences seen between biceps and triceps brachii EMG could be due to the difference in muscle fibre distribution between them (Elder et al., 1982). Looking at animal models, there is evidence that shows the triceps brachii has a larger amount of slow twitch (type I) muscle fibers in comparison to the biceps brachii (Morton et al. 1984; Van Den Hoven et al., 1985; Roy et al., 1984). Looking at the

EMG profile curves (Figure 2), the biceps brachii has a more distinct on and off pattern when compared to the triceps brachii, since fast twitch (type II) muscle fibers are more suited to rapid, explosive on/off activations as opposed to slow twitch muscle fibres, which are better suited to more sustained activation patterns (Colliander et al., 1988). The biceps brachii shows a much steeper slope, culminating in a pronounced peak, followed by decruitment, which is more in-line with the recruitment curve for fast twitch fibres (Maglischo, E.W., 2011). Whereas the triceps brachii recruitment slope is shallower, leading to more of a plateau, than a pronounced peak, which is more in line with the recruitment curve of slow twitch fibres (Maglischo, E.W. 2011). The differing proportions of type I and type II muscle fibers are a potential explanation for the differing EMG profile differences of the biceps brachii and triceps brachii during a 30s WAnT; where the biceps brachii are primary cyclical movers and the triceps brachii are primarily stabilizing the movement.

2.3.3 Time does not completely account for changes in EMG amplitude when considering fatigue development during a 30s WAnT.

Within the confines of this study, WAnT section did not significantly impact mean EMG amplitude, apart from the biceps brachii from the beginning (first three complete cycles) to middle section (middle three complete cycles) of the WAnT. This shows that neural drive to both the biceps and triceps brachii remains relatively constant when performing a 30s WAnT. Additionally, the observed deficit or facilitation remains relatively consistent across beginning, middle, and end segments (Figure 3). The similar EMG outputs for beginning, middle, and end segments of the WAnTs suggest that neural drive remains relatively consistent, even as the participant fatigues. However, Figure 2

highlights a potential issue or consideration to make when only utilizing mean EMG amplitude during a fatiguing task. While the amplitudes may have been similar for beginning, middle, and end segments, when plotted, the duration of each cycle increased with as time progressed, indicating an increase with fatigue. The observation of mean EMG remaining similar across time, but each cycle taking longer to suggests that while the neural drive may be similar from beginning to end, fatigue is occurring as each rotation takes longer to complete. Suggesting that, if the fatigue indicated by the slower movement speed is a primary consideration during a study, it may be more relevant to assess area under the curve, in addition to EMG amplitude.

2.3.4 There is likely an optimal WAnT resistance for generating maximal neural drive.

Within this study, there was a clear increase in EMG amplitude seen from 3% BW WAnTs compared to the 4% BW WAnTs, which supports the notion that as intensity (in this case resistance) of activity increases, so do the neural drive required to perform the task (Miller et al., 2020; Tøien et al., 2021). However, an interesting finding was that when comparing the 4% BW WAnTs to the 5% BW WAnTs, a decrease in EMG amplitude occurred. This finding indicates that there appears to be maximum output for effort to drive the biceps and triceps brachii for a given task (not designed to result in failure) that has multiple resistances. During a given WAnT, if the resistance is set higher than what yields maximal EMG amplitude, EMG amplitude may decrease when compared to a lower resistance that was either at or below the optimal load intensity, as was observed in this study with the 4% and 5% BW WAnTs. Thus, there is an optimal resistance for generating

maximum drive to the biceps and triceps brachii (and potentially neural drive) and obtaining maximum EMG amplitude during an upper body WAnT.

2.3.5 EMG amplitudes are significantly different for the same muscle across unilateral and bilateral WAnTs.

While this study did not assess BLD in context of intensity or duration, there was a clear and evident difference in EMG activity when comparing the unilateral WAnTs to their bilateral counterparts. The delta EMG activity for each muscle along with intensity, clock phase, and section of the WAnT shown in Figure 3 illustrates whether an activation deficit or facilitation took place. The findings of this study are supported by previous studies that demonstrated that a BLD during dynamic movements can be observed by analyzing EMG data (Hay et al., 2006; Pleša et al., 2022). Additionally, the findings in this paper may add support to the notion that any resulting EMG BLD does not exist in isolation and can be influenced by body position (Škarabot et al., 2016). Variance in EMG deficit at varying intensities and between the dominant and non-dominant arms also supports the idea that BLD is the result of higher order neural processes (Škarabot et al., 2016). Specifically, the variance in observed EMG deficit supports the idea that one hemisphere is able to inhibit the other, leading to a reduction in EMG output (Škarabot et al., 2016). However, based on the variance observed in this study, that reduction may not be constant

As shown in Figure 3, across all the intensities aside from the 5% non-dominant 12 o'clock, the delta EMG was more pronounced for the triceps brachii than the biceps brachii. The triceps brachii exhibiting more pronounced EMG deltas, suggests that activity is more affected by unilateral/bilateral movements, when compared to the biceps brachii. Within the non-dominant arm, the triceps brachii demonstrated a negative delta from unilateral to

bilateral WAnTs, indicating that it was more active unilaterally, as opposed to bilaterally. An increase in EMG activity of the agonist muscle during either contraction or extension is seen as co-contraction (Yoshitake et al., 2016) which could explain the observed delta EMG. When comparing the observed delta EMG across the dominant and non-dominant arms, the amplitude was reversed for all triceps brachii WAnT intensities, and all biceps brachii WAnT intensities at the 12 o'clock position, except for the beginning 3% WAnT. That is to say that if there was a negative delta for the dominant arm, the non-dominant arm showed a positive delta, and vice versa. This change in EMG delta across the non-dominant and dominant arms further supports the idea that activity pattens for the dominant and non-dominant arms may differ (Adam et al., 1998; Wang & Sainburg, 2007). If both arms expressed activity in the exact same pattern for the unilateral WAnTs compared to the bilateral WAnTs, then the EMG deltas for the dominant and non-dominant arms would have likely shared the same patterns.

2.4 Methodological considerations

The first main consideration was using normalized rectified mean EMG values for data analysis. This data analysis method was chosen because it allowed for normalization of each participants EMG data during cycling to their own isometric EMG data at both the 6 o'clock and 12 o'clock position (Burden, 2010). Normalizing the cyclical EMG value to the isometric values (peak biceps brachii to 6 o'clock MVIC and peak triceps brachii to 12 o'clock MVIC), allowed an estimate of the activity of the muscles for each participant (Norcross et al., 2009) and the generation of a reliable way to normalize means across all participants and create an aggregate mean at each trial (Knutson et al., 1994). This was necessary because activity and excitability do fluctuate between participant (Krause &

Kadosh, 2014; Metsomaa et al., 2021) and between days for within participant (Xu & Barak, 2017). Another method that could have been used to analyze the data was area under the curve. Using area under the curve is a method for analyzing EMG activity over the duration of a task. The resulting value is a function of EMG amplitude (mV) and time (s). Since time is accounted for with area under the curve, and not just voltage, the same normalization of utilizing a sustained MVIC would not be appropriate? for normalizing a cyclical movement, because one of the factors that creates the value (time) is not controlled for during a WAnT, which could result in different results.

Participants performed the WAnTs on a custom-built upper body cycle ergometer. The use of a cycle ergometer, specifically made for arm cycling, allowed for an effective way to ensure that trials were consistent across days and participants. This was aided by the harness attached to the seat, which limited excess upper body movement. Limiting unwanted movement during the WAnTs was a priority for this study because excess movement can sully the data obtained from the test (Lericollais et al., 2010). Additionally, the electromagnetic brake, featured on the cycle ergometer allowed for precise tuning of the resistance (Nayak et al., 2023) for each participant. However, it is unknow if the position of the participant relative to the arm crank allows for optimal power production during the WAnT. This would affect EMG activity of the muscles used in the arm cycling task.

Trial intensities were another consideration within this study. This was the first time our lab had performed a unilateral WAnT. While resistance intensities for leg cycling WAnTs can reach as high as 10%BM (Jaafar et al., 2015), a maximum intensity of 5%BW was decided in part due to it being the standard bilateral arm cycling WAnT resistance used

in the lab (Pearcy et al. 2016), and partly because it was not too challenging to perform unilaterally. If the intensities were higher than 5%, the results may have been different. Furthermore, unilateral cycling at resistances beyond 5% BW may be too high to complete a WAnT.

All WAnT experiments were quasi-randomized whereby the participant completed 3 WAnTs per experiment at the same resistances and the arm was randomized. If the experiment was completely randomized the results may have differed, since each participant may not have performed all of 3%, 4%, or 5%BM sessions or the all of the unilateral dominant, unilateral non-nondominant, or bilateral WaNTs for each session.

A 20-minute recovery period was selected due to the fatiguing nature of the WAnTs, both centrally and peripherally (Fernandez-del-Olmo et al., 2011). While fatigue was expected over the course of a WAnT, it was imperative to minimize the chance of fatigue from one WAnT affecting subsequent WAnTs. As such, 20 minutes was chosen as a slightly longer passive rest period compared to other studies (Kirkpatrick & Burrus, 2019) to ensure each participant had enough time to properly recover before starting the next WAnT. However, simply looking at EMG values cannot say for certain that some level of fatigue occurred from one WAnT to the next, even with 20 minutes of rest between them. Having a starting power output would determine this, however power data was not available for this study. If there was overall fatigue it may have affected the study's outcomes.

Lastly, this study predominantly sampled from a right arm dominant population. While there are asymmetries in the motor cortex between left and right arm dominant individuals (Amunts et al., 1996b), it is unclear if evenly including left-handed and right-

handed participants would have affected the study. Since only 12 participants had enough data completed to use for the entirety of this study, there were not enough participants to look at sex differences, nor were there sufficient participants to assess whether there would be a difference between untrained and trained participants.

2.5 Conclusion

This study demonstrated that EMG amplitude over the course of a fatiguing arm cycling task is highly dependent on crank position, applied resistance, and whether the task is performed with the dominant arm, non-dominant arm, or bilaterally with both arms. Fatigue was clearly observable when looking at the raw EMG curve, since each rotation increased in duration over the course of the WAnT. This study was also the first of its nature to clearly outline that a delta in EMG activity exists between unilateral and bilateral arm cycling. This finding added further support for a BLD/BLF being observable by EMG during a dynamic task.

Further research could be useful to determine if the different activation strategies are also observed in male and female populations, or a trained and untrained population. Lastly, while delta EMG amplitude for the biceps and triceps brachii from unilateral to bilateral WAnTs was observed within the EMG profiles of this study, it was not assessed based on performance measures. As such, an interesting avenue for future research would be to fully assess WAnT EMG profiles along with performance measures for BLD/BLF and determine which factor; arm dominance, intensity, sex, WAnT segment (fatigue) have the greatest impact on BLD/BLF.

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Table 1. Rectified and smoothed EMG data mean and standard deviation values during WAnT experimental trials. All EMG data is in mV.

	12 o'clock								
		UD		UND		BLND		BLD	
Rectified and smoothed (0.1ms) EMG	**Biceps**	M	SD	M	SD	M	SD	M	SD
3%	Beginning	0.125	0.073	0.137	0.122	0.120	0.054	0.130	0.084
	Middle	0.119	0.069	0.127	0.098	0.249	0.232	0.148	0.133
	End	0.104	0.050	0.115	0.076	0.134	0.076	0.133	0.096
4%	Beginning	0.099	0.068	0.088	0.044	0.142	0.159	0.094	0.064
	Middle	0.092	0.050	0.088	0.033	0.147	0.157	0.103	0.064
	End	0.069	0.024	0.080	0.028	0.088	0.045	0.086	0.039
5%	Beginning	0.091	0.049	0.093	0.061	0.261	0.274	0.108	0.051
	Middle	0.093	0.058	0.104	0.058	0.177	0.157	0.119	0.057
	End	0.077	0.038	0.083	0.028	0.108	0.045	0.102	0.049
	6 o'clock								
		UD		UND		BLND		BLD	
Rectified and smoothed (0.1ms) EMG	**Biceps**	M	SD	M	SD	M	SD	M	SD
3%	Beginning	0.337	0.109	0.374	0.141	0.357	0.118	0.396	0.129
	Middle	0.401	0.167	0.445	0.145	0.357	0.159	0.462	0.197
	End	0.442	0.199	0.473	0.177	0.391	0.164	0.465	0.153
4%	Beginning	0.325	0.144	0.348	0.140	0.323	0.139	0.366	0.148
	Middle	0.406	0.185	0.448	0.195	0.341	0.145	0.439	0.204
	End	0.367	0.174	0.500	0.290	0.351	0.177	0.471	0.249
5%	Beginning	0.339	0.128	0.348	0.195	0.323	0.113	0.409	0.165
	Middle	0.402	0.140	0.431	0.211	0.388	0.145	0.499	0.213
	End	0.379	0.141	0.418	0.290	0.376	0.177	0.500	0.248
	12 o'clock								
		UD		UND		BLND		BLD	
Rectified and smoothed (0.1ms) EMG	**Triceps**	M	SD	M	SD	M	SD	M	SD
3%	Beginning	0.295	0.132	0.593	0.401	0.300	0.132	0.598	0.374
	Middle	0.325	0.160	0.660	0.418	0.301	0.157	0.658	0.507
	End	0.305	0.154	0.656	0.410	0.285	0.112	0.635	0.386
4%	Beginning	0.280	0.144	0.553	0.363	0.271	0.141	0.635	0.392
	Middle	0.293	0.136	0.591	0.350	0.278	0.132	0.666	0.372
	End	0.276	0.133	0.541	0.318	0.309	0.153	0.627	0.345
5%	Beginning	0.326	0.188	0.577	0.389	0.266	0.148	0.651	0.419
	Middle	0.374	0.209	0.616	0.400	0.357	0.132	0.716	0.479
	End	0.354	0.227	0.538	0.318	0.410	0.153	0.678	0.459
	6 o'clock								
		UD		UND		BLND		BLD	
Rectified and smoothed (0.1ms) EMG	**Triceps**	M	SD	M	SD	M	SD	M	SD

	Beginning	0.168	0.083	0.432	0.321	0.169	0.066	0.439	0.331
	Middle	0.179	0.091	0.473	0.373	0.204	0.126	0.512	0.420
3%	End	0.170	0.078	0.389	0.223	0.160	0.080	0.488	0.410
	Beginning	0.325	0.144	0.332	0.233	0.154	0.057	0.383	0.294
	Middle	0.406	0.185	0.357	0.197	0.150	0.047	0.409	0.318
4%	End	0.367	0.174	0.349	0.206	0.132	0.057	0.340	0.227
	Beginning	0.339	0.128	0.377	0.324	0.204	0.076	0.442	0.391
	Middle	0.402	0.140	0.366	0.224	0.196	0.047	0.410	0.293
5%	End	0.379	0.141	0.360	0.206	0.154	0.057	0.348	0.235

UD: Unilateral Dominant arm, UND: Unilateral Non-dominant arm, BLD: Bilateral Dominant arm, BLND: Bilateral Non-dominant arm, M: mean, SD: standard deviation.

Table 2. Table of average MVIC. All EMG data is in mV.

Mean MVIC for U Normalization	DT @ 12	DB @ 6	NT @ 12	NB @ 6
5%	0.673	0.377	0.339	0.384
4%	0.712	0.421	0.301	0.345
3%	0.692	0.369	0.308	0.334
SD of Mean MVIC for U Normalization	DT @ 12	DB @ 6	NT @ 12	NB @ 6
5%	0.406	0.193	0.174	0.180
4%	0.404	0.360	0.118	0.194
3%	0.376	0.205	0.168	0.205
Mean MVIC for BL Normalization	DT @ 12	DB @ 6	NT @ 12	NB @ 6
5%	0.687	0.484	0.374	0.559
4%	0.673	0.443	0.343	0.555
3%	0.723	0.443	0.334	0.466
SD of Mean MVIC for BL Normalization	DT @ 12	DB @ 6	NT @ 12	NB @ 6
5%	0.423	0.240	0.173	0.363
4%	0.392	0.212	0.153	0.375
3%	0.365	0.278	0.189	0.279

U: Unilateral, BL: Bilateral, SD: Standard Deviation, BL: Bilateral, U: Unilateral, DT: Dominant Triceps, DB: Dominant Biceps, NT: Non-dominant triceps, NB: Non-dominant Biceps. 6: MVIC taken at crank position at 6 o'clock. 12: MVIC taken at crank position at 12 o'clock position.

Table 3. Normalized average EMG (mean and standard deviation) during WAnT experimental trials.

	12 o'clock								
		UD		UND		BLND		BLD	
Normalized	**Biceps**	M	SD	M	SD	M	SD	M	SD
	Beginning	55.74	54.24	64.66	64.27	40.56	32.98	44.61	39.33
3%	Middle	51.52	46.28	53.74	135.72	68.28	52.16	47.97	48.04
	End	45.18	36.08	45.57	157.67	56.17	46.79	44.31	39.94
	Beginning	61.70	115.71	42.88	44.64	47.92	62.01	29.84	32.86
4%	Middle	55.95	95.30	39.83	93.23	48.60	60.74	32.63	35.94
	End	45.08	67.19	32.71	112.20	44.16	57.99	28.04	32.51
	Beginning	41.06	48.22	31.43	26.64	76.33	92.60	38.39	57.01
5%	Middle	44.63	59.48	30.03	33.71	66.93	119.99	40.77	64.06
	End	32.68	29.56	26.97	67.38	63.92	85.69	35.61	48.73
	6 o'clock								
		UD		UND		BLND		BLD	

Normalized	**Biceps**	M	SD	M	SD	M	SD	M	SD
3%	Beginning	132.24	83.64	143.01	88.96	102.20	56.61	107.02	49.74
	Middle	155.55	89.89	172.05	215.52	111.82	84.09	126.96	69.79
	End	154.77	62.22	190.34	216.54	110.81	75.03	130.09	62.80
4%	Beginning	151.36	134.35	125.92	74.02	86.36	51.06	80.81	37.36
	Middle	161.84	120.68	149.30	181.66	93.88	52.18	92.07	36.66
	End	177.04	137.15	125.33	182.21	89.67	38.86	96.72	34.71
5%	Beginning	117.22	51.75	109.86	60.80	79.91	57.41	95.83	54.05
	Middle	139.26	53.28	126.29	135.27	99.90	67.38	115.64	57.55
	End	136.69	61.31	111.57	145.46	90.85	48.40	119.64	59.67
	12 o'clock								
		UD		UND		BLND		BLD	
Normalized	**Triceps**	M	SD	M	SD	M	SD	M	SD
3%	Beginning	97.14	30.19	115.54	36.95	51.10	26.87	233.75	147.38
	Middle	109.46	30.11	124.38	128.83	47.31	20.40	254.38	183.20
	End	107.21	32.13	117.63	122.55	59.51	61.66	260.47	184.06
4%	Beginning	91.33	20.54	98.13	29.72	47.12	25.39	206.88	101.04
	Middle	95.86	18.72	102.30	111.79	47.88	20.88	225.29	112.30
	End	88.93	24.05	95.58	106.18	46.12	22.57	211.92	100.69
5%	Beginning	98.84	27.71	120.66	80.07	43.23	23.62	208.63	124.59
	Middle	102.93	19.59	133.00	192.99	60.39	36.77	235.14	150.09
	End	90.01	16.46	119.67	193.15	55.78	35.00	224.31	130.86
	6 o'clock								
		UD		UND		BLND		BLD	
Normalized	**Triceps**	M	SD	M	SD	M	SD	M	SD
3%	Beginning	69.76	33.71	70.36	36.82	30.31	18.94	182.42	140.23
	Middle	76.69	38.21	72.98	94.29	35.95	23.43	210.20	160.98
	End	64.82	28.04	70.53	96.27	34.95	33.82	200.29	153.54
4%	Beginning	55.82	26.99	56.88	32.03	29.47	17.91	141.95	121.42
	Middle	57.07	25.71	59.36	80.17	28.37	17.03	155.51	126.23
	End	57.08	31.24	61.69	78.81	28.84	21.81	131.10	101.11
5%	Beginning	63.66	35.94	60.73	31.00	36.13	22.51	144.45	131.67
	Middle	63.20	30.97	66.24	73.37	34.14	26.66	138.18	118.28
	End	64.18	40.89	62.93	104.10	35.78	30.01	120.87	106.31

UD: Unilateral Dominant arm, UND: Unilateral Non-dominant arm, BLD: Bilateral Dominant arm, BLND: Bilateral Non-dominant arm, M: mean, SD: standard deviation.

Table 4. Main effect and post hoc test summary (based on normalized EMG) and percentage differences.

	Biceps	Triceps	Biceps	Triceps	Biceps	Triceps	Biceps	Triceps
	UND vs UD		BLND vs BLD		UND vs BLND		UD vs BLD	
ARM	-	-	-	+ BLD 359.5% > BLND	+ UND 17.4% > BLND	+ UND 90.0% > BLND	+ UD 33.2% > BLD	+ BLD 141.9% > UD
	Beginning vs Middle		Middle vs End		Beginning vs End			
WAnT SECTION	+ Middle 11.3% > Beginning	-	-	-	-	-		
	3% BM vs 4% BM		3% BM vs 5% BM		4% BM vs 5% BM			

INTENSITY	+ 3%BM 16.4% > 4%BM	+ 3%BM 21.4% > 4%BM	+ 3%BM 22.4% > 5%BM	+ 3%BM 13.9% > 5%BM	-	-	

+ Indicates significant difference ($P < 0.05$).
- Indicates no significant difference ($P > 0.05$).
UD: Unilateral Dominant arm, UND: Unilateral Non-dominant arm, BLD: Bilateral Dominant arm, BLND: Bilateral Non-dominant arm.

Table 5. Interaction P value table (based on normalized values). P values only reported for significant interactions.

P value interactions	Biceps	Triceps
INTENSITY * CRANK POSITION	+ (0.019)	-
INTENSITY * ARM	+ (<0.001)	+ (<0.001)
INTENSITY * WAnT SECTION	-	-
CRANK POSITION * ARM	+ (<0.001)	+ (<0.001)
CRANK POSITION * WAnT SECTION	+ (0.001)	-
ARM * WAnT SECTION	-	-
INTENSITY * CRANK POSITION * ARM	-	-
INTENSITY * CRANK POSITION * WAnT SECTION	-	-
INTENSITY * ARM * WAnT SECTION	-	-
CRANK POSITION * ARM * WAnT SECTION	-	-
INTENSITY * CRANK POSITION * WAnT SECTION * ARM	-	-

+ Indicates interaction reached significance ($P = < 0.05$).
- Indicates interaction did not reach significance ($P = > 0.05$).

Figure Legend

Figure 1. Experimental upper body WAnT protocol. Sessions 2, 3, and 4 were experimental sessions. Sessions were randomized.

Figure 2. Study group aggregate average rectified and smoothed (0.1ms) EMG during a single revolution (360°) of the crank for the beginning, middle, and end section of each WAnT for all arm configurations and WAnT intensities. **A)** 3% BM, **B)** 4% BM, and **C)** 5% BM WAnT. The first column represents Bilateral Non-dominant EMG, second column represents Bilateral Dominant EMG, third column represents Unilateral Non-dominant EMG, and the fourth column represents Unilateral Dominant EMG.

Figure 3. Study group mean aggregate percentage shift in normalized biceps and triceps brachii EMG activation from unilateral to bilateral WAnTs for **A)** the dominant arm and **B)** the non-dominant arm. Represented as the mean percentage increase or decrease of normalized EMG activity when comparing unilateral arm cycling to bilateral arm cycling WAnTs. A positive delta indicates that normalized EMG for a muscle at a given WAnT intensity was greater during a bilateral WAnT when compared to a unilateral WAnT. A negative delta indicates that normalized EMG for a muscle at a given WAnT intensity was greater during a unilateral WAnT than during a bilateral WAnT. Bars indicate standard error.

Figure 4. Study group aggregate mean EMG amplitude shift of the biceps brachii and triceps brachii from the middle of WAnT to the end of WAnT across all intensities, crank

positions, and both muscles, expressed as a percentage of the beginning EMG amplitude for **A)** Unilateral and **B)** Bilateral WAnTs. The left column represents the Non-dominant arm, and the right panel represents the Dominant arm.

Figure 1

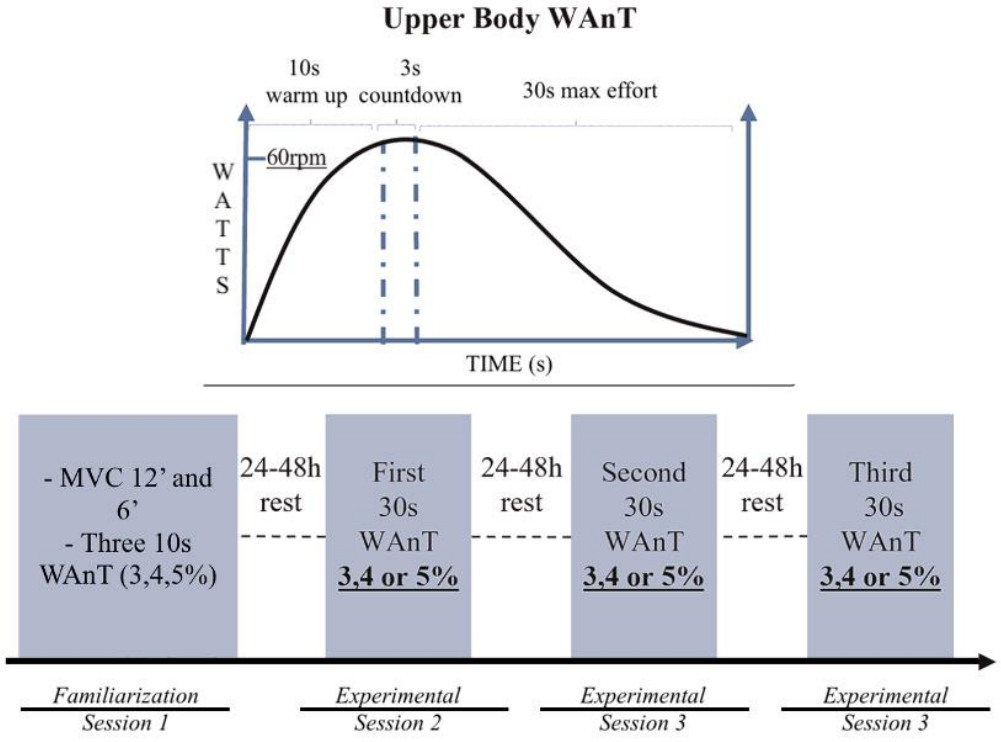

Upper Body WAnT
10s
warm up
3s
countdown
30s max effort
60rpm
WATTS
TIME (s)
- MVC 12' and 6'
- Three 10s WAnT (3,4,5%)
24-48h rest
First 30s WAnT 3,4 or 5%
24-48h rest
Second 30s WAnT 3,4 or 5%
24-48h rest
Third 30s WAnT 3,4 or 5%
Familiarization Session 1
Experimental Session 2
Experimental Session 3
Experimental Session 3

Figure 2

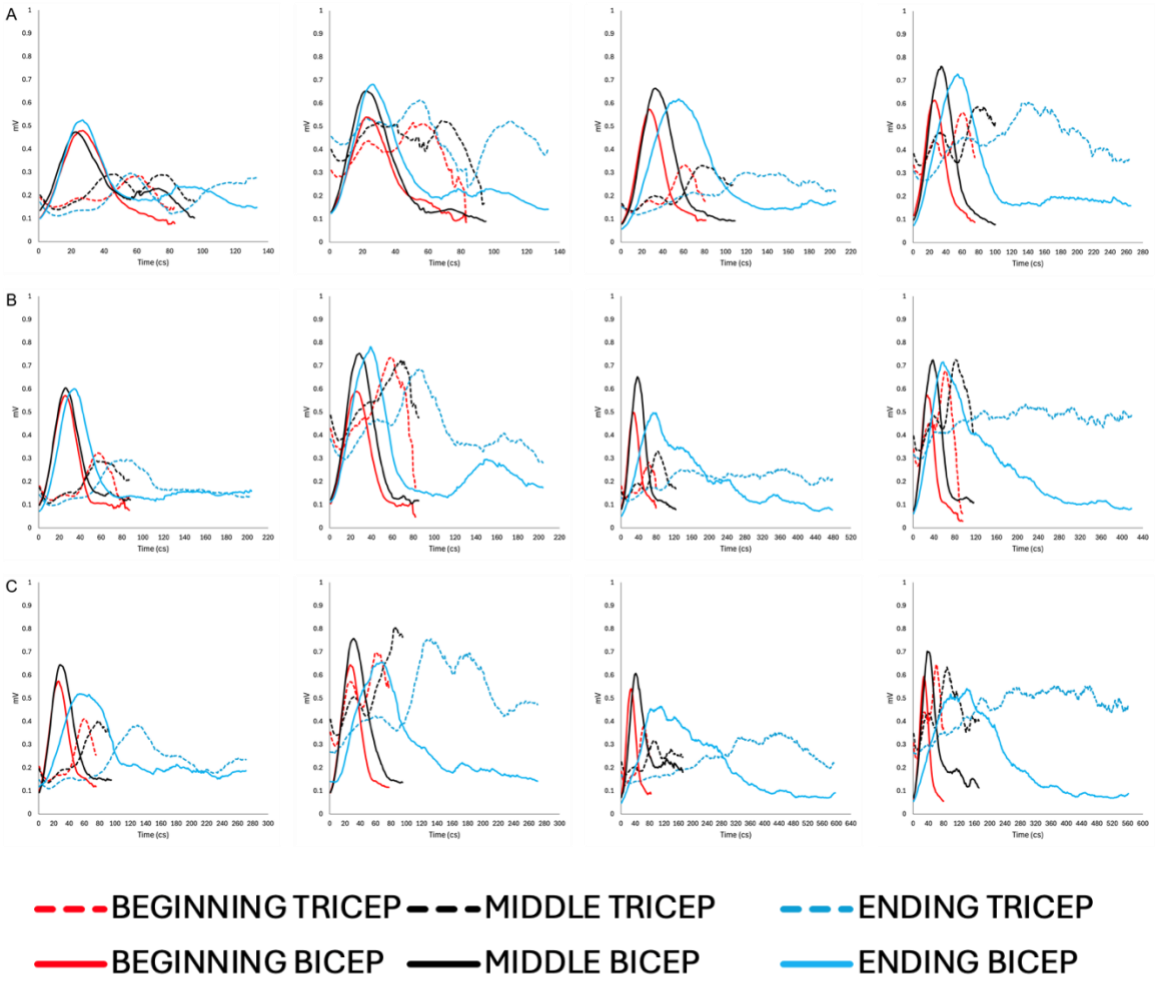

A
B
C
mV
Time (cs)
BEGINNING TRICEP
MIDDLE TRICEP
ENDING TRICEP
BEGINNING BICEP
MIDDLE BICEP
ENDING BICEP

Figure 3

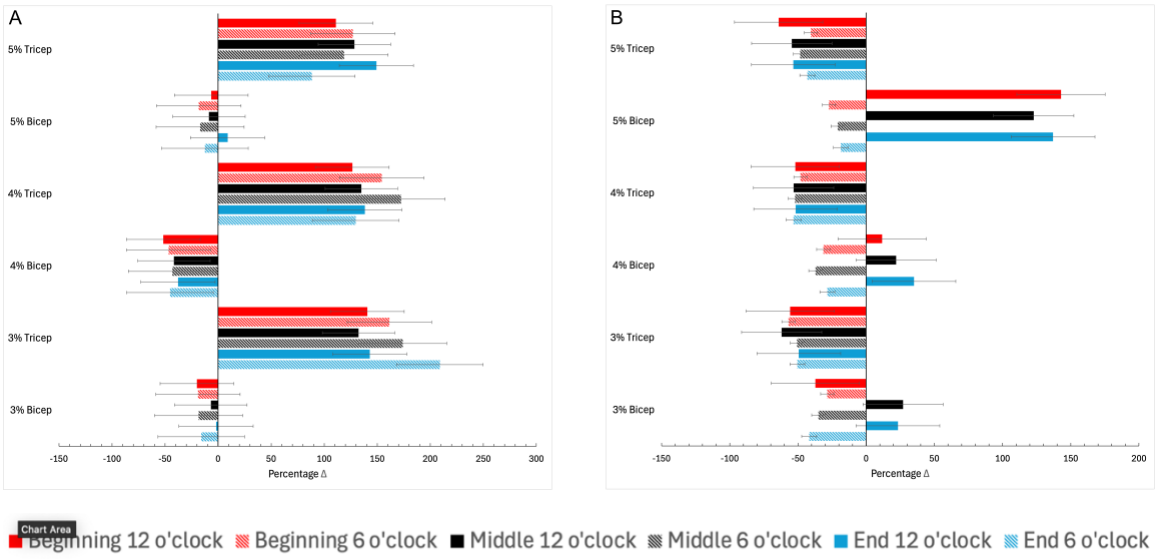
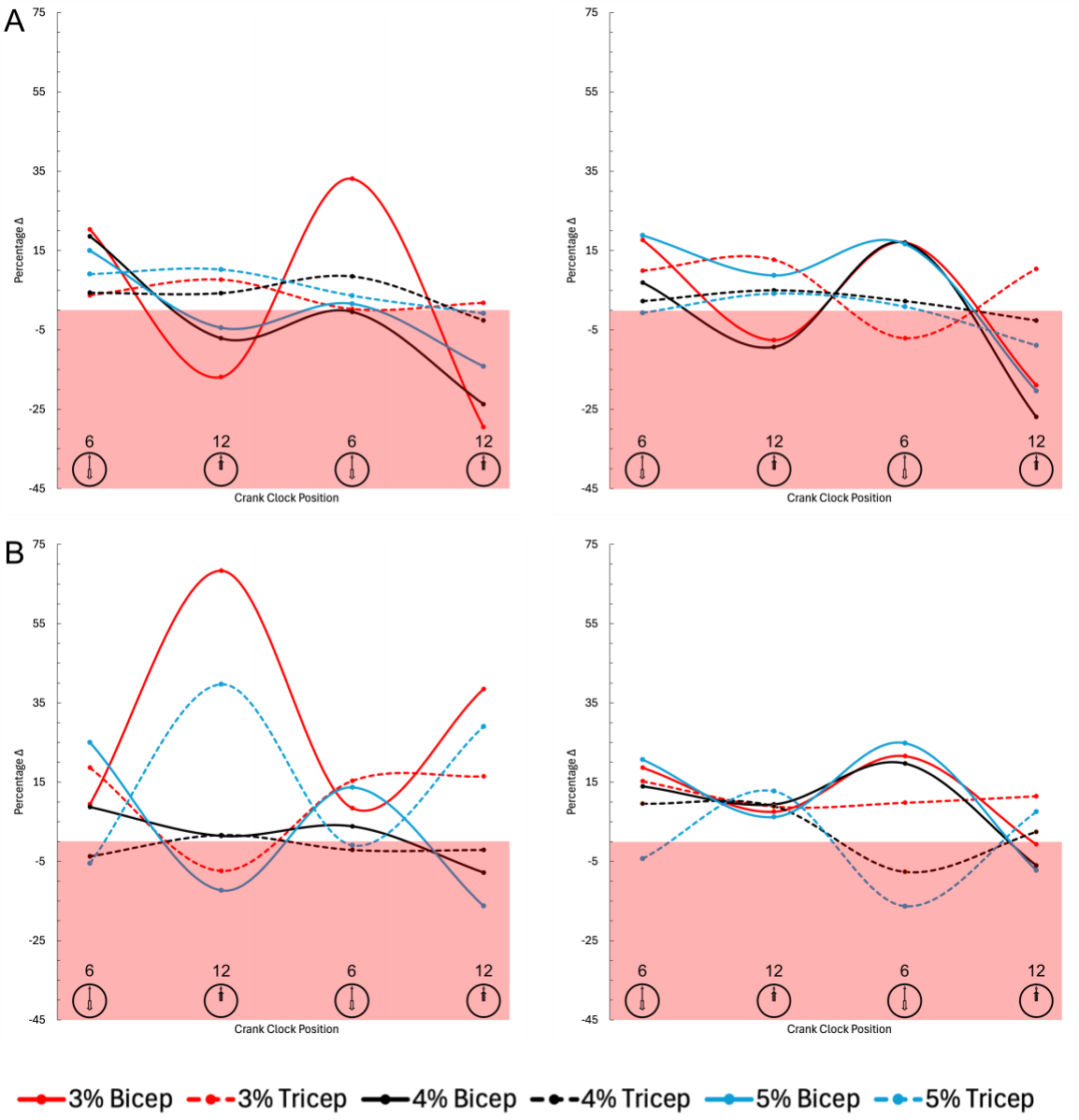

A
5% Tricep
5% Bicep
4% Tricep
4% Bicep
3% Tricep
3% Bicep
-150
-100
-50
0
50
100
150
200
250
300
Percentage Δ
B
5% Tricep
5% Bicep
4% Tricep
4% Bicep
3% Tricep
3% Bicep
-150
-100
-50
0
50
100
150
200
Percentage Δ
Chart Area
Beginning 12 o'clock
Beginning 6 o'clock
Middle 12 o'clock
Middle 6 o'clock
End 12 o'clock
End 6 o'clock

Figure 4


A
B
75
55
35
15
-5
-25
-45
Percentage Δ
6
12
6
12
Crank Clock Position
3% Bicep
3% Tricep
4% Bicep
4% Tricep
5% Bicep
5% Tricep

Appendix

Appendix 1- Ethics Approval Letter

MEMORIAL UNIVERSITY

Interdisciplinary Committee on
Ethics in Human Research (ICEHR)

St. John's, NL Canada A1C 5S7
Tel: 709 864-2561 icehr@mun.ca
www.mun.ca/research/ethics/humans/icehr

ICEHR Number:	**20250289-HK**
Approval Period:	May 27, 2024 – May 31, 2025
Funding Source:	
Responsible Faculty:	Dr. Duane Button School of Human Kinetics and Recreation
Title of Project:	*Exploring EMG activity during WinGate trials*

Title of Parent Project:	*Does a bilateral deficit exist in arm cycling and is it task dependant?*
ICEHR Number:	*20230904-HK*

May 27, 2024

Mr. Chioke Swann
School of Human Kinetics and Recreation
Memorial University

Dear Mr. Swann:

Thank you for your submission to the Interdisciplinary Committee on Ethics in Human Research (ICEHR) seeking ethical clearance for the above-named research project. The Committee has reviewed the proposal and agrees that the project is consistent with the guidelines of the *Tri-Council Policy Statement on Ethical Conduct for Research Involving Humans* (TCPS2). *Full ethics clearance* is *granted to* May 31, 2025. ICEHR approval applies to the ethical acceptability of the research, as per Article 6.3 of the *TCPS2*. Researchers are responsible for adherence to any other relevant University policies and/or funded or non-funded agreements that may be associated with the project. If funding is obtained subsequent to ethics approval, you must submit a Funding and/or Partner Change Request to ICEHR so that this ethics clearance can be linked to your award.

The *TCPS2* **requires** that you **strictly adhere to the protocol and documents as last reviewed** by ICEHR. If you need to make additions and/or modifications, you must submit an Amendment Request with a description of these changes, for the Committee's review of potential ethical issues, before they may be implemented. Submit a Personnel Change Form to add or remove project team members and/or research staff. Also, to inform ICEHR of any unanticipated occurrences, an Adverse Event Report must be submitted with an indication of how the unexpected event may affect the continuation of the project.

The *TCPS2* **requires** that you submit an Annual Update to ICEHR before May 31, 2025. If you plan to continue the project, you need to request renewal of your ethics clearance and include a brief summary on the progress of your research. When the project no longer involves contact with human participants, is completed and/or terminated, you are **required** to provide an annual update with a brief final summary and your file will be closed. All post-approval ICEHR event forms noted above must be submitted by selecting the ***Applications: Post-Review*** link on your Researcher Portal homepage. We wish you success with your research.

Yours sincerely,

James R Drover

James Drover, Ph.D.
Chair, Interdisciplinary Committee on
Ethics in Human Research

JD/bc
cc: Supervisor – Dr. Duane Button, School of Human Kinetics and Recreation