Kinetic and kinematic analysis of the upper body Wingate test using different relative loads.

By

© Angie Katherin Antolinez Romero

A thesis proposal submitted to the

School of Graduate Studies

In partial fulfillment of the requirements for the degree of

Master of Science in Kinesiology

School of human kinetics and recreation

Memorial University of Newfoundland

November 2023

St John's, Newfoundland and Labrador

Abstract

The Upper Body Wingate Anaerobic Test (WAnT) is widely used as an assessment tool in sports training, rehabilitation, and exercise physiology. This test involves participants pedalling an ergometer with their upper body against a fixed resistance for 30 seconds, providing valuable insights into anaerobic power and capacity. In recent years, researchers have turned their attention to understanding the influence of sex differences and relative load on performance outcomes in this test; however, to our knowledge, no previous study describes these differences in terms of kinetics, kinematics, and performance outcomes.

This thesis presents the results of an experiment involving 18 participants, equally divided between males and females, to investigate the presence of significant sex-related differences and the impact of relative load and crank position in the upper body WAnT. The participants were recreationally active university-students, aged between 20 and 30. All participants were familiarized with the Upper Body Wingate Anaerobic Test protocol to minimize learning effects. Each participant performed the test at three different resistances, 3, 4, and 5% of their body weight. The primary measures recorded were crank forces, video capture, and performance variables, including peak power, mean power, and fatigue index.

The results indicated significant differences in upper body anaerobic power between males and females. Males exhibited higher effective force, peak, and mean power than their female counterparts. These findings align with existing literature on sex differences in muscle mass and hormonal profiles, contributing to upper-body muscle strength and endurance disparities. Additionally, the study revealed that relative load substantially impacted test performance. Participants with higher relative loads experienced a reduction in mean power; however, this did not have the same effect for males and females. Regarding kinematics, we identified different kinematic strategies between males and females to overcome the relative load, with males using greater neck flexion and females using greater elbow and wrist

extension. This suggests that the resistance applied when scaled to an individual's body weight might not be the most accurate method to determine the relative load for males and females; this information is essential for standardizing the test across participants of varying sizes and developing precise exercise prescriptions based on relative load.

In conclusion, this thesis presents evidence of significant sex differences in upper body anaerobic power and the substantial impact of relative load on the Upper Body Wingate Anaerobic Test. These findings underscore the importance of considering sex-related disparities and relative load adjustments when using this test for assessment and training purposes.

Acknowledgments

I want to express my heartfelt appreciation to many individuals who have played pivotal roles in my research journey. Firstly, I want to extend my deepest gratitude to my friends, whose unwavering support and encouragement have constantly motivated me. Their enthusiasm and willingness to engage in countless discussions, share ideas, and provide feedback have enriched my academic pursuits.

To my partner, your boundless patience and unwavering belief in my abilities are the cornerstone of my research endeavors. Your steadfast support during the late nights and challenging moments is immeasurable. Thank you for being my rock.

To my family, your unwavering encouragement, love, and belief in my potential have been the foundation upon which my passion for research has flourished. Your sacrifices and understanding during this journey have been a trustworthy source of strength.

My gratitude extends to Dr. Button and all the mentors, teachers, and colleagues who have been instrumental in teaching me the art of research. Your guidance, knowledge, and passion for inquiry have been infectious, sparking a deep love for academia.

Lastly, to myself, for believing in my dreams, for waking up every day and continuing to walk towards my goals, for never giving up even when all odds were against me, for pursuing secondary education, for changing my family's history and for being one of the most resilient women I know.

Por siempre, gracias.

Table of contents

List of tables

List of Figures

Chapter 1: Introduction

The Wingate Anaerobic Test (WAnT) is the "gold standard" measurement of lower and upper body power production. Since its first application in 1974, the test has been applied in healthy and clinical populations with high reliability and validity for peak power measurement. However, multiple factors could change the participant's performance during the test, like the type of ergometer, age, physical condition, and sex-related differences.

The test consists of a 10-second warm-up cycling at 60 rpm with no load, followed by a 3-second countdown where the relative load is dropped, and the participant performs a 30-second all-out cycling. The relative load is calculated as a percentage of the individual's body weight. Previous research has been interested in finding the optimal relative load, defined as the load at which the person produces the higher power output; however, no previous study explored concomitant kinetic and kinematic changes during the WAnT at different relative loads.

This study explores how the relative load can affect the forces exerted on the upper body during the WAnT. Since the load increases relative to body weight, the forces required to move the upper body and perform the test increase accordingly, which could result in higher mechanical stress on the upper body joints and muscles. Regarding kinematics, the relative load can influence the movement patterns of the upper body. Individuals may adopt strategies to overcome the increased load, such as altering their posture, range of motion, or movement speed, impacting the joint angles, velocities, and accelerations during the test.

Intending to describe the effect of the relative load and the possible sex-related effects in the kinetics and kinematics of the Wingate test, this thesis contains five chapters, each containing various sub-sections. Chapter 1 is the introductory chapter that includes the scope of this thesis and background information on the WAnT. Chapter 2 reviews the literature, including the history, protocol, previous kinetic and kinematic studies, and optimal load research on the upper body Wingate test. Chapter 4 (i.e., the manuscript) describes the methodology, including a kinematic approach of Markerless Motion Capture Analysis and power and force production measurements, summarizes the results and description of the effects of the relative load, explains the conclusions gathered from this project, and gives recommendations for future considerations when choosing the relative load of the upper body WAnT.

Purpose of the study

The primary goals of this study are to 1) describe the effect of the relative load on the kinetics and kinematics of the upper body WAnT and 2) determine if goal 1 is sex-dependent.

Hypothesis

The three hypotheses for this study are as follows:

1. Higher relative loads will lead to higher crank forces and power production.
2. Higher relative loads will impact the upper body's kinematics, such as increased joint angles in the upper extremity and decreased relative velocity.
3. The kinetics, kinematics, and performance during the upper body WAnT will be sex-dependent.

Chapter 2: Literature Review

This literature review takes a multifactorial approach to understanding the Wingate anaerobic test (WAnT). The first part of this review consists of describing the history, creation, and variations of the test, followed by a review of the physiological mechanisms involved in muscle power production that include understanding factors like the generation of maximum force, rate of force development, neural coordination, activation-relaxation Kinetics, maximum shortening velocity, fatigue resistance, and muscle oxidative capacity.

In the second part, we described the kinematic variables of the upper body WAnT, such as the range of movement of the upper extremity, movement pattern, and the influence of the motion path in arm cycling, as well as the available tools to measure this variable such as markerless motion capture. Regarding Kinetics, we described the forces that interact between the ergometer and the person when performing arm cycling. While most of the literature has focused on the impact of different relative loads on performance measurements like peak or mean power, fewer studies have described crank forces in arm cycling, especially during the upper body WAnT.

The final part of this review emphasizes the relative load effect and the sex differences in upper body power production. It is well known that anatomical and physiological factors allow men to produce more power with their upper body than females; however, since there is no consensus on the relative optimal load for each sex, it is challenging to compare their power production. Other factors, such as the biomechanics of arm cycling WAnT, might differ between males and females but, to our knowledge, have not been explored.

2.1 The Wingate anaerobic test

2.1.1 History.

The WAnT was designed by the Department of Research and Sports Medicine of the Wingate Institute of Physical Education and Sports in Israel during the 1970s (1). The development of the WAnT started in 1974 with a study that compared measurements of explosive strength and anaerobic power (2). This study looked at the relationship between 5 different tests: two tests explored explosive strength with a single all-out effort over 150 degrees of motion to measure the power of the lower limb in the propulsive phase of cycling. The other two tests consisted of 30 seconds of pedalling test measurements, a lower body one with a relative load of 40g per kg of body weight (BW) and the upper body with a relative load of 20g per kg of BW, and the Margarita step test. The results of this study showed that the 30-second pedalling test and the explosive power test had higher reliability (r=0.93) among all the anaerobic power measurements.

Bar-Or (1) published an update on the test's methodology, reliability, and validity factors. He described the test as a 30-second pedalling or arm-cranking task at maximal speed against a constant force that must induce a noticeable development of fatigue. The three initial indices measured were peak power (highest mechanical power produced), mean power (average power sustained over 30 seconds), and the rate of fatigue (degree of power drop-off from start to finish). The test was designed to be administered in the mechanical ergometer of preference, with a conventional pedal crank length of 17.5cm and an optimal load of 0.75 kp/kg. The optimal load was defined as the individual force setting that would elicit the highest peak and mean power. Additional considerations for the leg pedalling were toe stirrups and, for arm cranking, a cylindrical grip with no additional attachments.

The WAnT is considered the gold standard for measuring power production in the lower and upper body primarily due to its high reliability and validity, well-established protocols, and standardization of testing procedures (3). Several factors contribute to its reliability; the test is conducted in a controlled environment, typically in a laboratory setting, where variables like temperature, humidity, and air resistance are kept constant, reducing the impact of external factors on the results.

This test has been extensively studied and validated with various individuals, including athletes from various sports, fitness, functional levels, and different age groups (4–7). The WAnT has shown a high correlation with other measures of anaerobic power, such as peak power, mean power, and fatigue index, strengthening its validity as an accurate assessment of muscle power production. Overall, the reliability of the WAnT makes it a trusted and widely accepted tool for evaluating an individual's capacity for rapid and intense muscle power production, providing valuable insights for training and performance optimization.

2.1.2 Upper vs. lower body WAnT

The WAnT can be conducted using either the upper body (e.g., arm crank ergometer) or the lower body (e.g., cycle ergometer) to assess anaerobic power and capacity (1). While the fundamental principles of the test remain the same, there are some differences between the upper body and lower body Wingate tests, like the type of muscles engaged, the relative power output, the fatigue factor, and the specificity for different sports (8–10). The primary difference lies in the muscles being engaged during the test. In the lower body Wingate test, the main muscle groups involved are the quadriceps, hamstrings and gluteus maximus, as the individual pedals on a stationary cycle ergometer. It is important to mention that the upper body is also active during the lower body WAnT providing stability to the participant by grasping the handles. In contrast, the upper body Wingate test primarily engages the arms, shoulders, and chest muscles as the individual performs the test on an arm crank ergometer with the legs and feet of the participants strapped in place to provide stability (11). Due to the larger muscle mass and greater force-generating capacity of the lower-body muscles, the lower-body Wingate test yields higher peak and mean power outputs compared to the upper-body Wingate test. As a result, the upper body Wingate test may be more suitable for individuals with lower body injuries or conditions that limit lower body function (12).

The fatigue pattern in the upper body Wingate test may differ from the lower body test due to variations in muscle fibre recruitment and energy systems. For instance, the upper body may experience earlier fatigue due to the reliance on smaller muscle groups, while the lower body may sustain power output for a longer duration due to the involvement of larger muscle groups. (13). However, choosing between upper body and lower body Wingate tests should consider the individual's sport-specific requirements or functionality level. For example, athletes in rowing, wheelchair racing, or combat sports may benefit from the upper body Wingate test, while athletes in sprinting or cycling events may find the lower body Wingate test more relevant (14,15).

2.1.3 How does the WAnT measure power production?

Muscle power production primarily depends on the contractile mechanisms within skeletal muscle fibres. The sliding filament theory, proposed by Huxley and Niedergerke in 1954 and refined by Huxley and Hanson in 1957, remains a foundational concept in muscle physiology (16). According to this theory, the interaction between actin and myosin filaments within the sarcomeres leads to muscle contraction, generating force and power (17). The rate of force development and the extent of force produced is determined by multiple factors, such as the type and arrangement of muscle fibres, motor unit recruitment, and the frequency of neural stimulation (Table 1). In addition, the size and architecture of muscles play pivotal roles in muscle power production, with pennate muscles, for example, exhibiting greater power output than parallel muscles due to their increased physiological cross-sectional area (18).

Maximum Force	- Muscle CSA - MyHC IIa/x area ratio - MyHC IIa/x composition (positive effect) - Neural drive - Muscle architecture
Rate of force development	- Maximum force - Neural drive MyHC IIa/x area ratio - MyHC IIa/x composition (positive effect) - MTU stiffness
Neural coordination	-Magnitude of muscle activation

	-Timing of muscle activation and relaxation
Activation-relaxation kinetics	- MyHC IIa/x composition (positive effect) -Sarcoplasmic reticulum structure and function
Maximum shortening velocity	- MyHC IIa/x composition (positive effect) - Muscle architecture
Fatigue resistance	- MyHC IIa/x composition (negative effect) -Anaerobic substrate availability and enzyme activity -Metabolite buffering capacity -Pain tolerance
Muscle oxidative capacity.	- Muscle CSA (negative effect) - MyHC IIa/x composition (negative effect) - Mitochondrial and capillary density -Oxidative enzyme activity

Table 1. Summary of physiological determinants of power production. CSA cross-sectional area, MTU muscle-tendon unit, MyHC myosin heavy chain isoform Modified from (19)

Muscle power production also depends on efficient energy systems rapidly supplying adenosine triphosphate (ATP). The primary energy systems in power production are phosphagen, glycolysis, and oxidative phosphorylation. The phosphagen system, dominated by creatine phosphate, provides rapid but limited ATP for short bursts of high-intensity activities. Glycolysis offers a moderate ATP supply but is limited by the accumulation of lactic acid, leading to fatigue. Conversely, oxidative phosphorylation occurs in mitochondria and is the most sustainable energy system, producing ATP by oxidizing carbohydrates and fats. However, this system has a slower rate of ATP production, making it more suitable for endurance rather than explosive power activities (20). The coordination and balance of these energy systems are crucial for optimizing muscle power production during high-exertion physical tasks.

The WAnT assesses power production, which can be described through the force-velocity and power-velocity relationships (Figure 1); this test primarily engages the anaerobic energy systems to meet the high demands of intense exercise (21). At the start of the test, the phosphagen system provides rapid energy through the breakdown of stored creatine phosphate to regenerate ATP (ATP-PCr system). This immediate energy source supports the explosive burst of power needed to initiate pedalling on the stationary cycle ergometer. As the test progresses, the glycolytic system becomes more prominent, breaking down stored glycogen into glucose to produce ATP anaerobically. However, this energy pathway is less efficient and results in the accumulation of hydrogen ions, contributing to the onset of fatigue (22).

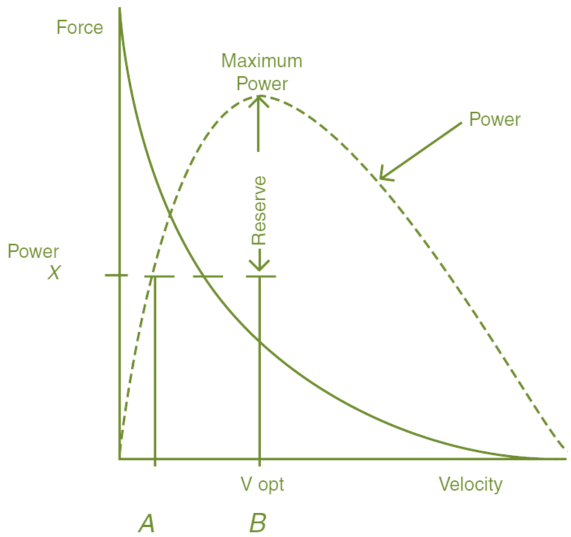

Figure 1. Force-velocity and power-velocity relationship. At a power requirement, X only 50% of the muscle power-generating capability is needed at optimum velocity (point B)(21)

During the 30-second all-out effort, oxygen demand exceeds oxygen supply, and the body's inability to meet the energy demands aerobically relies on anaerobic processes to sustain power production. After the test is over, there is a phenomenon called excess post-exercise oxygen consumption, characterized by the oxygen consumption remains elevated, facilitating the clearance of accumulated lactic acid associated with muscle fatigue and providing valuable information about an individual's anaerobic capacity and tolerance to high levels of muscle acidity (23). The intensity of the Wingate Test also leads to a significant depletion of phosphocreatine, a crucial component of the phosphagen energy system. The rate at which phosphocreatine is depleted during the test provides insights into an individual's immediate anaerobic energy availability and capacity for rapid power generation (24).

2.2 Performance variables of the WAnT

Multiple variables can be obtained from the test as a reliable and valid tool to measure anaerobic performance (3). The primary information that can be extracted from the test is as follows:

- Peak Power (PP): Peak power is the highest power output achieved during the 30-second test, usually achieved within the first 5 to 10 seconds of the test. It represents the participant's maximum anaerobic power capacity and is typically expressed in watts (W) or watts per kilogram of body weight (W/kg), also called Anaerobic Power. This is an indirect measurement of the power generation capacity of the ATP-PCr system.
- Mean Power (MP): Mean power is the average power output over the 30-second test. It indicates the participant's sustainable anaerobic power output during a brief high-intensity effort and is also expressed in watts (W) or watts per kilogram of body weight (W/kg). This is a measurement of the energy production capacity of the represents the capacity of ATP-PCr and glycolytic system.
- Fatigue Index (FI): The fatigue index quantifies the decline in power output during the test, indicating the rate at which fatigue occurs. It is expressed as a percentage and is calculated using the following formula:

$$FI=\frac{(PeakPower-LowestPower)}{PeakPower}\times 100$$

- Total Work (TW): The total work is the cumulative energy expenditure during the 30-second test. It represents the total amount of mechanical work performed during the test and is calculated by integrating the instantaneous power output (PO) over time, which represents the area under the power-time curve. The formula for calculating the total work done is as follows:

$$TW=\int_0^{30} PO(t)dt$$

- Anaerobic Capacity: This index is calculated by dividing the mean power by body weight. The units are watts per kilogram; however, the scientific community has not accepted this index because of its low reliability (23).

These performance variables are crucial in assessing anaerobic capacity and power during short-duration, high-intensity efforts (21). The WAnT is widely used in various fields, including sports science, exercise physiology, rehabilitation set-ups, and athletic training, to evaluate anaerobic performance and guide training and rehabilitation programs. Proper interpretation of these variables can help to optimize training strategies, track performance progress, and make informed decisions related to exercise, performance, and functional goals.

2.3 Kinematics of arm cycling

While the physiological performance determinants of recumbent arm cycling have been investigated extensively, little is known about arm cycling biomechanics or arm cycling configuration. Understanding the kinematics of arm cycling can provide valuable insights into optimizing performance, injury prevention, and enhancing overall efficiency during arm-powered activities. The key kinematic parameters in arm cycling include the joint angles at the shoulder, elbow, and wrist joints; these angles determine the arm's position relative to the body and the cycling equipment, affecting the overall range of motion and efficiency during the movement (25).

The study of arm movement patterns helps identify the most effective and efficient techniques for generating power during cycling and changes in the cadence, typically measured in revolutions per minute (rpm) (26). It is important to note that the specific kinematics of arm cycling may vary depending on the type of activity, equipment used, and individual biomechanical characteristics. As with any physical activity, proper form, technique, and training play a significant role in enhancing performance and preventing injuries during arm cycling, where variables like arm synchronization and arm positioning will play a fundamental role (27).

Arm cycling involves repetitive movements that can be categorized into two different phases, such as the pushing phase (downstroke) and pulling phase (upstroke), two propulsive moments called top dead centre (or 12 o'clock position) and bottom dead centre (or 6 o'clock position) (see Figure *2*). Those phases are relevant to characterizing the optimal performance of cycling (28,29). The 12 o'clock position begins the pushing phase, the primary propulsive stage of arm cycling. It involves exerting maximal force through the arms and shoulders, specifically activating the triceps brachii and the deltoid anterior, to move the cycling equipment forward (30). In this phase, most of the power is generated, so proper biomechanics, arm positioning, and coordination of muscle groups are essential to ensure efficient force production and minimize energy loss.

The 6 o'clock position is the start of the pulling phase. In this phase, the arms continue their motion backward, pulling through the cycle to complete the movement. This phase assists in maintaining momentum and sets up for the next pushing phase. The main muscles active during this phase are the biceps brachii and the posterior deltoid (31). The pulling phase is essential for maintaining a smooth and continuous arm cycling motion, which can help reduce joint and muscle stress and promote a more fluid movement pattern(32).

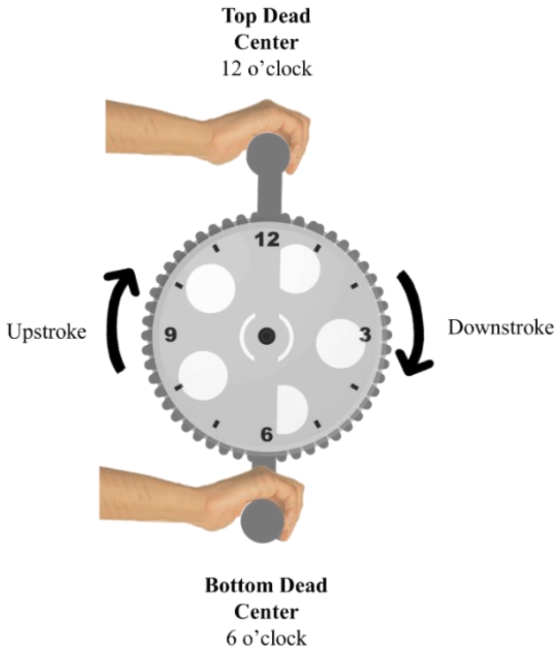

Figure 2. Phases of cycling. The 12 o'clock position begins the downstroke phase, also known as the pushing phase. At the 6 o'clock position, the start of the upstroke phase is also known as the pulling phase.

To our knowledge, no previous study has evaluated the biomechanics of the WAnT in the lower or upper body. However, the kinematic patterns in cycling have been of great interest, especially for high-performance athletes (19,29,33). The measuring of kinematic variables in sports sciences and also in rehabilitation sciences has been conducted using mainly optical marked systems such as 3D Vicon, the gold standard for motion capture systems; however, with the development of artificial intelligence, there has been an interest in markerless motion capture systems for their promising features like high accuracy, flexibility for indoor and outdoor applications, low cost, and versatility during different motor tasks. (34,35).

2.3.1 Markerless motion capture.

There are multiple optical and non-optical methods to perform kinematic analysis. Motion capture technology has revolutionized the field of kinematic analysis by providing valuable insights into human and animal movement (36). Traditional marker-based motion capture systems require attaching reflective markers to the subject's body, which can be cumbersome and time-consuming. Marker-based motion capture systems have been widely used in biomechanics, neuroscience, robotics, and sports sciences to study human and animal movements. While these systems have proven valuable, they have certain limitations. Placing markers on subjects can disrupt natural movements and may not be suitable for certain species or contexts. Marker occlusion and drift are common challenges leading to data inaccuracies (37). However, recent advancements in computer vision and machine learning have led to the development of markerless motion capture techniques, eliminating the need for physical markers (38). Markerless motion capture emerged as a solution to these drawbacks, allowing researchers to analyze movements more naturally and unobtrusively. It relies on computer vision algorithms and deep learning techniques to track critical points on the subject's body without any markers (39).

Deeplabcut, developed by Mathis et al. (40), is a state-of-the-art markerless motion capture software that has gained immense popularity in the scientific community. The tool is built upon deep neural networks, specifically convolutional neural networks (CNNs), which excel in identifying and tracking complex patterns in images and videos. The process involves a two-step pipeline: training and using the network to infer new videos. There are multiple advantages to using Deeplabcut. The deep learning algorithms allow for robust and precise tracking of anatomical landmarks, achieving accuracy comparable to or surpassing traditional marker-based systems. The model learns from the training data and generalizes its knowledge to new, unseen movements, resulting in reliable kinematic analyses. Initially, the system was created to track animal behaviour; however, its flexibility and adaptability allowed it to be used with various species, body shapes, and environments.

Deeplabcut is also an open-source algorithm, fostering collaboration and continuous improvement from different scientific communities. The community actively contributes to its development, sharing insights, updates, and optimized models, making it a dynamic tool for researchers worldwide (https://github.com/DeepLabCut/DeepLabCut). No previous study has implemented the use of Deeplabcut on arm cycling. Its potential is attractive since marker obstruction is one of the most common challenges when tracking human-object interactions, and this is not a limitation with this type of technology since the labelling is performed manually.

The analysis of kinematic factors is essential to describe characteristics like joint angles, range of motion, movement variability, and velocity for each body part involved in the motor task. However, they are not sufficient to understand the complexity of human movement. Kinetic factors are often measured to describe changes in the interaction between the subject and object.

2.4 Kinetics of arm cycling.

Another area of interest in studying cycling is understanding the forces that move most efficiently and effectively. In cycling, the attachment between the cyclist and the bicycle is defined as a three-point link involving the handlebars, saddle, and pedals; in the case of arm cycling, that system is composed of the individual, the cranks, and the chair or point of contact with the ground. These links serve as force transfer points from the body to the bicycle and viceversa (41). The main force studied in cycling is the pedal or crank force in arm cycling.

Measuring pedal forces has interested researchers for the last 30 years. Technology has allowed pedal force measurement to advance to the stage where it is possible to measure three components of force (Fx, Fy, and Fz) and three associated moments (Mx, My, and Mz). Tools like angular potentiometers (42), videography (43), and digital encoders (45) have been used to measure and calculate the effective force during cycling. The ratio of the force perpendicular to the crank (effective force) to the overall force applied to the pedal (resultant force) has been used to define the effectiveness of pedal force during cycling (see **Figure *3***). The pedal force has been measured with mainly strain gauges, and since the sagittal plane is the one that contributes mainly to the tasks, the forces tend to be calculated just in this plane.

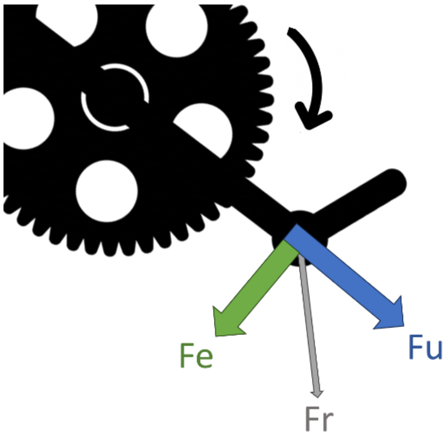

Figure 3. Visual representation of forces acting in the arm crank. Fe = effective force Fr = resultant force Fu: unused force.

In arm cycling, no study has explored the forces applied in the crank, creating a knowledge gap in the study of force efficiency and a lack of understanding of how the forces are produced by the person performing the task. This is particularly true in a power production task like the upper body WAnT. Arm crank ergometry has been studied to explore what type of crank facilitates more power generation or if the crank configuration, synchronous versus asynchronous, affects torque production. Smith et al. (46) investigated the effect of crank configuration on muscle activity and torque production during submaximal arm crank ergometry. They found that patterns of torque production during asynchronous and synchronous cranking were distinct and different during different workloads (50 vs. 100 watts).

Another essential force is muscle force. In (47), Bressel et al. studied the effect of different handgrip positions (supinated, pronated, and neutral) on the neuromuscular activity of the biceps brachii, triceps brachii, middle deltoid, infraspinatus, and brachioradialis. They found that the biceps brachii has a 63% greater activation in a neutral handgrip, and the infraspinatus exhibited 36% less activation in a supinated vs a neutral position. Another study in 2015 explored the muscle coordination of the upper body while cycling at 150 and 300 watts with a cadence of 90 RPM; they found that the EMG activation patterns remained similar; however, increased workloads showed an increased structure in the EMG that most likely represents a more orderly recruited pattern during the cycling task at higher effort levels (48). The most recent study that classified muscle activation during an arm cycling task was by Chaytor et al.(31). They found that integrated electromyography (EMG) amplitude increased significantly with increased power outputs and a difference in iEMG between the flexion and extension phases in all muscles except the triceps brachii and extensor carpi radialis.

The joint torques are essential when combining the study of muscle and crank forces. Joint torque refers to the mechanical stress the joint undergoes when performing a task. The shoulder joint is the central pivot point in the joint kinematics of arm cycling. During the pushing phase, the shoulder abductors and flexors generate torque to push the hand cranks forward, and during the pulling phase, the shoulder extensors and adductors work to pull the hand cranks backward. The second main joint is the elbow joint. The elbow joint produces the main torque during the pushing phase via the elbow extensors and during the pulling via the elbow flexors. Finally, as the point of transference of the forces to the crank, the wrist joint contributes to arm cycling by providing stability and control. The wrist extensors and flexors may generate torque during the downstroke and upstroke, respectively (49). It is essential also to consider that other joints like the neck or the trunk have not been included in the analysis of upper body cycling; however, their proximity and high functional correlation might give insight into a more holistic approach to the kinematics of arm cycling.

Another factor considered in a kinematic analysis is joint stiffness, which refers to a joint's resistance to movement. Dynamic joint stiffness is thought to be influenced by a combination of active muscle contraction and passive soft-tissue resistance. In the context of the upper body WAnT, joint stiffness plays a role in optimizing power transfer and minimizing energy losses during the pedalling motion. Proper joint stiffness allows efficient force transmission from the muscles to the hand cranks. If joint stiffness is too low, excessive energy may be lost as the joints "give" during the pedal stroke. On the other hand, if joint stiffness is too high, it can lead to increased fatigue and muscle stress (50).

Individuals must balance joint flexibility and stiffness to maximize upper body WAnT performance. Proper joint coordination and stability are critical for generating and transferring power effectively. Measuring joint torques and stiffness directly during the upper body Wingate test can be challenging, as it may require specialized equipment and invasive procedures. However, researchers and sports scientists can use motion capture systems, dynamometers, and surface EMG to infer joint torques and assess muscle activation patterns during the test.

The study of combined kinetic and kinematic variables of the upper body WAnT has not been explored yet, but it has been done for lower body WAnT. **Figure *4*** represents how the kinetic and kinematic factors interact in a power production task. First, the power production has a hyperbolic behaviour regarding the velocity, and there is an optimal point for power generation around 120 rpm. Conversely, torque almost has an inverse linear relationship with velocity, as the force-velocity relationship curve illustrates in Figure 4, also, as time passes during the test, the muscle reduces its ability to produce torque because of a fatigue effect. The capacity to generate maximal force, or a very high submaximal force, is limited to medium velocities because of the energy depletion it takes to increase the velocity to high RPMs, leading to a detrimental effect on power and force production over time.

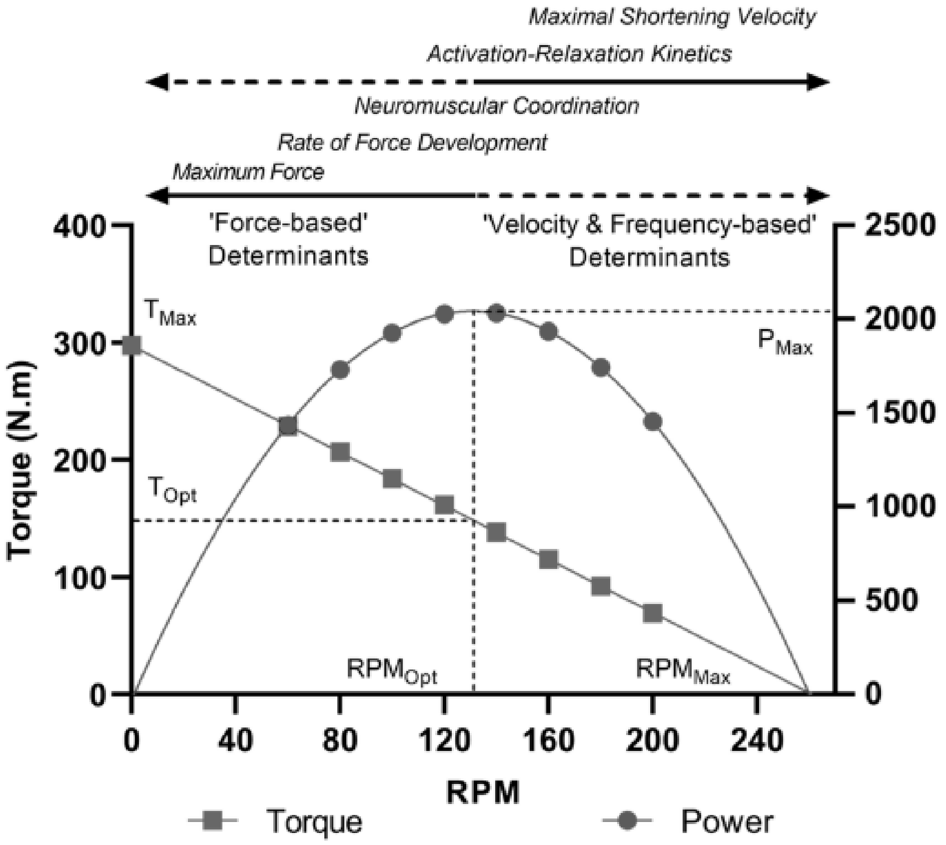

Figure 4. The torque- and power-pedalling rate relationship, parameters, and determinants.: PMax, maximum power; RPMMax, maximum pedalling rate; RPMOpt, optimum pedalling rate (i.e., optimal frequency); TMax, maximum torque; TOpt, optimum torque. From (19)

In conclusion, power output can be improved by modifying the arm cycling task technique to a more circling action (i.e., greater force effectiveness for similar magnitude of forces). The performance will be potentiated from a mechanical perspective if the pedal forces are applied perfectly perpendicular to the crank in the direction of crank motion (i.e., 100% of effective force). However, this effect requires a perfect kinematic alignment; therefore, the kinetic and kinematic factors should not be studied separately due to the interdependence for improving performance.

It is important to consider that in the WAnT, the forces applied by the participant depend on the resistance chosen to perform the test. This resistance has been named relative load, and a percentage of body weight is the most commonly used factor to determine it (51).

2.5 Relative load in arm cycling

The use of arm cycling from rehabilitation to high-performance populations has led to multiple questions relating to prescription parameters. Thus, one of the significant parameters includes optimal load and optimal load force to maximize performance. A systematic review interested in the factors that influence bicycle fitting, defined as the detailed process of evaluating the cyclist's requirements to adjust the bicycle to meet the cyclist's goals and needs, found that the dynamic bicycle fitting methods need to consider intensity (i.e., relative load) as a fundamental factor (33).

The effect of the relative load has been explored in the upper and lower body and has mainly focused on neuromuscular/neuromechanical performance. For example, Holliday et al. (29) studied the effect of different relative loads of cycling training and racing on whole-body joint kinematics and muscle activity. They found significant changes in the kinematic parameters of ankle dorsiflexion, knee extension, elbow flexion, and trunk orientation, with more than 30% difference between low and high intensities. In muscle activation, there was a higher EMG signal amplitude in all low-extremity muscles at higher relative loads.

Botzheim et al. (32) described the effects of body position with respect to gravity (sitting and supine), crank length (10cm vs. 15cm), and cranking mode (synchronous vs. asynchronous) on muscle synergies in arm cycling. EMG was recorded from the main muscles activated during arm cycling, and their results showed that synergies were affected by body position and cranking mode, suggesting that the central nervous system may employ different motor control strategies in response to external constraints.

Most of the research developed in understanding arm cycling and the effect of different relative loads has been developed in neurophysiology. Forman et al. (2015) found that phase and cadence impacted corticospinal excitability (CSE) in elbow flexion and extension during asynchronous arm cycling (52). Spence et al. (53) compared CSE at 5% and 15% of peak power output (PPO), finding that the CSE is load-dependent, increasing the excitability as the load increases in the biceps and triceps brachii. Another study that compared different PPOs (5-50%) found that iEMG is influenced by the flexion/extension cycling phases and also showed a linear relationship between iEMG amplitude and the %PPO for all muscles (31). In addition to the cadence, phase, and load dependence factors, it was also shown that the direction of the cycling (forward vs. backward) significantly impacts the CSE of the biceps brachii, being higher during forward cycling compared with backward cycling (54).

Less literature studies the effect of the relative load on the kinematics of arm cycling. Mravcsik et al. (32) developed a study to determine the effects of different relative loads on the kinematics of arm cycling. Specifically, they wanted to determine the effect of three different crank resistances (1.16 Nm, 3.09 Nm, and 6.14 Nm) on arm configuration and muscle activation variances during arm cycling. The range of angular motion of the shoulder, elbow, and wrist was 42.75°±0.63, 68.87° ±0.49, and 23.35° ±1.16 (mean ±SEM) during the three resistances. The study reported no significant differences between the crank resistances regarding the muscle activation variances. The resistance used in their study did not consider body weight as a factor to determine the relative loads, and the task did not require high intensities and velocities as needed during the upper body WAnT.

The denominated kinematic-force-control theories have studied the interest in forces and motion patterns in different motor outputs (55). These theories try to explain the mechanisms and implications of the central nervous system to control movement and force (most studies have only explored isometric tasks). Piovesan et al. (56) studied the control of motion and contact forces in dynamic motor output. They found that the force regulation process employed by the central nervous system involves not only the pure force controller but also a coupled motion controller mediated via the mechanical impedance of the arm. Also, as an adaptation to the environment and the task, the person creates an internal environment model that estimates the kinematic parameters and drives the coupled force-motion controllers to provide increasingly more accurate feed-forward compensation. However, the exact mechanism of force-motion control, especially for power production tasks in dynamic motor outputs, is still under study.

In a dynamic motor output like arm cycling and particularly in power production tasks like the WAnT, there is a knowledge gap on the effect of relative load on the biomechanical characteristic of the movement itself. Filling this gap could help us understand the relevance of force efficiency and optimal motor activation patterns when performing the WAnT.

2.5.1 Impact of relative load in WAnT.

The original test weight/load used for the WAT was 7.5% of the participant's body mass (BM) or 0.075 kg per kg of body weight; however, this recommendation was based on a study of a small number of untrained young people. The best way to empirically answer the WAnT ideal strength is to repeat the test with varying loads on different populations (57). Selecting a force load for each subject that yields both optimal mean and peak power outputs is essential.

The initial investigation into the proper resistance for the WAnT was conducted by Evans et al. (58) on 12 highly trained individuals. They concluded that the WAnT resistance must be determined by the subject's anthropometric measures (weight and leg volume). Dotan et al. suggested a resistance of 0.087 kp/kg for active male students, arguing that the resistance should be set as a function of physical fitness. Additionally, they confirmed that the ideal resistance for peak power was higher than required for maximal mean power.

After this study, Bar-Or published an update on the methodology, reliability, and validity of the WAnT for the upper and lower body in different populations and established the relative load values for active adults, as shown in Table 2. They also established that choosing the ideal force based on total body weight may not be the most accurate strategy (muscle mass or fat-free weight, for example, may be superior options). However, using body weight as a criterion appears sensible in terms of application (1).

Subjects	Upper/Lower body	Sex	Optimal load (kp/kg)
Physical education students	Upper body	Male	0.087
		Female	0.085
	Lower body	Male	0.062
		Female	0.048

Table 2. References values for optimal load for upper and lower body WAnT in active males and females. Modified from (1)

Forbes et al. (59) conducted a study with 40 participants (20 males and 20 females) divided into recreationally active and trained participants. They performed multiple upper body WAnT with relative load factors ranging from 0.050 to 0.080 kg $\cdot$ kg^{-1} BM for recreational males, 0.060–0.090 kg $\cdot$ kg^{-1} BM for trained males, and 0.040–0.070 kg $\cdot$ kg^{-1} BM for both recreational and trained females. The results showed that the highest peak (5s) power outputs for trained males, recreational males, trained females, and recreational females were 0.075, 0.070, 0.065, and 0.060 kg $\cdot$ kg^{-1} BM, respectively. The relative load factors that elicited the highest mean 30s power outputs for trained males, recreational males, trained females, and recreational females were 0.075, 0.055, 0.050, and 0.050 kg $\cdot$ kg $^{-1}$ BM (59).

More recently, a study by Michalik et al. (60) compared two methods to determine the optimal load for the WAnT: the force-velocity relationship, the power-velocity relationship, and the body weight model. The participants performed nine all-out lower body sprints against a relative load of 3 to 14 kg. They found that peak power output was similar between tested models, but the study did not describe factors like velocity or force production (60). Thus, the relative load used for WAnT potentially impacts WAnT performance. Selecting an optimal relative load resistance considering applied forces and joint kinematics of an upper body cycling WAnT could create a novel and powerful approach.

2.6 Sex differences in power production

As mentioned at the beginning of this literature review, power production depends on the type and arrangement of muscle fibres, motor unit recruitment, and neural stimulation frequency. Previous studies have identified over 3,000 genes expressed differently in male and female skeletal muscle. Those genetic differences are expressed in different skeletal muscle fibre composition, myosin heavy chain expression, contractile function, and the hormonal regulation of these physiological differences (18).

A study that included 55 women and 95 men to study the composition of the vastus lateralis found that type-I fibres account for 36% of the total biopsy area in men and 44% in women, and type-IIA fibres account for 41% in men and only 34% in women. The greater prevalence of slower type fibres in females compared with males parallels the lower contractile velocity in females compared with males. Also, men's muscle fibres have significantly larger cross-sectional areas (CSA) than women's (61). The capacity of fatigue of a muscle has also been studied during isometric contractions, and it was found that male muscles are more fatigable than female muscles, and it is related to muscle substrate utilization, neuromuscular activation, and muscle morphology (62).

These sex differences in force and power production are relevant to deciding parameters like the relative load during anaerobic testing and performance reference values to determine if the person has a high, medium, or low power production. Because of the anatomic and physiological muscle differences between males and females, it would make sense for females to have a lower peak power and a low fatigue index since male muscles can produce high power quickly but are more fatigable.

2.6.1 Sex differences in the WAnT

One of the first research publications comparing performance during the lower body WAnT and muscle morphology in males and females was by Froese et al. (63). They had 30 participants (18 females and 12 males) perform a 30-second lower body WAnT, and then blood samples were taken to measure lactate levels as well as biopsies of the vastus lateralis on the right leg following the WAnT. There was a significative positive correlation between the performance variables of peak power and total work, post-test lactate concentration, and the percentage and area of fast twitch fibres for men. However, no significant correlations were found for females. One of the possible explanations for the lack of correlation between variables for women is that the method used to calculate the optimal load was a regression equation incorporating body mass (kg) and leg volume (L) determined by water displacement. Since women tend to have a more significant proportion of body fat and lower muscle mass than men, this led to assigning loads that were likely too high in terms of active muscle mass for the female subjects.

Multiple studies have used different relative loads and compared the performance of males versus females in the upper and lower body. However, there is no consensus on the optimal load regarding sex differences. Wozniak et al. (64) evaluated the performance of male and female wrestlers using an upper body (5.5% and 4.5%, respectively) and lower body (7.5% and 6.5%, respectively) relative load of body mass. They found that relative peak and mean power during both tests were higher in males than female wrestlers. However, the sex differences were reduced when power was normalized for body or fat-free mass, suggesting that anaerobic power is closely related to lean body mass.

In (65), Lovell et al. studied the effect of body strength and power on the performance during the upper body WAnT in 24 men and 16 women. Muscular strength was measured with one repetition of maximum bench press and muscular power with bench throws. The protocol of the WAnT was not specified. Their findings showed that the best indicators of upper body Wingate performance in males are body mass and, to a lesser extent, muscular power. Physical strength in women was the best predictor of performance of the upper body Wingate test. These results indicate that the quality of muscle power and force production may be used differently between males and females when performing arm cycling sprints.

The effect of sex-related differences on performance during upper body WAnT has been described in high-performance athletes. Sandbakk et al. (66) published a review summarizing the scientific knowledge concerning sex differences in world record performance. Sex differences in performance between the world's best athletes in sports in which upper-body power is a significant contributor (swimming, rowing, kayaking, canoeing) show more than 15% difference between males and females. One of the morphological explanations for this difference is the proportionally larger upper-body muscle mass of men compared to females, who generally have 25-40% less muscle mass. Regarding peak power output, during WAnT, the muscle mass of the lower extremities seems to be the main reason men perform better than women. However, for the upper-body power, this may not be the case (67,68). Another factor that seems to contribute to sex differences in power production performances is the pacing strategy. The pacing strategy is defined as the power output distribution during the exercise. During the WAnT, where they are instructed to do a high-intensity burst of exercise, males are shown to be more competitive and do not employ the pacing strategy as effectively as females (69).

To our knowledge, the only study that has explored the relative load effect and sex-related differences in power production of the upper body was conducted by Hegge et al. (70). They compared three different power outputs during a 4-minute submaximal, 3-minute submaximal, and 30-second maximal all-out test during double poling on a ski ergometer. The peak power in men was 88%, 95%, and 108% higher during the 4-minute, 3-minute, and 30-second tests, compared to females, respectively. No differences between sex or intensity were found.

Conclusion

The WAnT is a gold standard in power production testing. Most of the literature has centred around the lower body and test performance measurements. Less is known about upper body WAnT. More specifically, we know little about the biomechanical aspects of WAnT performance, how it changes due to relative load resistance during WAnT, and whether it is sex-dependent. Furthermore, this has not been analyzed with crank forces or other WAnT performance measures. The following study will be the first to use joint kinematics, crank forces, and performance measurements under different relative loads to evaluate the upper body WAnT's performance and determine if the difference (or lack thereof) in WAnT performance measures are sex-dependent.

Chapter 3: Co-authorship statement.

The following are the contributions of each author in this manuscript:

I. Conceptualization: Angie K. Antolinez, Philip Edwards, Dr. Button.

II. Research methodology design: Angie K. Antolinez, Philip Edwards, M Holmes, DC. Button.

III. Data collection: Angie K. Antolinez, Philip Edwards

IV. Data analysis and interpretation: Angie K. Antolinez, M Holmes, S Beaudette, and DC Button.

V. Drafting the article: Angie K. Antolinez, S Beaudette, and DC Button.

VI. Final approval: Angie K. Antolinez and DC Button.

Chapter 4: The effects of load, crank position, and sex on the biomechanics and performance during an upper body Wingate anaerobic test.

Antolinez Angie K.[1], Edwards Philip F. [1], Holmes Mike W.[3], Beaudette Shawn M.[3], Button Duane C. [1,2].

[1] School of Human Kinetics and Recreation and [2]Faculty of Medicine, Memorial University of Newfoundland. akantolinez@mun.ca , pfedwards@mun.ca , dbutton@mun.ca

[3]Faculty of Applied Health Sciences Brock University.

holmesmike@gmail.com , sbeaudet11@gmail.com.

Corresponding author information :

Dr. Duane C. Button, Memorial University of Newfoundland, School of Human Kinetics and Recreation, 230 Elizabeth Avenue St. John's, Newfoundland, Canada

2024

4.1 Abstract

Introduction: The upper body Wingate Anaerobic Test (WAnT) is a 30-second maximal effort sprint against a set load (percentage of body mass). However, there is no consensus on the optimal load and no differential values for males and females, even when there are well-studied anatomical and physiological differences in muscle mass for the upper body. Our goal was to describe the effects of load, sex, and crank position on the kinetics, kinematics, and performance of the upper body WAnT.

Methods: Eighteen participants (9 females) performed three WAnTs at 3, 4, and 5% of body mass. Arm crank forces, 2D kinematics, and performance variables were recorded during each WAnT.

Results: Our results showed an increase of ~49% effective force, ~36% peak power, ~5° neck flexion, and ~30° shoulder flexion from 3-5% load (p<.05). Mean power and anaerobic capacity decreased by 15%, with no changes in fatigue index (p<.05). The positions of higher force efficiency were at 12 and 6 o'clock. The least force efficiency occurred at 3 o'clock (p<.05). Sex differences showed that males produced 97% more effective force and 109% greater mean power than females, with 11.7% more force efficiency (p<.001). Males had 16° more head/neck flexion than females, and females had greater elbow joint variability with 17° more wrist extension at higher loads. Males cycled ~32% faster at 3 vs 5% load with a 65% higher angular velocity than females. Grip strength, MVIC, mass, and height positively correlated with peak and mean power (p<.001).

Conclusion: In conclusion, load, sex, and crank position have a significant impact on performance of the WAnT. These factors should be considered when developing and implementing an upper body WAnT.

Keywords: cycle ergometry, kinematics, kinetics, power.

4.2 Introduction

The study of human performance has long been a focal point in exercise science, with researchers aiming to understand the interplay of physiological, biomechanical, and sex-related factors that influence an individual's capacity for physical exertion. An essential aspect of this goal is examining anaerobic power, a critical component of overall athletic performance and a parameter for assessing an individual's physical capabilities. The Wingate Anaerobic Test (WAnT) is a widely used and established protocol that has been a gold standard in assessing anaerobic power, mainly in the lower body (3). The upper body WAnT has been used for its application in rehabilitation settings and sports but to a lesser extent.

Power production in the upper body WAnT has been primarily studied in males. Sex-related differences in physiological attributes, including muscle mass, fibre composition, and hormonal profiles, contribute to distinct performance outcomes between males and females (62,71). The disparities in the aforementioned differences in physiological attributes and upper body muscular power production between sexes underscore the need for investigation. Existing research shows consistent disparities in lower body Wingate test outcomes based on sex, with males generally displaying higher peak power, mean power, and fatigue index than females (71). However, the disparities between males and females have been less explored in terms of upper body power production. Lovell et al. (65) studied the influence of body strength (1reptitition maximum), body mass (BM), and power production (bench throws) on upper body WAnT performance (peak and mean power for both males vs females). They found that BM and muscular power best predict upper body Wingate performance for men, while only muscular power predicted performance for women. While a few sex differences have been explored regarding performance variables, characteristics like crank forces or kinematic strategies have not. In terms of upper body cycling, biomechanics studies in wheelchair propulsion showed that females had a lower gross mechanical efficiency, lower comfortable propulsion speed, higher local perceived exertion, and higher push percentage than males (72). However, no previous study has explored the sex differences in the biomechanics of the upper body WAnT. This information could give us insights into the mechanics of the movement for males and females and the possibility of developing individualized protocols to measure and calculate the maximum power production in the upper body.

These upper body WAnT mechanical factors can be categorized into kinematic and kinetic. These factors are relevant in sports and rehabilitative sciences since they describe athletes' and patients' techniques to develop strategies for performance enhancement, fatigue, injury prevention, and energy expenditure optimization (29,33,41,73). In lower-body cycling, studies have described kinematic variables like joint angles and torques and how they are influenced by bike configuration, posture, fatigue patterns, training level, and workload (29,33,74). Regarding kinetic factors, previous studies in upper and lower body cycling have described the crank forces applied during the test and how they are highly influenced by the external load, typically known as relative load (51,75,76). The relative load for the WAnT is determined as a percentage of the participant's body weight (51). The most recent study that attempts to determine a body mass factor for males and females (59) found that the optimal load for upper body WAnT was 5% BM for active and trained females, 7.5% for trained males, and 7% for active males. These relative loads have not been widely used in the literature since there are more factors than sex and training level in power production, so multiple studies have set up their loads based on the maximum possible relative load for their specific sample (9,15). Furthermore, the influence of relative load in the WAnT has only been considered in terms of peak and mean power production and kinetic and kinematic factors have not been considered.

The gap in the literature regarding the influence of sex and relative load and potential underlying mechanisms for these effects on the performance of the upper body WAnT needs to be addressed to develop more accurate testing techniques and individualized training protocols. Therefore, the objectives of this study were to 1) perform a comprehensive analysis of the kinetics, kinematics, and overall performance measures on an upper body WAnT and 2) determine the effects of external load, crank position, and sex on objective 1. We hypothesized that 1) cycling against a higher relative load would lead to increased crank forces, joint angles, and altered performance measures and 2) that changes to biomechanical and performance measures would be sex-dependent.

4.3 Methodology

4.3.1 Participants

Using a G-power analysis, it was determined that a sample of 16 participants was needed to achieve an alpha of .05 with a power of 0.8. Eighteen healthy recreationally active adults (nine males - height:175.7 ± 7.7 cm, weight: 88.5 ± 22.5 kg, age: 28 ± 4 years and nine females - height: 164.75 ± 11.25 cm, weight: 71 ± 18 kg, age: 24 ± 4 years) volunteered to participate in the study. All participants were university students at Memorial University of Newfoundland. Exclusion criteria included having a previous history of upper limb injury in the last 6 months or pain that prevented them from completing vigorous exercise. After signing the written consent form, the participants completed the Physical Activity Readiness Questionnaire (PAR-Q+) (77) to screen for contraindications to perform physical activity. Hand dominance was determined using the Edinburgh Handedness Inventory (78) questionnaire as kinetic and 2D kinematics measurements were analyzed from the dominant limb (see Experimental Protocol). Using this inventory, it was determined that 15 of the 18 participants were right-hand dominant. The Interdisciplinary Committee on Ethics in Human Research approved the study at Memorial University of Newfoundland (ICEHR No 20230904) following the Tri-Council Policy Statement for Ethical Conduct on Research Involving Humans (TCPS2) in Canada, with full disclosure of potential risks to participants.

4.3.2 Experimental Setup

4.3.2.1 Cycle Ergometer

All experimental data were recorded from the participant while being seated on a computer-controlled, electrically braked cycle ergometer modified for arm cycling (Dynafit Pro, Racemate, Seattle, Wash., USA). Participants were seated in a padded armless chair, strapped with a shoulder belt, and situated at a comfortable distance from the crankshaft so that with the crank position on the opposite side of the body and the hand grasping the handles, the elbow joint was almost in full extension (165–175°). The chair was manipulated so that the participant's shoulders were approximately in line with the axis of rotation of the ergometer with their feet strapped to the floor to minimize compensatory movements. This chair positioning was recorded for each participant and was subsequently used in every data collection session.

Crank forces were measured with a Smart Fit Power Force (SmartFit, Radlabor GmbH, Freiburg, Germany) system with a sampling frequency of 500 Hz, a measurement range from 500-1500 N, and a 12-bit resolution. The analog signal was recorded in volts and converted to newtons. The force transducers were mounted between the crank and the pedal with a 12° forward rotation about the crank. Its measuring principle determined the magnetic field variations because of the angular displacement of a small magnet with respect to 2 sensors, a sensor for radial forces and one for tangential forces. The device was calibrated every five sessions.

4.3.2.2 Grip Strength

Maximal grip strength was collected bilaterally with a Smedley Digital Hand Dynamometer (Model: 12-0286, Baseline® Digital hand dynamometer, White Plains, New York, USA) in two static asynchronous cycling positions, with both wrists in a pronated position. The first hand grip test was recorded bilaterally, with the dominant arm at 12 o'clock and the non-dominant arm at 6 o'clock. Two trials were performed. The second hand grip test was recorded bilaterally in the opposite position, with the non-dominant arm at 12 o'clock and the dominant arm at 6 o'clock; two trials were also performed. A total of 8 measurements of maximal handgrip strength were performed.

4.3.2.3 Maximal voluntary isometric contractions (MVIC)

In the same position where grip strength was recorded, the participant performed 6 total MVICs. Three MVICs of the elbow extensors with the dominant arm pushing at the 12 o'clock position while the non domminant arm was pulling at the 6 o’clock position and other three MVICs of the elbow flexors with the dominant arm pulling at the 6 o'clock position and the non-dominant arm pushing at the 12 o’clock position. The participant held each MVIC for 5 seconds while verbal encouragement was delivered, and the force data was recorded.

4.3.3 Arm Cycling WAnT

The WAnT protocol consisted of a brief 10-second warm-up at a 60 rpm cadence, followed by a visual and verbal 3-second countdown prior to the electro-mechanical brake applying a 3,4, or 5% factor of the participant body mass, and then the participants were asked to arm cycle to volitional fatigue during a 30-second all-out sprint (3). Visual cadence feedback was provided on a computer screen in front of the participants, and verbal encouragement was provided throughout the test. Crank force, performance, and kinematic data were recorded during the 30-second all-out phase of the test.

During the Arm cycling WAnt, the force applied to the crank was continiuosly measured bilaterally and extracted with the IMAGO software in two components: the effective/tangential force (FE) as the force vector that is always perpendicular to the crank and responsible for propulsion. The unused/radial force (FU) is the force component lengthwise to the crank that does not generate propulsion. Thea total force was calculated by summing the FE and FU. Performance data was collected with a 10 Hz sampling rate with the Velotron cycle ergometer

(Dynafit Pro, Racemate, Seattle, Wash., USA). The variables of mean watts, peak watts, minimum watts, mean rpm, peak rpm, peak reach (time), minimum rpm, anaerobic capacity, anaerobic power, fatigue index, and total work were all recorded.

Kinematic data was recorded using three Go Pro 10 cameras in Ultra slow-motion capture mode, 240 frames per second (fps), and a resolution of 2704x1520 pixels. Two cameras were positioned for a right and left sagittal view, and one camera was front (45 degrees off to the participant's dominant side). All cameras were 50cm from the participant. The recording was synchronized by using a voice command. A red flashlight was displayed at the start of the kinetic data recording to align with the kinematic data. See **Figure *5*** for the experimental setup.

4.3.4 Experimental protocol

The experiment protocol consisted of 4 sessions with 24 to 48 hours between sessions. The first session was a familiarization where the participant's anthropometrics were measured. Following this, the participant performed maximal grip strength bilaterally, MVICs of the dominant arm elbow flexors and elbow extensors, and three 10-second Wingate tests at the 3, 4, and 5% of their body weight. These percentages were selected after performing three trials with randomly selected participants with intensities from 3 to 7.5%. Resistances higher than 5% BM were very difficult for participants to complete the 30-second upper body WAnT. Following the familiarization session, the participants randomly completed the other three sessions, which included a 30-second upper body WAnT at a relative load of either 3, 4, or 5% of the participant's body weight. For the experimental protocol, see **Figure *6***.

4.4 Data Analysis

Grip strength and MVIC averages were calculated for the 12 and 6 o'clock positions. All crank force data during arm cycling were extracted from the IMAGO system, filtered, and processed in MATLAB R2022a using a 4^{th}-order Butterworth filter (low pass, dual pass, cut-off frequency: 50 Hz). Kinetic data were normalized to the participant's body weight. The crank angle was defined as 0 when the crank was at the 6 o'clock position. A complete cycle was defined by the crank angle, with each cycle starting at a crank angle of 0 and ending at a crank angle of 360, which was represented as a percentage from 0 to 100%. The ratio FE/FU was calculated for each trial to indicate force efficiency. Sprint performance measures, including mean and peak power output, mean and peak rpm, peak reach (time), anaerobic capacity, and anaerobic power fatigue index, were recorded using the Velotron Wingate Version 1.0.1 software.

For the kinematic data, videos were clipped to the 30-second all-out cycling, using the red flashlight emitted at the start of the recording of the kinetic data as a reference. Adobe Premiere Pro 2021 was used to clip the videos and down-sampled them to 120 fps, keeping the original resolution. Markerless pose estimation was performed using human pose–estimation software DeepLabCut (DLC) (79). This software has not been used during arm-cycling; however, an inter-class correlation coefficient of 0.98 with Kinovea for 2D kinematic analysis has been reported (38). DLC performs frame-by-frame prediction based on transfer learning with deep neural networks. Specifically, users can track novel landmarks from time-varying videos with minimal requirements for additional training data. The DeepLabCut toolbox was developed as a Python Environment in Anaconda3. The procedure for video processing in DLC was as follows: 1) creation of a project, 2) frame extraction of 1020 frames using the k means algorithm from the dominant side and frontal view videos, 3) all frames extracted underwent manual labelling of 12 reference points per frame (

Figure 7): nose, tragus, acromion (shoulder), lateral epicondyle (elbow), styloid process of the ulna (wrist), 3^{rd} metacarpophalangeal joint (knuckle), crank, the centre of the crank, chair point (hip), knee lateral joint line (knee), lateral malleolus (ankle) and first toe (toe). The greater trochanter was not possible to track because of the seat belt occluding the hip joint, so the chair point was used as a surrogate.

Manually labelled frames from each participant were then used to transfer train an already existing neural network architecture (resnest50). Specifically, 500,000 iterations (i.e., epochs) were used to train the DLC neural network with a split of a 95% train subset and a 5% test subset. The performance of the trained network was evaluated, obtaining a Train error = 4.59px and Test error = 4.31px, which represents less than 1 cm of error when tracking each labelled point (

Figure *7*). The final step included processing all videos to obtain the 2D (x,y) spatial coordinates of all tracked points. X was defined as the horizontal axis, and Y was the vertical axis. All video data were processed using a GPU parallelization (NVIDIA-RTX 6000), and all XY outputs were smoothed (moving median) using Deeplabcut.

Coordinate data was then processed using Matlab (R2022a, Mathworks Inc., Natick, MA, USA). Coordinates with a confidence level below 95% were removed, and gaps were interpolated using adjacent data points (MATLAB interp1 function, piecewise shape preserving cubic spline; PCHIP). The 2D kinematic variables included neck angle, shoulder angle, shoulder path length, elbow angle, elbow path length, and wrist angle. The intersegmental angles at the neck, shoulder, elbow, and wrist were computed from filtered marker coordinates by trigonometric equations. The neck angle was calculated as the angle between a vertical line and the head orientation (a horizontal line between the nose tip and the right tragus). The shoulder angle was defined as the angle between a vertical line and the upper arm (acromion, lateral epicondyle) segment. The elbow angle was defined as the angle between the upper arm and forearm (lateral epicondyle, styloid process of the ulna) segments. The wrist angle was defined as the angle between the forearm and the hand (styloid process of the ulna, 3rd metacarpophalangeal process) segments. The shoulder and elbow path length variables were calculated as the displacement of the acromion and lateral epicondyle points, respectively, during the 30-second WAnT. Joint paths were normalized to the number of cycles done on each trial; these variables were dimensionless. The lower body kinematics were not calculated, assuming that the most significant changes in the Upper body WAnT will occur in the upper body segments. See

Figure *7* for kinematic setup and analysis.

4.5 Statistical Analysis

Statistical analyses were completed using SPSS 28.0 (SPSS for Windows, IBM Corporation, Armonk, New York, USA). The normality of the data was assessed using both Shapiro–Wilk and Kolmogorov–Smirnov tests, and it was found that all the variables were normally distributed ($p > 0.05$). See all independent and dependent variables in Table 3. Three-way repeated measures ANOVAs were performed with factors LOAD (3,4 and 5%) x POSITION (12,3,6,9 o'clock) to evaluate all kinetics variables and joint angles; SEX was set as the between-subjects factor. For the range of motion of all joints and the shoulder and elbow path, two-way repeated ANOVAs were performed with a with-in-subjects factor, LOAD, and a between-subjects factor, SEX. A two-way ANOVA with the same model was also performed for each performance variable. If violating the assumption of sphericity, p values were adjusted using the Greenhouse–Geisser correction. Statistical significance for main tests was set at $p \leq 0.05$. In the event of a statistically significant ANOVA outcome, pairwise comparisons were completed post hoc using the Bonferroni correction. The text, tables, and figures show data as mean ± SD. Partial eta-squared (pη2) measures indicating the magnitude of changes associated with significant main effects were provided and reported as small (< 0.01), medium (≥ 0.06), or large (≥ 0.14) (80).

Two multivariate ANOVAs were performed to find the differences in Grip strength and MVIC at the 12 and 6 o'clock positions. Last, simple bivariate correlations (Pearson's r) were calculated between Grip strength, effective force, and performance measurement (peak power and mean power) and between effective force, all joint angles, and performance measurements (peak power and mean power) for all positions. The strength of the correlation coefficients (r) was interpreted as <0.3 (negligible), 0.3–0.5 (weak), 0.5–0.7 (moderate), 0.7–0.9 (strong), and > 0.9 (very strong). (81)

INDEPENDENT	DEPENDENT			Control
	Kinetic	Kinematic	Performance	
Relative load (3,4,5%)	-Effective force -Unused force -Force efficiency	-Neck angle -Shoulder angle -Elbow angle -Wrist angle -Shoulder path length -Elbow path length -Range of motion	-Peak power -Peak power reach -Mean power -Peak RPM -Mean RPM -Anaerobic capacity -Anaerobic power -Fatigue index -Total work	-Age -Height -Weight -Handedness -Grip strength -MVIC
Sex (M, F)				

Table 3. List of variables.

4.6 Results

The demographic information of the participants is summarized in Table 4. Grip strength at the 12 o'clock ($F_{(1, 16)} = 28.13$, $p < .001$, $\eta_p^2 = .64$), and 6 o'clock ($F_{(1, 16)} = 20.42$, $p < .001$ $\eta_p^2 = .56$)

positions, as well as MVIC at 12 o'clock ($F_{(1, 16)} = 28.57$, $p < .001$, $\eta_p^2 = .64$) and 6 o'clock ($F_{(1, 16)} = 82.04$, $p < .001$, $\eta_p^2 = .83$) positions were higher in males than females by 56%, 59%, 99% and 93%, respectively.

Sex	Age (years)	Height (cm)	Mass (kg)	Handedness L=left R= right	Grip 12 (kg)	Grip 6 (kg)	MVIC Dominant arm at 12 (Newtons)	MVIC Dominant arm at 6 (Newtons)	Net Load (kg)
Males n= 9	26.7 ± 2.5	176.6 ± 4.8	85.5 ± 16.7	R=8 L = 1	36.5 ± 5.9	37.9 ± 8.5	332.3 ± 72.1	278.7 ± 32.4	3% = 2.6 ± 0.5 4% = 3.4 ±0.7 5% = 4.3 ±0.8
Females n= 9	25.8 ± 2.6	163.2 ± 7.4	66.8 ± 11.8	R=7 L = 2	23.2 ± 4.3	23.8 ± 9.7	167.1 ± 58.3	144.6 ± 30.4	3% = 2 ± 0.3 4% = 2.7 ± 0.5 5% = 3.3 ± 0.6

Table 4. Demographic information of participants. Grip: grip strength and MVIC: maximum voluntary isometric contraction.

4.6.1 Kinetics

Effective force

There were significant main effects for LOAD ($F_{(2, 32)} = 20.05$, $p < .001$ $\eta_p^2 =.56$), POSITION ($F_{(3, 48)} = 94.76$, $p < .001$ $\eta_p^2 =.85$), and SEX ($F_{(1, 16)} = 20.8$, $p < .001$ $\eta_p^2 =.61$) on effective force. Post hoc comparisons showed that as the load increased, there was a significant ($p < .05$) increase in effective force by 25% from 3 to 4% load and 24% from 4 to 5% load. The effective force was significantly greater and lower at the 12 ($p < .05$) and 3 ($p < .05$) o'clock positions, respectively, compared to all other positions. Males produced 97% ($p < .001$) more effective force than females (Figure 8).

There were significant interactions for LOAD x SEX ($F_{(2, 32)} = 3.66$, $p < .001$ $\eta_p^2 =.18$), LOAD x POSITION ($F_{(6, 96)} = 7.17$, $p < .001$ $\eta_p^2 =.31$) and POSITION x SEX ($F_{(3, 48)} = 11.6$, $p < .001$ $\eta_p^2 =.42$) on effective force. Post hoc comparisons for LOAD x SEX showed that males produced a higher force at 5% vs. 3% (+53%) and 5% vs 4% (+33%) ($p < .05$) with no difference between 3 and 4% loads. Females produced a higher force at 5% vs. 3% (+58.8%) ($p < .05$), with no difference between 5% vs 4% or 4% vs 3% loads. Post hoc comparisons for LOAD x POSITION and POSITION x SEX showed that males produced 109% more (p < .05) force than females, at 5% load in all positions. At 3% (+117%) and 4% (+70%) loads, males also produced more force in all positions than females except for the 3 o'clock position ($p < .05$) (Figure *9*).

Unused Force

There were significant main effects for LOAD ($F_{(2, 32)} = 7.74, p = .002\ \eta_p^2 = .33$) and POSITION ($F_{(1.68, 26.8)} = 12.7, p < .001\ \eta_p^2 = .44$) on unused force. Post hoc comparisons revealed that as the load increased, there was a significant ($p < .05$) increase in unused force by 18% from 3 to 4% load and 28.5% from 4 to 5% load. Unused force was significantly ($p < .05$) different between all positions, being greatest at the 9 o'clock position, followed by the 12, 3, and 6 o'clock positions (see Figure *9*). There were no significant main effects for SEX ($F_{(1, 16)} = 2.7$ $p = .11\ \eta_p^2 = .14$) or any interactions on unused force for POSITION x SEX ($F_{(1.68, 26.8)} = .57, p = .54\ \eta_p^2 = .03$), LOAD x POSITION ($F_{(3.62, 58)} = 1.75, p < .15\ \eta_p^2 = .0.9$) or LOAD x POSITION x SEX ($F_{(3.62, 58)} = .85, p = .48\ \eta_p^2 = .05$).

Force efficiency

Significant main effects were found for POSITION ($F_{(2.24, 35.9)} = 36.06, p < .001\ \eta_p^2 = .69$) and SEX ($F_{(1, 16)} = 4.82$ $p = .043\ \eta_p^2 = .23$), but not LOAD ($F_{(1.24, 19.96)} = 1.41, p = .26\ \eta_p^2 = .08$) on force efficiency. Post hoc comparisons showed that force efficiency was significantly (p < .05) different between all positions, being highest at the 12 o'clock position, followed by the 6, 9, and 3 o'clock positions. Males were 11.7% ($p < .05$) more efficient than females.

Significant interactions for LOAD x POSITION x SEX ($F_{(3.5, 56.09)} = 3.22, p = .023$ η p2 =.17) on force efficiency. Post hoc comparisons showed that at 3% load, males were more efficient ($p < .05$) than females by 14% and 24% at the 12 and 6 o'clock positions, respectively. At 5% load, males were more efficient ($p < .05$) than females by 19% at the 9 o'clock position (Figure 8). Furthermore, males were more efficient $p < .05$) by 26% at the 3 o'clock position during 5%

vs 3% load. There was no difference in force efficiency at 4% load nor differences in female force efficiencies between loads (Figure *9*).

4.6.2 Kinematics

Joint angles

Since the joint angles are constantly changing following the circular path of the crank, there was a significant (p < .001) main effect for POSITION on all joint angles of the upper arm, shoulder ($F_{(1.8, 29.6)}$ = 509.6, p < .001 η_p^2 =.97), elbow ($F_{(2.14, 34.3)}$ = 181.8, p < .001 η_p^2 =.91) and wrist ($F_{(1.75, 28.04)}$ = 53.3, p < .001 η_p^2 =.76). Main effect of POSITION for neck angle was reported individually (see below) since there is no previous description of how the joint moves during high-intensity arm cycling.

Neck

There was a significant main effect for POSITION ($F_{(3, 48)}$ = 16.47, p < .001 η_p^2 =.51) and SEX ($F_{(1, 16)}$ = 17.29, p < .001 η_p^2 =.52), but not LOAD ($F_{(2, 32)}$ = 1.11, p = .34 η_p^2 =.06) on neck angle. There was a significant interaction for POSITION x SEX ($F_{(3, 48)}$ = 3.35, p = .03 η_p^2 =.17). Post hoc comparisons showed that the neck angle was significantly (p < .001) different between all positions, with greater flexion at the 9 o'clock position followed by the 6 and 3 o'clock positions. Males had ~16° more flexion (p < .001) than females across all positions (Figure 10).

Shoulder

There was no significant main effect for any factor on shoulder angle. There were significant interactions for POSITION x SEX ($F_{(1.8, 29.6)}$ = 26, p < .001 η_p^2 =.62), LOAD x POSITION ($F_{(1.8, 29.6)}$ = 61.31, p < .001 η_p^2 =.79) and LOAD x POSITION x SEX ($F_{(1.8, 29.6)}$ = 509.57, p < .001 η_p^2 =.81) on shoulder angle. Post hoc analysis showed that at 4% load, females had 36° and 29° (p < .001) more shoulder flexion than males at the 12 and 6 o'clock positions, respectively. In terms of LOAD x POSITION, at the 12 o'clock position, males flexed their shoulders 28° ($p < .001$) more at 5% and 3% vs. 4% loads, and at the 9 o'clock position, females had 5° ($p < .001$) more shoulder flexion at 4 and 5% than at 3% loads (see Figure 10).

Elbow

There were no significant main effects for LOAD ($F_{(2, 32)} = .09$, $p = .90$ η p2 =.06), or SEX ($F_{(1, 16)} = .54$, $p = .47$ $\eta_p^2 = .03$) , nor significant interactions.

Wrist

There were no significant main effects for any factor on wrist angle. However, there were significant interactions for POSITION x SEX ($F_{(3, 48)} = 3.87$, $p < .05$ $\eta_p^2 = .19$) and LOAD x POSITION ($F_{(6, 96)} = 7.38$, $p < .001$ $\eta_p^2 = .32$) on wrist angle. Post hoc analysis showed that at 5% load, females had 17° (p <.001) more wrist extension than males at the 9 o'clock position.

Range of motion

There was a significant main effect for LOAD ($F_{(2, 32)} = 5.93$, $p < .05$ $\eta_p^2 = .27$) but not for SEX ($F_{(1, 16)} = .005$, $p = .94$ $\eta_p^2 = .00$) on neck ROM. The post hoc analysis showed that neck ROM increased from 3% (7.63° ± 4.54) to 4% (9.22° ± 5.18°) to 5% (11.87° ± 6.5°) load, with only a significant ($p < .001$) difference between 3% and 5% loads. There was no significant interaction of LOAD x SEX ($F_{(2, 32)} = .60$, $p = .53$ $\eta_p^2 = .037$) on neck ROM.

There was no significant main effect for LOAD ($F_{(2,32)} = .85$, $p = .43$ $\eta_p^2 = .051$) or SEX ($F_{(1, 16)} = 1.87$, $p = .19$ $\eta_p^2 = .10$), nor an interaction of LOAD x SEX ($F_{(2,32)} = .61$, $p = .55$ $\eta_p^2 = .04$) on shoulder ROM. There was no significant main effect for LOAD ($F_{(1.4, 23.06)} = 2.7$, $p = .103$ $\eta_p^2 = .14$) or SEX ($F_{(1, 16)} = 1.23$, $p = .28$ $\eta_p^2 = .07$), nor an interaction of LOAD x SEX ($F_{(1.4, 23.06)} = .31$, $p = .66$ $\eta_p^2 = .02$) on elbow ROM. There was no significant main effect for LOAD ($F_{(2,32)} = 2.5$, $p = .09$ $\eta_p^2 = .13$) SEX ($F_{(1, 16)} = .600$, $p = .45$ $\eta_p^2 = .04$), nor interactions of LOAD x SEX ($F_{(2,32)} = 1.3$, $p = .28$ $\eta_p^2 = .07$) on wrist ROM.

Shoulder path length

There was a significant main effect for LOAD ($F_{(2, 32)} = 22.5$, $p < .001$ $\eta_p^2 = .58$) but not SEX ($F_{(1, 16)} = .19$, $p = .67$ $\eta_p^2 = .01$) on shoulder path length. Post hoc comparisons showed that shoulder joint displacement significatively (p <.05) increased from 3% (236.3 ± 51.07) to 4% (271.7 ± 67.05) to 5% (300.4 ± 69) load. There was no significant interaction of LOAD x SEX ($F_{(2, 32)} = 2.29$, $p = .11$ $\eta_p^2 = .12$).

Elbow path length

There was no significant main effect for LOAD ($F_{(2, 32)} = 317$, $p = .055$ η p2 =.16) or SEX ($F_{(1,16)} = .62$, $p = .44$ $\eta_p^2 = .04$) on elbow path length. There was a significant interaction for LOAD x SEX ($F_{(2, 32)} = 5.45$, $p < .05$ $\eta_p^2 = .25$) on elbow path length. Post hoc analysis revealed that females had a greater ($p < .001$) elbow path length at 5% (888.9 ± 93.4) vs 3% (808.8 ± 63.12).

4.6.3 Performance

Peak power

There was a significant main effect for LOAD ($F_{(1.25, 20)} = 72.29$, $p < .001$ $\eta_p^2 = .82$) and SEX ($F_{(1, 16)} = 10.4$, $p < .05$ $\eta_p^2 = .39$) on peak power. Post hoc analysis revealed that the peak power was 18% ($p < .05$) higher as the load increased from 3 to 4% and from 4 to 5%. Males produced 37% more ($p < .05$) peak power than females (Figure 11). There was no significant interaction of LOAD x SEX ($F_{(1.25, 20)} = .76$, $p = .42$ $\eta_p^2 = .04$) on peak power.

Peak power reach

There was a significant main effect for LOAD ($F_{(1.23, 19.7)} = 15.15$, $p < .001$ $\eta_p^2 = .48$) and SEX ($F_{(1, 16)} = 49.3$, $p < .001$ $\eta_p^2 = .75$) and there was a significant interaction of LOAD x SEX ($F_{(1.23, 19.7)} = 11.84$, $p = .002$ $\eta_p^2 = .42$) on peak power reach. Post hoc comparisons showed that males took 10 times and 5 times longer ($p < .05$) to reach peak power than females at the 3% and 4% loads, respectively, with no difference at the 5% load. For males, as load percentage increased from 3-5%, the time to peak power decreased ($p < .001$) from 14.5 to 3.4 to 1.9 seconds, respectively (

Table **5**).

Mean power

There was a significant main effect for LOAD ($F_{(2, 32)} = 8.99$, $p < .001$ $\eta_p^2 = .36$) and SEX ($F_{(1, 16)} = 31.55$, $p < .05$ $\eta_p^2 = .66$) on mean power. Post hoc analysis revealed that mean power decreased ($p < .001$) by 4% from 3 to 4% and by 11.5% from 4 to 5% load. Males produced 109% ($p < .001$) more mean power than females (

Table **5**). There was no significant interaction of LOAD x SEX ($F_{(2, 32)} = .39$, $p = .68$ $\eta_p^2 = .02$) on mean power.

Peak RPM

There was a significant main effect for LOAD ($F_{(2, 32)} = 13.67$, $p < .001$ $\eta_p^2 = .46$) and SEX ($F_{(1, 16)} = 31.55$, $p < .05$ $\eta_p^2 = .66$) on peak RPM. Post hoc comparisons showed that peak RPM decreased ($p < .05$) by 8.7% from 3 to 4% load and 2.4% from 4 to 5%. Males produced 13% higher ($p < .05$) peak RPM than females. There was a significant interaction of LOAD x SEX ($F_{(2, 32)} = 4.62$, $p = .047$ $\eta_p^2 = .22$) on peak RPM. Post hoc analysis showed that males were 25% faster ($p < .05$) than females at 3% load. For males, peak RPM decreased ($p < .001$) by 16% from 3 to 4% load and 16% from 4 to 5% load (

Table **5**).

Mean RPM

There was a significant main effect for LOAD ($F_{(2, 32)} = 88.52$, $p < .001$ $\eta_p^2 = .84$) and SEX ($F_{(1, 16)} = 21.51$, $p < .05$ $\eta_p^2 = .57$) on mean RPM. Post hoc analysis showed that mean RPM decreased ($p < .001$) by 36% from 3 to 4% and 33% from 4 to 5% load. Males produced 65% more ($p < .001$) mean RPM than females (Figure 11). There was no significant interaction of LOAD x SEX ($F_{(2, 32)} = 2.54$, $p = .09$ $\eta_p^2 = .14$) on mean RPM.

Anaerobic capacity

There was a significant main effect for LOAD ($F_{(2, 32)} = 11.11$, $p < .001$ $\eta_p^2 = .41$) and SEX ($F_{(1, 16)} =$, $p < 18.62.05$ $\eta_p^2 = .54$) on anaerobic capacity. Post hoc test showed that anaerobic capacity decreased ($p < .001$) by 3% from 3 to 4% and by 11.5% from 4 to 5% load. Males had a 61% ($p < .001$) higher anaerobic capacity than females (
Table **5**). There was no significant interaction of LOAD x SEX ($F_{(2, 32)} = .08$, $p = .92$ $\eta_p^2 = .005$) on anaerobic capacity.

Anaerobic power

There was a significant main effect for LOAD ($F_{(1.16, 18.66)} = 115.23$, $p < .001$ $\eta_p^2 = .88$) but not SEX ($F_{(1,16)} = 1.96$, $p = .18$ $\eta_p^2 = .11$) on anaerobic power. Post hoc analysis revealed that anaerobic power increased ($p < .001$) by 20.5% from 3 to 4% and 18.6% from 4 to 5% load. There was a significant interaction for LOAD x SEX ($F_{(1.26, 18.66)} = 5.78$, $p = .023$ $\eta_p^2 = .26$). Post hoc analysis revealed that males have a 21% higher ($p < .05$) anaerobic power than females at 3% (
Table **5**).

Fatigue index

There was a significant main effect for LOAD ($F_{(1.3, 20.84)} = 20.39$, $p < .001$ $\eta_p^2 = .56$) but not SEX ($F_{(1, 16)} = 1.9$, $p = .18$ $\eta_p^2 = .11$) on fatigue index. Post hoc analysis showed that the fatigue index increased ($p < .001$) by 31% from 3 to 4% and by 37.6% from 4 to 5% load (see
Table **5**). There was no significant interaction of LOAD x SEX on the fatigue index ($F_{(1.3, 20.84)} = 1.42$, $p = .25$ $\eta_p^2 = .08$).

Total work

There was a significant main effect for LOAD ($F_{(2, 32)} = 8.99$, $p < .001$ $\eta_p^2 = .36$) and SEX ($F_{(1, 16)} = 31.58$, $p < .05$ $\eta_p^2 = .66$) on total work. Post hoc analysis showed that total work increased

($p < .05$) by 4.2% from 3 to 4% and 11.4% from 4 to 5% load. Males performed 109% higher ($p < .001$) total work than females (

Table **5**). There was no significant interaction of LOAD x SEX on total work ($F_{(2, 32)} = .39$, $p = .64$ $\eta_p^2 = .02$).

Performance measurement	3% Males	3% Females	*P-values*	4% Males	4% Females	*P-values*	5% Males	5% Females	*P-values*	%	*P-values* Between loads Males	*P-values* Between loads Females
PP	361.6 ± 102.6	238.6 ± 35.5	*.004 **	411.2 ± 90.3	298.3 ± 49.5	*.005 **	468 ± 91.9	369.1 ±70.3	*.021 **	3-4 4-5 3-5	*<.001 ** *<.001 ** *<.001 **	*<.001 ** *<.001 ** *<.001 **
PPREACH	14.5 ± 7.8	1.21 ± 1	*<.001 **	3.4 ± 3.6	0.5 ± 0.3	*.026 **	1.9 ± 2.8	0.4 ± 0.3	*.150*	3-4 4-5 3-5	*<.001 ** *.278* *<.001 **	*1.000* *1.000* *1.000*
MP	323.4 ± 84	159.78 ± 22.1	*<.001 **	315 ± 97.2	148.9 ± 21.2	*<.001 **	284.2 ± 81.7	131.7 ± 20	*<.001 **	3-4 4-5 3-5	*1.000* *.011 ** *.015 **	*1.000* *.224* *.100*
PRPM	143.1 ± 26.7	114.8 ± 6.2	*.007 **	123.2 ± 18.1	114 ± 8.5	*.185*	119.3 ± 19	112.2 ± 5.4	*.298*	3-4 4-5 3-5	*<.001 ** *.771* *<.001 **	*1.000* *1.000* *1.000*
MRPM	129 ± 22.6	88.4 ± 13.89	*<.001 **	95.4 ± 26.9	58.8 ±13.2	*.002 **	74.8 ± 27.4	41.3 ± 7.6	*.003 **	3-4 4-5 3-5	*<.001 ** *.003 ** *<.001 **	*<.001 ** *.010 ** *<.001 **
AC	3.8 ± 0.7	2.4 ± 0.3	*<.001 **	3.7 ± 1	2.3 ± 0.5	*.002 **	3.4 ± 1	2 ± 0.4	*.001 **	3-4 4-5 3-5	*1.000* *.016 ** *.018 **	*1.000* *.057* *.022 **
AP	4.2 ± 0.8	3.5 ± 0.5	*.035 **	4.8 ± 0.7	4.5 ± 0.3	*.191*	5.5 ± 0.7	5.5 ± 0.3	*.934*	3-4 4-5 3-5	*<.001 ** *<.001 ** *<.001 **	*<.001 ** *<.001 ** *<.001 **
FI	8 ± 4.3	5 ± 1.9	*.083*	8.6 ± 3.8	8.3 ± 1.8	*.865*	12.1 ± 2.4	11.1 ± 2.1	*.363*	3-4 4-5 3-5	*1.000* *<.001 ** *.020 **	*.074* *<.001 ** *.001 **
TW	9708 .1 ± 2511.7	4795.8 ± 665.4	*<.001 **	9449 ± 2917.7	4468.6 ± 639.4	*<.001 **	8530.8 ± 2449	3957.2 ± 601.4	*<.001 **	3-4 4-5 3-5	*1.000* *.011 ** *.015 **	*1.000* *.234* *.101*

Table 5. Performance value differences between males and females and between loads. PP: peak power, PPREACH: peak power reach, MP: mean power, PRPM: peak revolutions per minute, MRPM: mean revolutions per minute, AC: anaerobic capacity, AP: anaerobic power, FI: fatigue index, and TW: total work. *indicates significant differences (p <.05)

4.6.4 Change in power outputs between the loads

There was a significant main effect for LOAD ($F_{(2, 32)} = 38$, $p < .001$ $\eta_p^2 = .704$) but not SEX ($F_{(1, 16)} = .85$, $p = .371$ $\eta_p^2 = .05$) on the change in peak power. Post hoc analysis revealed there was significant ($p < .001$) power loss between the loads, with the highest power loss from 3 to 5% (-53%, 54.7 ± 29.9 watts) followed by 4 to 5% (-46%, 63.7 ± 34.8 watts) and the smallest loss was from 3 to 4% (-14%, 118.4 ± 55.3 watts). There was no significant effect for LOAD ($F_{(1.21, 19.38)} = 2.82$, $p = .103$ $\eta_p^2 = .150$) or SEX ($F_{(1, 16)} = .42$, $p = .52$ $\eta_p^2 = .02$) on change in peak power.

4.6.5 Correlations

For the whole group, there was a positive correlation between effective force, peak power, mean power, MVIC, and grip strength in all loads. For males, there was a medium correlation between effective force and peak power with handgrip at 4% and a strong correlation between peak power at 5% and MVIC. There were no significant correlations between any variables for Females (**Error! Reference source not found.**).

		All grip strength		All MVIC		Males grip strength		Males MVIC		Females grip strength		Females MVIC	
		ρ	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*
3%	EF	.780*	<.001	.735*	<.001	.65	.058	.23	.54	.012	.97	.354	.35
	PP	.750*	<.001	.669*	<.001	.66	.052	.49	.17	-.001	.99	-.18	.64
	MP	.841*	<.001	.835*	<.001	.60	.083	.65	.054	.46	.21	.178	.64
4%	EF	.802*	<.001	.509*	<.001	.68*	.044	-.27	.47	.67	.047	.49	.17
	PP	.742*	<.001	.739*	<.001	.67*	.047	.59	.09	.092	.81	.32	.39
	MP	.804*	<.001	.741*	<.001	.58	.102	.46	.21	.16	.67	-.28	.46
5%	EF	.772*	<.001	.754*	<.001	.50	.16	.32	.39	.017	.96	.21	.59
	PP	.625*	<.001	.776*	<.001	.52	.15	.83*	.006	.13	.72	.44	.23
	MP	.774*	<.001	.772*	<.001	.45	.23	.46	.21	.10	.79	.08	.84

Table 6. Pearson correlations between EF: effective force, PP: peak power, MP: mean power, MVIC: maximal voluntary isometric contraction, and Grip strength. *indicates a significant correlation ($p < .05$). The reference position for all variables was the 12 o'clock position.

There was a positive, strong correlation for the whole group ($p < .05$) between mass and height and peak power and mean power at all loads. In males, there was a strong correlation between height and weight for peak power in all loads and for mean power at 3%. Finally, there was a strong correlation between weight and peak power at 4 and 5% for females but no correlation for the other loads or mean power (Table 7).

		All weight		All height		Males weight		Males height		Females weight		Females height	
		ρ	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*	*ρ*	*P-values*
3%	PP	.80*	.001	.70*	.001	.77*	.015	.73*	.025	.54	.132	.20	.604
	MP	.79*	<.001	.76*	<.001	.76*	.016	.74*	.023	.63	.068	.004	.99
4%	PP	.88*	<.001	.74*	<.001	.77*	0.14	.69*	.038	.93*	<.001	.43	.24
	MP	.59*	.005	.66*	.005	.39	.301	.44	.232	-.15	.69	-.66	.05
5%	PP	.90*	.001	.69*	.001	.79*	.011	.72*	.030	.97*	<.001	.42	.26
	MP	.62*	.004	.65*	.004	.36	.336	.36	.340	.37	.328	-.42	.26

Table 7 Correlations between weight and height and peak power and mean power. *shows significant correlations. PP: peak power and MP: mean power.

4.7 Discussion

To our knowledge, this is the first study determining the relative load, crank position, and sex-specific differences in kinetics, kinematics, and performance parameters during the upper body WAnT. The effect of relative load on performance showed an increase in peak power, anaerobic power, fatigue index, and total work as the load increased, whereas mean power and anaerobic capacity decreased as the load increased. Power production was also sex-dependent. Males had higher peak power and produced approximately twice the mean power than females. These results were similar to the kinetics findings. Participants produced more effective and unused force as the load increased, with no difference in force efficiency. Males produced twice the effective force and were more efficient than females during the upper body WAnT. We also found sex-dependent kinematic differences between males and females during the upper body WAnT. Males held a greater neck flexion throughout the WAnT, which increased with load, whereas females had higher variability at the elbow joint and used greater wrist extension to overcome the 5% relative load. Thus, males and females employ different biomechanical strategies during the upper body WAnT. The crank position also affected participant biomechanics during the WAnT. The 12 o'clock position was shown to be the point with the highest force efficiency, and the 3 o'clock position showed the lowest efficiency. Postural and kinematic adjustments of the shoulder and wrist joints were also load and position-dependent. Finally, we found non-sex-dependent positive correlations between grip strength and MVIC with peak and mean power. There was also a positive correlation between weight and height, peak power, and mean power. Overall, upper body WAnT performance is load- and sex-dependent. The differences in WAnT performance as the load is increased and due to sex are, in part, biomechanically driven.

4.7.1 Load and sex differences in the performance variables.

Peak and mean power have been the main focus of WAnT performance (82). Our results showed that peak power increased and mean power decreased from 3-5% load. The peak power increase at higher loads is consistent with previous literature in the lower body WAnT with load increments from 8 to 9% BW (83,84) and in the upper body WAnT with loads from 5% to 8% BW (59). However, in terms of mean power, the studies above performed on the lower body showed that the mean power also increased at higher loads, but this was not the same for the upper body. Forbes et al. (59) showed that mean power tended to decrease at higher loads for trained and recreationally active males (it reduced after 7.5% relative load) and for trained and recreationally active females (it reduced after 5% relative load). A potential reason for the reduction in mean power in the upper body but not the lower body might be attributed to differences in muscle size, force generation capacity, and fibre type distribution (11). Also, in our study, males and females had different power production patterns at 5%. Females suddenly dropped in power 15 seconds into the test vs. males, who progressively lost power over the 30-second window load (see Figure 11). This difference could have influenced our results, leading to an underestimation of mean power for the whole group. Further research into the power development at different time points of the WAnT is needed to understand sex and load effects in power production.

In terms of specific sex differences, males produced 37% and 109% more peak and mean power output, respectively, than females. These results agree with previous literature comparing male and female power production. Bartolomei et al. (71) found that males produced 157% and 178% more power during a bench press throw (BPT) and countermovement jump power (CMJP), respectively, than females. Lovel et al. (65) compared males and females performance on the lower body WAnT with a 5% BW load, finding that males produced 70% more peak power and 124% mean power than females. Heller et al. (85) used five 7-second, ascending loads sprints with interspersed breaks of 5 minutes for the upper (load:19.6 N to 58.8 N) and lower (load: 29.4 to 68.6 N) bodies. They found that males had a higher peak power than females, 43% in the lower limbs and 62% in the upper limbs. A study comparing anaerobic performance in female and male wrestlers during lower body (load, males: 7.5%, females 6.5% BW) and upper body (load, males: 5.5%, females 4.5% BW) WAnT found that males had a 62% higher peak and a 57% higher mean power than females in the lower body and a 102% higher peak and an 82% higher mean power than females in the upper body (64). These studies are comparable to our results in that males produced more power overall than females. However, the percentage of power production difference between males and females reported in these studies differs from ours. This can be explained because different load factors and experimental set-ups were used in these studies with respect to ours. The power output differences between males and females have been attributed to physiological differences such as males having 1) 40% more muscle mass than females in the upper body (86), 2) a higher number of type II fibres (87), and 3) a greater glycolytic capacity and a greater reliance on the glycolytic pathways compared to females (88). Interestingly, the 100-150% difference in power production surpasses the percentages in anatomical differences between males and females (around 40%), all the factors that contribute to the sex-related effects in power production have not been described yet.

Another important factor regarding power production in the WAnT is the time in the test at which the participant reaches their peak power, commonly known as peak power reach. Previous literature has reported that peak power reach occurs within the first 5 seconds of the WAnT (14). In our study, the average male reached their peak power at 14 seconds against a 3% load, 3.4 seconds at 4%, and 1.9 seconds at 5%. This suggests that a 3% load is insufficient to elicit the early onset of peak power in males. On average, females reached their peak power in the first 2 seconds at 3% and in less than half a second at 4 and 5% loads. Therefore, female power production is depleted within the first couple of cycles. These results suggest that the optimal load for males and females needs to be determined individually. The first study on load optimization of the WAnT defined the optimal load as the load at which the participant produced the highest power output (peak power) (51). However, there is no agreement on the most reliable measurement of anaerobic performance between peak and mean power to determine the optimal load. It has been found that all-out tests lasting more than 10 seconds largely depend on aerobic metabolism (57). Therefore, the optimal duration of the test and the most reliable measures of true anaerobic performance are still up for debate. Most of the literature in WAnT has set the relative load in terms of a body weight percentage without implementing a sex factor (57). Even when performance is normalized to body weight, as is the case of anaerobic capacity (mean power/body weight) and anaerobic power (peak power/ body weight), our study showed that males still had 61% higher anaerobic capacity and 21% higher anaerobic power, at 3% load, than females.

Similar to previous literature (89), we found a higher fatigue index and power drop at higher relative loads during the WAnT. However, we did not find sex differences like previous studies have reported during the lower and upper body WAnT, where males had a higher fatigue index (21,65). The possible explanation for this phenomenon is that in our study, the difference in peak power was 37% vs previous studies that reported up to 60% differences.

In order to find other potential factors that influenced performance in the upper body WAnT, we performed Pearson correlations between MVIC, grip strength, weight, height peak power, and mean power. Sex differences were found, with males producing higher MVIC and grip strength than females, which is consistent with previous studies (90,91). Interestingly, there were positive correlations at all loads between effective force ($\rho > 0.5$, $\rho > 0.78$), peak power ($\rho > 0.66$, $\rho > 0.75$), mean power ($\rho > 0.75$, $\rho > 0.74$) and MVIC and grip strength respectively. However, when separating males and females, the only significant correlations for males were between effective force and grip strength and peak power and grip strength at 4% load and between peak power and MVIC at 5% load. To our knowledge, no previous studies have considered MVIC in the hand-cycling position or grip strength as potential indicators for determining optimal load for upper body WAnTs. These factors can potentially establish a relative load factor corresponding to individual characteristics, like sex, age, or training level since previous studies have shown a correlation between force-velocity and power-velocity development (60,85).

In terms of the correlations between height and weight and peak power/mean power production, we found stronger correlations between weight and peak power ($\rho > 0.8$) than between weight and mean power ($\rho > 0.6$) at all loads. Height correlated higher with peak power and mean power at lower loads ($\rho > 0.7$) than at higher loads ($\rho > 0.6$). When correlations were run separately for males and females, males weight correlated with peak power ($\rho > 0.8$) at all loads, but females just did at 4 and 5%. Males' height also correlated with peak power at all loads ($\rho > 0.7$), but there was no correlation for females. This result indicates that height may influence power production because males have longer body segments, leading to a potential mechanical advantage. Further research is needed to establish the net impact of body segments length in upper body power production and how beneficial it would be to find an optimal position to perform the test that reduces the mechanical advantage that some individuals may have over others.

4.7.2 Load, crank position, and sex differences in the kinetic variables.

Our results revealed that individuals produced more effective and unused force at higher loads. When increasing load from 3 to 4%, males and females increased their force by at least 50%. The similar increases in force for males and females were not the same for power. Thus, force about the crank is not the only factor to help explain the differences between male and female power output during an upper body WAnT.

In terms of force efficiency, males produced twice the effective force and were more efficient than females at 3% and 4% load at the 12 and 6 o'clock positions. At 5% load, males were also more efficient, but only at the 9 o'clock position (with the lowest mechanical advantage). It is important to mention that it has been found that kinetics, kinematics, and performance have been proven to be task-dependent. The current study had high relative forces (23 to 46 N) cadences (88 RPM to 130), and the power produced was between 240 and 360 watts. Drongelen et al. (92) studied arm cycling under resistances from 4.9 N to 24.30 N and showed that the effective force increased as the load increased and that the highest force point was between the 3 o'clock and 6 o'clock positions. Kraaijenbrink et al. (75) measured the crank forces during hand-cycling at different cadences (52 to 70 RPM) at three different resistances (0, 10, and 20 Watts). It was found that effective and unused forces and force efficiency increased as the resistance increased, and the position in the cycle with the highest effective force occurred between the 3 and the 6 o'clock positions. Kraaijenbrink et al. (72) found that effective force is highest in the asynchronous mode for the push/press down phases (12 to 3 o'clock). Quittmann et al. (26) examined the changes in hand-cycling propulsion due to increasing loads starting at 20 Watts, and they found that maximal torque occurred within the pull-down (3 to 6 o'clock) and push-up sector (9 to 12 o'clock), whereas the lowest torque and cadence occurred during the lift-up (6 to 9 o'clock) during hand-cycling propulsion. Kraaijenbrink et al. (72) compared the kinematics and kinetics of synchronous vs. asynchronous hand-cycling, finding that effective force is highest in the asynchronous mode for the push/press down phases (12 to 3 o'clock).

In summary, the differences between the aforementioned studies and ours suggest that during high-intensity cycling, the predominant phase is the pushing phase, whereas in low-intensity hand-cycling, the dominant phase is the pulling phase. However, we must look at both arms during the WAnT to describe the force production strategy. Is it that people are pushing more with their dominant arm or is it that they are pulling with the non-dominant arm to produce overall power production.

To our knowledge, no previous study has determined sex differences in force production during the WAnT. We found that males produced 97% more effective force, and they were 11.7% more efficient than females. Females typically have 40% less muscle mass in the upper limbs than males (86). Muscle strength, or force-generating capacity, is closely related to muscle mass and cross-sectional areas (87,93). This suggests that anatomical features might affect the ability of females to produce power or force. Even when normalized to the body weight, females have a disadvantage when performing the upper body WAnT at the same relative load as males. These findings in force production might be attributed to anatomical, physiological, and metabolic differences between males and females. In terms of force production in resistance training, it has been found that female strength as a percent of male strength is 46, 56, and 61% for the bench press, squat, and deadlift 1RMs, respectively (94). More data is needed to describe the differences between male and female force production changes during various exercise types since these differences are muscle and task-specific (95).

4.7.3 Load, crank position, and sex differences in the kinematic variables.

To our knowledge, this is the first study to measure biomechanical variables of the upper body WAnT. Most literature on the kinematics of upper-body cycling has focused on specific arm cycling tasks, such as competitive hand-cycling or wheelchair propulsion. In recumbent hand-cycling, peak elbow extension and shoulder adduction occur at the furthest reach, as the arms are maximally extended, while peak elbow flexion and shoulder abduction occur at the closest reach, as the arm is maximally flexed. Shoulder flexion and extension occurred at the highest and lowest points of the cycle (12 and 6 o'clock positions), respectively (96,48). In terms of the wrist, the main action occurs when extending ~35° at the beginning of the pushing phase (9 o'clock position) (28). The results above, as related to the current study of kinematics in the sagittal plane (flexion/extension movements), must be interpreted cautiously as they might not apply to arm-cycling sprinting. Additionally, in the current study, the configuration of the ergometer and the intensity and speed variables differed from previous studies (26,30,49,97). As Botzheim et al. (50) suggested, those differences affect muscle synergies activation and could potentially change the kinematics of the movement task.

We found that at higher loads, there was a higher neck and shoulder flexion, and these flexions were sex-dependent. Males flexed their necks 16° more than females during the whole WAnT duration, and males' neck ROM increased as the load increased. The neck and neck position has been shown to play a vital role in the kinetic chains during locomotion and also plays a vital role in balance and proprioception (98). Aruin and Latash (99) found that when humans perform voluntary arm movement, postural equilibrium suffers because limb and body geometry changes, reflecting a postural adjustment. Based on these findings, Hiraoka et al. (100) explored if immobilization of the neck and trunk on arm-cycling induced depression of leg motoneuron pool excitability. They found that the H-reflex depression during arm cycling in the non-immobilized condition was significantly larger than in the immobilized condition, suggesting a relationship between neck and trunk posture during arm cycling performance. The only study that explored the effect of posture on power production was done by Offley et al. (101), who found that a neck flexion posture can potentiate performance in power production during squatting. More research is needed on the role of the neck position in power production tasks and the influence of sex on these parameters.

We also found sex differences in the movement pattern of the shoulder joint. Males flexed their shoulders more at 5% and 3% compared to 4% load. Females flexed their shoulders more at 5% compared to 3 and 4% load. This could be related to the fact that 3% was a light relative load for males, and the joint has a higher degree of freedom to perform the task. However, at 5%, the shoulder worked as the primary joint to generate more power for males and females with increased shoulder flexion, especially at the 12 o'clock position.

The elbow joint did not show any significant differences between loads. A reason for this can be the limited degrees of freedom determined by the circular path of the crank, limiting its movement to a flexion-extension trajectory. Lower limb cycling has shown differences mainly in the hip and ankle joints at different loads (29,73,74). This might mirror our results in that the most proximal and distal joints have higher adaptations when changing loads than the middle joints (knee or elbow). Joints like the knee and elbow are more restricted to the circular path of the crank (102). However, in our study, females had higher variability in the elbow joint at higher loads, which might suggest the use of more distal joints when females cycle against heavier loads. Furthermore, females also showed an increased wrist extension by 17° when arm cycling against a 5% load at the 9 o'clock position, a position with a higher mechanical disadvantage. The fact that wrist extension did not occur in females suggests that wrist extension might represent a mechanical disadvantage for power production since previous studies have described wrist stiffness as an essential strategy to improve distal force transmission, especially at higher loads (26).

RPM or angular velocity is another kinematic factor interpreted as a performance index. As mentioned in previous sections, force is not the only determinant variable for power production since power is a product of force times velocity. Our study showed that there was a reduction in peak RPM for males at higher loads. However, females kept the same angular velocity between loads. Our findings are similar to previous studies that showed that males produce 25-30% higher peak RPM than females (65,84). This finding is important since it suggests that power production in males relies on their force production but not their capacity to generate speed. Females keep the same speed independently from the load even when producing significantly less force than males. There is less research on the impact of different loads on RPM and its potential to explain the different strategies used by males and females when performing the upper body WAnT.

Finally, we did not find statistical correlations between kinetic, kinematic, and performance factors. However, our results suggest that the big difference in performance between males and females might be attributed to a difference in force production and not to differences in angular velocities during the test. Additionally, the body weight factor might not be appropriate to compare within and between males and females since it overlooks anatomical and physiological differences in the power production of the upper body.

4.8 Methodological considerations.

In this study, several methodological aspects warrant careful consideration, such as using just 2D kinematics, evaluation of only one side of the body, variability in height and weight among participants, and the absence of body composition and metabolic data. Although a practical and highly validated approach, 2D kinematics presents inherent limitations in capturing the full range of movement. Since arm cycling occurs mainly in the sagittal plane, 2D kinematics have high reliability for uni-planar findings. However, using 3D kinematics could provide a more comprehensive assessment of three-dimensional motion, like shoulder adduction or abduction, that has been reported in previous studies (76). Therefore, any conclusions drawn from this data should be made with the understanding that they may not fully represent the complexities of upper body motion.

Limiting the analysis to only one side of the body rather than both might overlook potential bilateral differences in performance and kinematics. This approach may also neglect significant asymmetries, which can be critical in understanding upper body anaerobic power, especially in sports or activities that involve unilateral movements. Furthermore, when analyzing kinematics, previous studies on lower-body cycling (29,43) have included the upper and lower body, even when the main body segments in action were the legs. In our study, we isolated the analysis of the upper body, disregarding the potential contribution of the lower body and the trunk to the performance and the kinematic strategies during the upper body WAnT. Future studies should implement the analysis of joints like the trunk, scapula, hip, knee, and foot. Our observation shows considerable movement in the lower body, especially at higher intensities. This might be useful to understand how lower body and trunk postural perturbations during the upper body WAnT are affected by load, sex, and crank position and their effect on the upper body during the upper body WAnT.

The variability in height and weight among participants may introduce confounding factors for WAnT performance. Varying weight and height impact lever lengths, muscle attachments, and overall biomechanics, potentially influencing the performance and kinematic outcomes. While it is challenging to homogenize height and weight in a diverse sample, statistical control measures or subgroup analyses may help address this issue. Also, body weight is very relevant since it is currently used as the relative load factor. Similar to our study, most WAnT research normalizes all test participants to their total body mass. However, this factor was established more than 50 years ago, and from our results and previous studies, it needs further investigation to develop a factor that includes sex, kinetic, kinematic, and performance factors.

The absence of detailed body composition data, such as muscle mass, is a limitation in this study. Muscle mass variations between groups can significantly influence power production, particularly in anaerobic activities. The lack of body composition data makes it challenging to control for these differences. Future studies should consider collecting body composition information to pair-match participants based on muscle mass, reducing the influence of this variable. Also, anthropometric measures like body segment length could provide valuable information to understand the interaction between the upper limbs and the ergometer.

Including females in force or power production studies has been controversial in exercise and sports sciences (103). The main reason to exclude women was the potential variations in performance due to changes in the menstrual cycle. Our study did not control this factor since recent research has shown that menstrual cycle changes do not significantly impact the strength of female power production (104,105). However, more research is needed to determine if hormonal factors affect performance during the upper body WAnT test for both males and females.

Finally, it is recommended that future studies perform inverse dynamics analysis with the kinetic and kinematic data to evaluate joint torques and mechanical stress for each body part. This information could be beneficial in describing the optimal loads to enhance performance and reduce injury risk when using upper-body sprinting as a training method.

Conclusions

In conclusion, this study showed that load, crank position, and sex influence upper body WAnT. The conventional practice of selecting load relative to body weight appears to be suboptimal, especially for females, as it significantly hampers their performance. Our findings emphasize using mean power instead of peak power as a reliable and sensitive performance indicator, particularly for assessing the effects of load variations and sex differences.

A notable aspect for further investigation is the distinctive neck flexion strategy observed in males, which seems to enhance their performance. A deeper understanding of this strategy could provide insights for optimizing test outcomes and training protocols. Additionally, it becomes evident that there is a need to expand the analysis to include not only the upper body but also the trunk and lower body, which can substantially impact power output during the WAnT. By exploring these aspects further, a more comprehensive understanding of the factors at play in upper body WAnT performance can be achieved.

Given these findings, further research should focus on developing an approach that creates a relative load factor that accounts for sex differences, kinetics, and kinematic strategies. That would allow for a more equitable evaluation of upper body Wingate test performance and improved training and assessment practices guidance.

References

1. Bar-Or O. The Wingate Anaerobic Test An Update on Methodology, Reliability and Validity. Sport Med An Int J Appl Med Sci Sport Exerc. 1987;4(6):381–94.
2. Nelson R, Morehouse C. Biomechanics IV. Vol. 1, International Series on Sport Sciences. 1974. 217–233 p.
3. Castañeda-Babarro A. The wingate anaerobic test, a narrative review of the protocol variables that affect the results obtained. Appl Sci. 2021;11(16).
4. Legaz-Arrese A, Munguia-Izquierdo D, Carranza-Garcia L, Torres-Davila C. Validity of the wingate anaerobic test for the evaluation of elite runners. J Strength Cond Res. 2011;25(3):819–24.
5. Tharp GD, Newhouse RK, Uffelman L, Thorland WG, Johnson GO. Comparison of sprint and run times with performance on the wingate anaerobic test. Res Q Exerc Sport. 1985;56(1):73–6.
6. Zupan M, Arata A, Dawson L, Wile A, Payn T, Hannon M. Wingate anaerobic test peak power and anaerobic capacity classifications for men and women intercollegiate athletes. J ofStrength Cond Res. 2009;23(9):2598–604.
7. Armstrong N, Welsman JR, Kirby BJ. Performance on the Wingate anaerobic test and maturation. Pediatr Exerc Sci. 1997;9(3):253–61.
8. Kaupp C, Pearcey GEP, Klarner T, Sun Y, Cullen H, Barss TS, et al. Rhythmic arm cycling training improves walking and neurophysiological integrity in chronic stroke: The arms can give legs a helping hand in rehabilitation. J Neurophysiol. 2018;119(3):1095–112.
9. Franchini E. Upper-body Wingate test classificatory table for adult judo athletes. J Exerc Rehabil. 2019;15(1):55–9.
10. Vigotsky AD, Halperin I, Lehman GJ, Trajano GS, Vieira TM. Interpreting signal amplitudes in surface electromyography studies in sport and rehabilitation sciences.

Front Physiol. 2018;8(1).

11. Patton JF, Duggan A. Upper and lower body anaerobic power: comparison between biathletes and control subjects. Int J Sports Med. 1987;8(2):94–8.

12. Andreato L V., Leite AE, Ladeia GF, Follmer B, de Paula Ramos S, Coswig VS, et al. Aerobic and anaerobic performance of lower- and upper-body in Brazilian jiu-jitsu athletes. Sci Sport. 2022;37(2):145.e1-145.e8.

13. Vural F, Erman B, Ozkol M, Aksit T. The Relationship Between the Force Production in the Isometric Squat and Bench Press Exercises and the Lower and Upper Body Anaerobic Power Parameters. Spor Bilim Araştırmaları Derg. 2023;8(2):293–305.

14. Özbay S, Ulupınar S. Strength-Power Tests are More Effective When Performed after Exhaustive Exercise in Discrimination between Top-Elite and Elite Wrestlers. J Strength Cond Res. 2022;36(2):448–54.

15. Andrade VL, Foresti YF, Almeida PB, Araujo MYC, Kalva-Filho CA, Papoti M. Validity and reliability of a specific anaerobic test for mixed martial arts. Sci Sport. 2022;37(5–6):488.e1-488.e10.

16. Huxley AF, Niedergerke R. Structural changes in muscle during contraction. Nature. 1954;4412(4412):971–3.

17. Squire JM. Muscle contraction: Sliding filament history, sarcomere dynamics and the two Huxleys. Glob Cardiol Sci Pract. 2016;2016(2):5–23.

18. Fitts RH, McDonald KS, Schluter JM. The determinants of skeletal muscle force and power: Their adaptability with changes in activity pattern. J Biomech. 1991;24:111–22.

19. Douglas J, Ross A, Martin JC. Maximal muscular power: lessons from sprint cycling. Sport Med - Open. 2021;7(1):7–48.

20. Hargreaves M, Spriet LL. Skeletal muscle energy metabolism during exercise. Nat Metab. 2020;2(9):817–28.

21. Beneke R, Pollmann C, Bleif I, Leithäuser RM, Hütler H. How anaerobic is the wingate anaerobic test for humans? Eur J Appl Physiol. 2002;87(4–5):388–92.

22. Lovell D, Kerr A, Wiegand A, Solomon C, Harvey L, McLellan C. The contribution of energy systems during the upper body Wingate anaerobic test. Appl Physiol Nutr Metab. 2013;38(2):216–9.

23. Price M, Beckford C, Dorricott A, Hill C, Kershaw M, Singh M, et al. Oxygen uptake during upper body and lower body wingate anaerobic tests. Appl Physiol Nutr Metab. 2014;39(12):1345–51.

24. Dawson B, Goodman C, Lawrence S, Preen D, Polglaze T, Fitzsimons M, et al. Muscle phosphocreatine repletion following single and repeated short sprint efforts. Scand J Med Sci Sport. 1997;7(4):206–13.

25. Bressel E, Heise GD. Effect of arm cranking direction on EMG, kinematic, and oxygen consumption responses. J Appl Biomech. 2004;20(2):129–43.

26. Quittmann OJ, Meskemper J, Abel T, Albracht K, Strüder H. Changes in the kinematic and kinetic profile of handcycling propulsion due to increasing workloads. ISBS Proc Arch. 2017;35(1):706–9.

27. Mossberg K, Willman C, Topor MA, Crook H, Patak S. Comparison of asynchronous versus synchronous arm crank ergometry. Spinal Cord. 1999;37(8):569–74.

28. Faupin A, Gorce P, Watelain E, Meyer C, Thevenon A. A biomechanical analysis of handcycling: A case study. J Appl Biomech. 2010;26(2):240–5.

29. Holliday W, Theo R, Fisher J, Swart J. Cycling: joint kinematics and muscle activity during differing intensities. Sport Biomech. 2019;00(00):1–15.

30. Faupin A, Gorce P, Meyer C. Effects of type and mode of propulsion on hand-cycling biomechanics in nondisabled subjects. J Rehabil Res Dev. 2011;48(9):1049–60.

31. Chaytor CP, Forman D, Byrne J, Loucks-Atkinson A, Power KE. Changes in muscle activity during the flexion and extension phases of arm cycling as an effect of power

output are muscle-specific. PeerJ. 2020;8.

32. Botzheim L, Laczko J, Torricelli D, Mravcsik M, Pons JL, Barroso FO. Effects of gravity and kinematic constraints on muscle synergies in arm cycling. J Neurophysiol. 2021;125(4):1367–81.

33. Swart J, Holliday W. Cycling Biomechanics Optimization - The (R) Evolution of Bicycle Fitting. Curr Sports Med Rep. 2019;18(12):490–6.

34. Colyer SL, Evans M, Cosker DP, Salo AIT. A Review of the Evolution of Vision-Based Motion Analysis and the Integration of Advanced Computer Vision Methods Towards Developing a Markerless System. Sport Med. 2018;4(24).

35. Kanko RM, Laende EK, Davis EM, Selbie WS, Deluzio KJ. Concurrent assessment of gait kinematics using marker-based and markerless motion capture. J Biomech. 2021;127.

36. Lam W, Tang Y, Fong K. A systematic review of the applications of markerless motion capture (MMC) technology for clinical measurement in rehabilitation. J Neuroeng Rehabil. 2023;20(1):1–26.

37. Nakano N, Sakura T, Ueda K, Omura L, Kimura A, Iino Y, et al. Evaluation of 3D Markerless Motion Capture Accuracy Using OpenPose With Multiple Video Cameras. Front Sport Act Living. 2020;2(May):1–9.

38. McGuirk TE, Perry ES, Sihanath WB, Riazati S, Patten C. Feasibility of Markerless Motion Capture for Three-Dimensional Gait Assessment in Community Settings. Front Hum Neurosci. 2022;16(June):1–15.

39. Wade L, Needham L, McGuigan P, Bilzon J. Applications and limitations of current markerless motion capture methods for clinical gait biomechanics. PeerJ. 2022;10:1–27.

40. Mathis A, Warren R. On the inference speed and video-compression robustness of DeepLabCut. bioRxiv. 2018;

41. Bini RR, Carpes FP. Biomechanics of Cycling. Springer. 2014. 1–130 p.

42. Davis RR, Hull ML. Measurement of pedal loading in bicycling: II. Analysis and results. J Biomech. 1981;14(12).

43. Rossato M, Bini RR, Carpes FP, Diefenthaeler F, Moro ARP. Cadence and workload effects on pedaling technique of well-trained cyclists. Int J Sports Med. 2008;29(9):746–52.

44. Díaz-Bravo L, Torruco-García U, Martínez-Hernández M, Varela-Ruiz M. La Entrevista, Recurso Flexible y Dinámico-The interview, a Flexible and Dynamic Resource. Investig en Educ médica. 2013;2(7):162–7.

45. Martin JC, Brown NAT. Joint-specific power production and fatigue during maximal cycling. J Biomech. 2009;42(4):474–9.

46. Smith PM, Chapman ML, Hazlehurst KE, Goss-Sampson MA. The influence of crank configuration on muscle activity and torque production during arm crank ergometry. J Electromyogr Kinesiol. 2008;18(4):598–605.

47. Bressel E, Bressel M, Marquez M, Heise GD. The effect of handgrip position on upper extremity neuromuscular responses to arm cranking exercise. 2001;11:291–8.

48. Hendrik E, Von Tscharner V, Nigg B. Neuromuscular Strategies during Cycling at Different Muscular Demands. 2015;1450–9.

49. Nevin J, Kouwijzer I, Stone B, Quittmann OJ, Hettinga F, Abel T, et al. The Science of Handcycling: A Narrative Review. Int J Sports Physiol Perform. 2022;17(3):335–42.

50. Davis RB, DeLuca PA. Gait characterization via dynamic joint stiffness. Vol. 4, Gait and Posture. 1996. p. 224–31.

51. Dotan R, Bar-Or O. Load optimization for the wingate anaerobic test. Eur J Appl Physiol Occup Physiol. 1983;51(3):409–17.

52. Forman DA, Philpott DTG, Button DC, Power KE. Cadence-dependent changes in

corticospinal excitability of the biceps brachii during arm cycling. J Neurophysiol. 2015;114(4):2285–94.

53. Spence AJ, Alcock LR, Lockyer EJ, Button DC, Power KE. Phase- and workload-dependent changes in corticospinal excitability to the biceps and triceps brachii during arm cycling. Brain Sci. 2016;6(4):14–6.

54. Nippard AP, Lockyer EJ, Button DC, Power KE. Corticospinal excitability to the biceps and triceps brachii during forward and backward arm cycling is direction-and phase-dependent. Appl Physiol Nutr Metab. 2020;45(1):72–80.

55. Kolesnikov M, Piovesan D, Lynck KM, Mussa-Ivaldi FA. On Force Regulation Strategies in Predictable Environments Maxim. HHS Public Access. 2011;2(1):4076–81.

56. Piovesan D, Kolesnikov M, Lynch K, Mussa-Ivaldi FA. The Concurrent Control of Motion and Contact Force in the Presence of Predictable Disturbances. J Mech Robot. 2019;11(6):1–14.

57. Driss T, Vandewalle H. The measurement of maximal (Anaerobic) power output on a cycle ergometer: A critical review. Biomed Res Int. 2013;2013.

58. Evans J, Quinney H. Determination of resistance settings for anaerobic power testing. Can J Appl Sport Sci. 1981;6:53–6.

59. Forbes SC, Kennedy MD, Boule NB, Bell G. Determination of the optimal load setting for arm crank anaerobic testing in men and women. Int J Sports Med. 2014;35(10):835–9.

60. Michalik K, Smolarek M, Ochmann B, Zatoń M. Determination of optimal load in the Wingate Anaerobic Test is not depend on number of sprints included in mathematical models. Front Physiol. 2023;14(May):1–8.

61. Staron RS, Hagerman FC, Hikida RS, Murray TF, Hostler DP, Crill MT, et al. Fiber type composition of the vastus lateralis muscle of young men and women. J Histochem

Cytochem. 2000;48(5):623–9.

62. Hicks AL, Kent-Braun J, Ditor DS. Sex differences in human skeletal muscle fatigue. Exerc Sport Sci Rev. 2001;29(3):109–12.

63. Froese EA, Houston ME. Performance During the Wingate Anaerobic Test and Muscle Morphology in Males and Females. 1987;8:35–9.

64. Hubner-Wozniak E, Kosmol A, Lutoslawska G, Bem E. Anaerobic performance of arms and legs in male and female free style wrestlers. J Sci Med Sport. 2004;7:473–80.

65. Lovell D, Mason D, Delphinus E, Eagles A, Shewring S, McLellan C. Does upper body strength and power influence upper body wingate performance in men and women? Int J Sports Med. 2011;32(10):771–5.

66. Sandbakk Ø, Strøm G, Holmberg H-C. Sex Differences in World Record Performance: The Influence of Sport Discipline and Competition Duration. Jsep [Internet]. 2017;28(2):588–95. Available from: https://research.rug.nl/files/30892751/Den_Hartigh_et_al._Short_and_long_term_PM_JSEP_accepted_version_Pure.pdf

67. Moon JR. Role of muscle mass on sprint performance: Gender differences? by Jorge Perez-Gomez, German Vicente Rodriguez, Ignacio Ara, Hugo Olmedillas, Javier Chavarren, Juan Jose González-Henriquez, Cecilia Dorado and José A. L. Calbet. Eur J Appl Physiol. 2008;103(3):375.

68. Hegge AM, Bucher E, Ettema G, Faude O, Holmberg HC, Sandbakk Ø. Gender differences in power production, energetic capacity and efficiency of elite cross-country skiers during whole-body, upper-body, and arm poling. Eur J Appl Physiol. 2016;116(2):291–300.

69. Ogles BM, Masters KS. A typology of marathon runners based on cluster analysis of motivations. J Sport Behav. 2003;26(1):69–86.

70. Hegge AM, Myhre K, Welde B, Holmberg HC, Sandbakk Ø. Are gender differences in

upper-body power generated by elite cross-country skiers augmented by increasing the intensity of exercise? PLoS One. 2015;10(5):1–16.

71. Bartolomei S, Grillone G, Di Michele R, Cortesi M. A comparison between male and female athletes in relative strength and power performances. J Funct Morphol Kinesiol. 2021;6(1).

72. Kraaijenbrink C, Vegter R, de Groot S, Arnet U, Valent L, Verellen J, et al. Biophysical aspects of handcycling performance in rehabilitation, daily life and recreational sports; a narrative review. Disabil Rehabil. 2021;43(24):3461–75.

73. Bini RR, Diefenthaeler F, Mota CB. Fatigue effects on the coordinative pattern during cycling: Kinetics and kinematics evaluation. J Electromyogr Kinesiol. 2010;20(1):102–7.

74. Bini RR, Diefenthaeler F. Kinetics and kinematics analysis of incremental cycling to exhaustion. Sport Biomech. 2010;9(4):223–35.

75. Kraaijenbrink C, Vegter RJK, Hensen AHR, Wagner H, Van Der Woude LHV. Different cadences and resistances in submaximal synchronous handcycling in able-bodied men: Effects on efficiency and force application. PLoS One. 2017;12(8):1–15.

76. Kraaijenbrink C, Vegter RJK, Hensen AHR, Wagner H, Van Der Woude LHV. Biomechanical and physiological differences between synchronous and asynchronous low intensity handcycling during practice-based learning in able-bodied men. J Neuroeng Rehabil. 2020;17(1):1–13.

77. Der W, Jamnik VK, Bredin SSD and GN (PAR-QC. 2021 Par-Q +. Heal &Amp; Fit J Canada [Internet]. 2021;14(1):83–7. Available from: https://hfjc.library.ubc.ca/index.php/HFJC/article/view/351/370

78. Oldfield RC. The assessment and analysis of handedness: The Edinburgh inventory. Neuropsychologia. 1971;9(1):97–113.

79. Nath T, Mathis A, Chen AC, Patel A, Bethge M, Mathis MW. Using DeepLabCut for

3D markerless pose estimation across species and behaviors. Nat Protoc. 2019;14(7):2152–76.
80. Cohen J. A power primer. Am Psychol Assoc. 1992;112(1):155–9.
81. Mukaka MM. Statistics corner: A guide to appropriate use of correlation coefficient in medical research. Malawi Med J. 2012;24(3):69–71.
82. Vanderthommen M, Francaux M, Johnson D, Dewan M, Lewyckyj Y, Sturbois X. Measurement of the power output during the acceleration phase of all-out arm cranking exercise. Int J Sports Med. 1997;18(8):600–6.
83. Richmond SR, Whitman SA, Acree LS, Olson BD, Carper MJ, Godard MP. Power output in trained male and female cyclists during the wingate test with increasing flywheel resistance. J Exerc Physiol Online. 2011;14(5):46–53.
84. Vargas NT, Robergs RA, Klopp DM. Optimal loads for a 30-s maximal power cycle ergometer test using a stationary start. Eur J Appl Physiol. 2015;115(5):1087–94.
85. Heller J, Kinkorova I, Vodicka P, Nikolaidis PT. Force – Velocity Relationship in Cycling and Arm Cranking : A Comparison of Men and Women. 2023;
86. Janssen I, Heymsfield SB, Wang ZM, Ross R. Skeletal muscle mass and distribution in 468 men and women aged 18-88 yr. J Appl Physiol. 2000;89(1):81–8.
87. Haizlip KM, Harrison BC, Leinwand LA. Sex-based differences in skeletal muscle kinetics and fiber-type composition. Physiology. 2015;30(1):30–9.
88. Tarnopolsky MA. Gender Differences in Metabolism; Nutrition and Supplements Mark. J Sci Med Sport. 2000;287–98.
89. Jaafar H, Rouis M, Coudrat L, Attiogbe E, Vandewalle H, Driss T. Effects of load in Wingate Test performances and reliability. J strength Cond Res. 2014;28(12):3462–8.
90. Zaccagni L, Toselli S, Bramanti B, Gualdi-Russo E, Mongillo J, Rinaldo N. Handgrip strength in young adults: Association with anthropometric variables and laterality. Int J Environ Res Public Health. 2020;17(12):1–12.

91. Glenmark B, Nilsson M, Gao H, Gustafsson JÅ, Dahlman-Wright K, Westerblad H. Difference in skeletal muscle function in males vs. females: Role of estrogen receptor-β. Am J Physiol - Endocrinol Metab. 2004;287(6 50-6):1125–31.
92. Van Drongelen S, van den Berg J, Arnet U, Veeger DJ, van der Woude LHV. Development and validity of an instrumented handbike: Initial results of propulsion kinetics. Med Eng Phys. 2011;33(9):1167–73.
93. Lanza IR, Towse TF, Caldwell GE, Wigmore DM, Kent-Braun JA. Effects of age on human muscle torque, velocity, and power in two muscle groups. J Appl Physiol. 2003;95(6):2361–9.
94. Strömbäck E, Aasa U, Gilenstam K, Berglund L. Prevalence and Consequences of Injuries in Powerlifting: A Cross-sectional Study. Orthop J Sport Med. 2018;6(5):1–10.
95. Roberts BM, Nuckols G, Krieger JW. Sex Differences in Resistance Training: A Systematic Review and Meta-Analysis. J Strength Cond Res. 2020;34(5):1448–60.
96. Stone B, Mason BS, Warner MB, Goosey-Tolfrey VL. Shoulder and thorax kinematics contribute to increased power output of competitive handcyclists. Scand J Med Sci Sport. 2019;29(6):843–53.
97. Halloran KM, Focht MDK, Teague A, Peters J, Rice I, Kersh ME. Moving forward: A review of continuous kinetics and kinematics during handcycling propulsion. J Biomech. 2023;159.
98. Szczygieł E, Fudacz N, Golec J, Golec E. The impact of the position of the head on the functioning of the human body: A systematic review. Int J Occup Med Environ Health. 2020;33(5):559–68.
99. Aruin AS, Latash ML. Directional specificity of postural muscles in feed-forward postural reactions during fast voluntary arm movements. Exp Brain Res. 1995;103(2):323–32.

100. Hiraoka K, Taniguchi Y. The effect of immobilization of the head and trunk on arm-cycling-induced depression of soleus motoneuron pool excitability. Somatosens Mot Res. 2010;27(1):28–33.

101. Offley C, Moody J, Esformes J, Byrne P. Effects of head position on back squat power performance. Eur J Hum Mov. 2021;46(2018):66–74.

102. Bini RR, Senger D, Lanferdini F, Lopes AL. Joint kinematics assessment during cycling incremental test to exhaustion. Isokinet Exerc Sci. 2012;20(2):99–105.

103. Carmichael MA, Thomson RL, Moran LJ, Wycherley TP. The impact of menstrual cycle phase on athletes' performance: a narrative review. Int J Environ Res Public Health. 2021;18(4):1–24.

104. Colenso-Semple LM, D'Souza AC, Elliott-Sale KJ, Phillips SM. Current evidence shows no influence of women's menstrual cycle phase on acute strength performance or adaptations to resistance exercise training. Front Sport Act Living. 2023;5(March).

105. Dasa MS, Kristoffersen M, Ersvær E, Bovim LP, Bjørkhaug L, Moe-Nilssen R, et al. The Female Menstrual Cycles Effect on Strength and Power Parameters in High-Level Female Team Athletes. Front Physiol. 2021;12(February):1–9.

Figures.

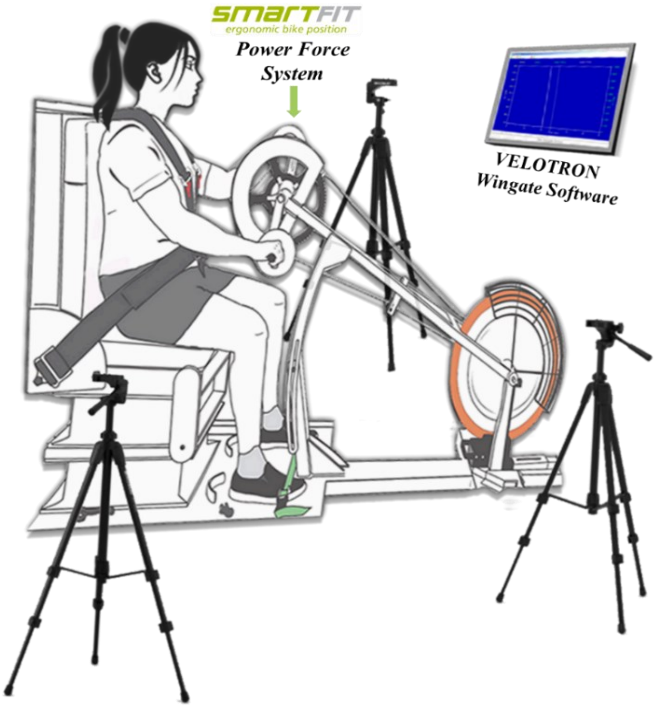

Figure 5. Experimental Setup

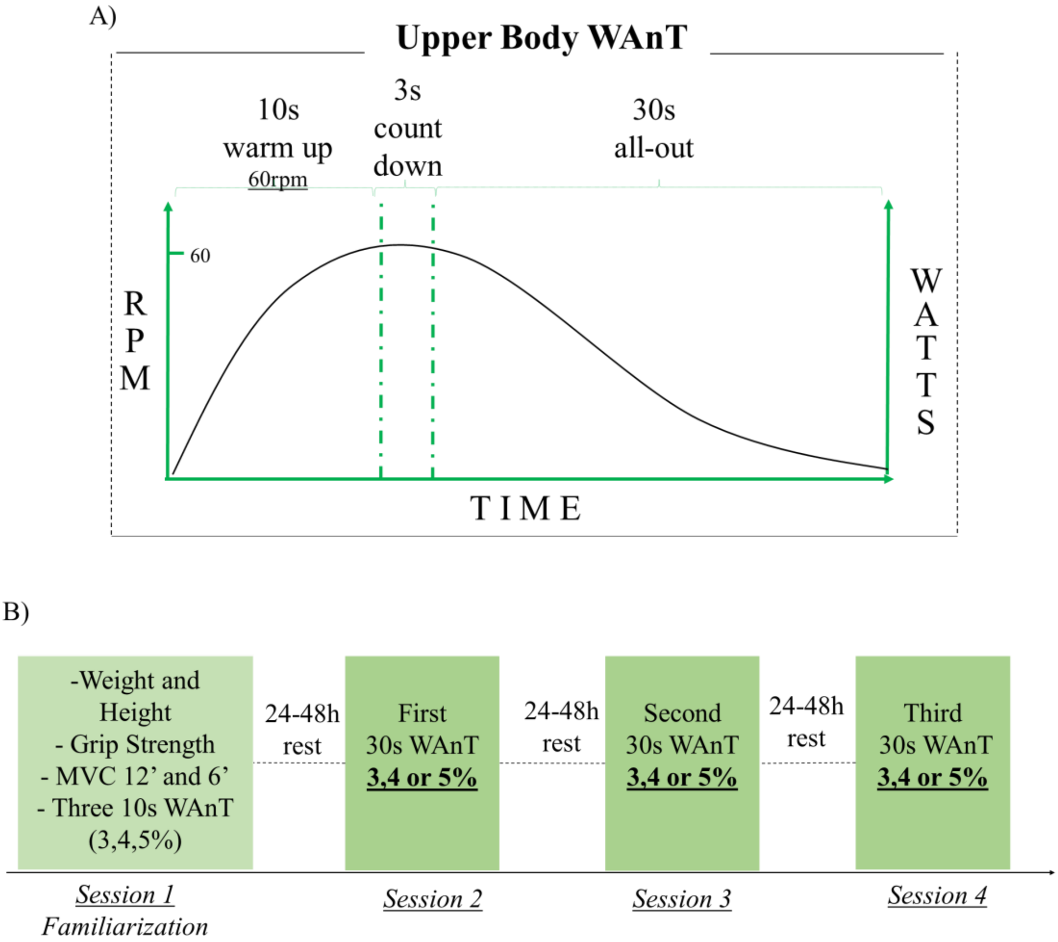

Figure 6. A) Upper body WanT protocol. B) Experimental Protocol. Sessions 2, 3, and 4 were randomized.

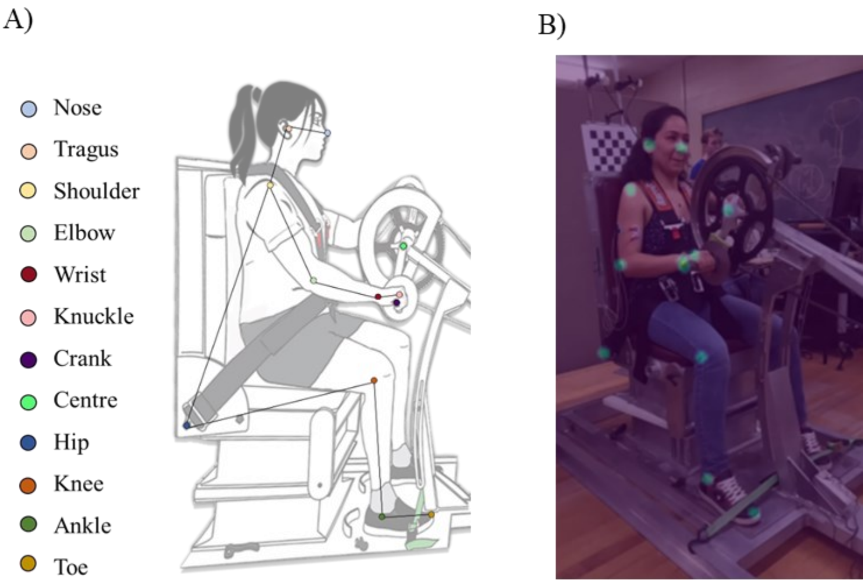

Figure 7. A) Labels selected in DeepLabCut for the markerless motion capture analysis. B)Visual representation of the test error: the green dots represent the tracking area of each label.

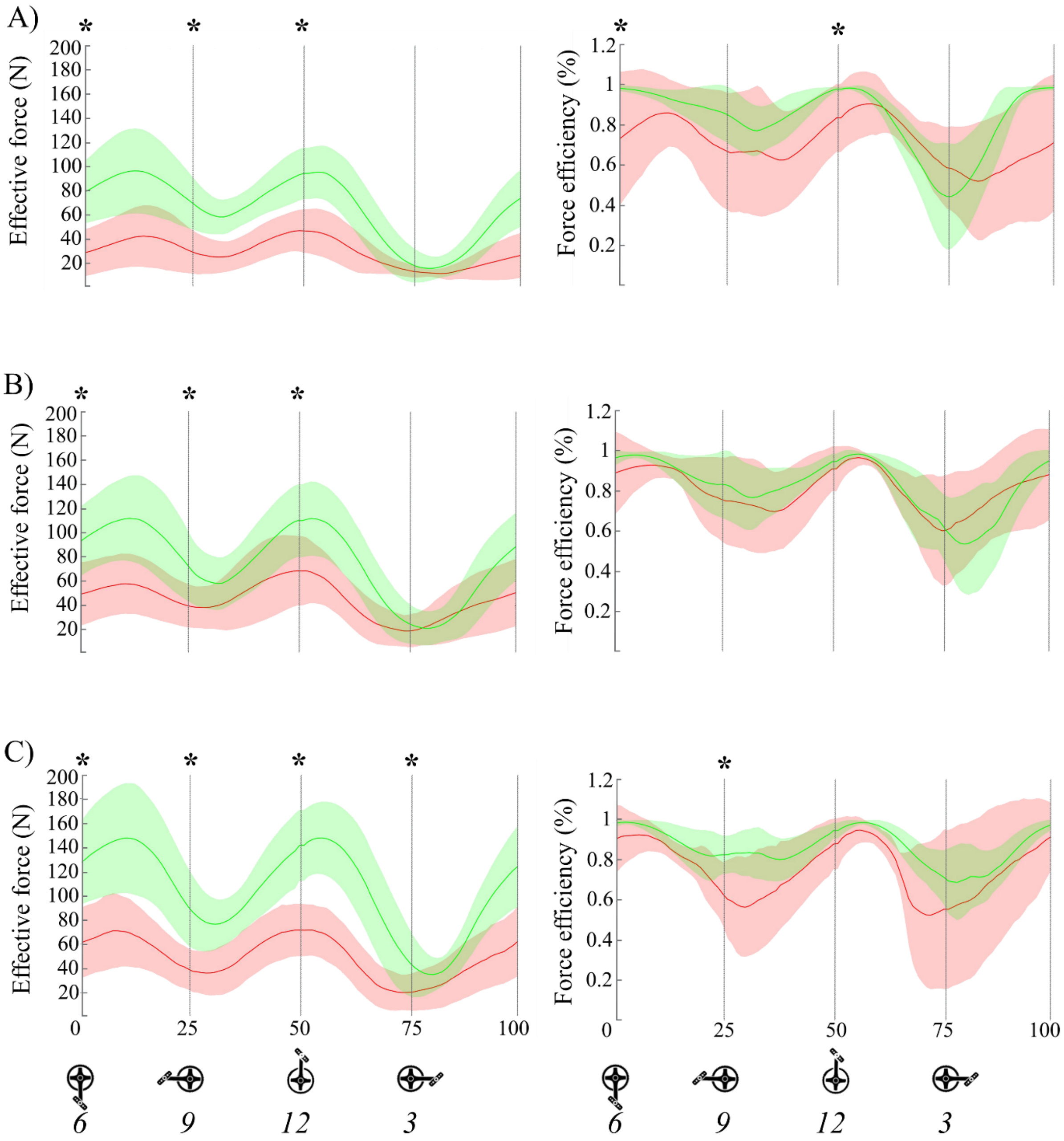

Figure 8. Sex differences in effective force and the corresponding force efficiency at A) 3% B) 4% and C) 5%. Females are represented in red and males in green. Means are shown as solid lines, and the SDs are shown as the matching colour shadowed areas about each mean. Symbol * represents significant differences (p <.05) between sexes at each position for each load. The X-axis represents a complete cycle from 0-100%; under the axis, the crank graphic represents the cycle positions, starting at the 6 o'clock. The Y-axis represents the force in newtons for effective and for force efficiency represents efficiency from 0-1.2; 0=0% efficiency and 1.2=120% efficiency.

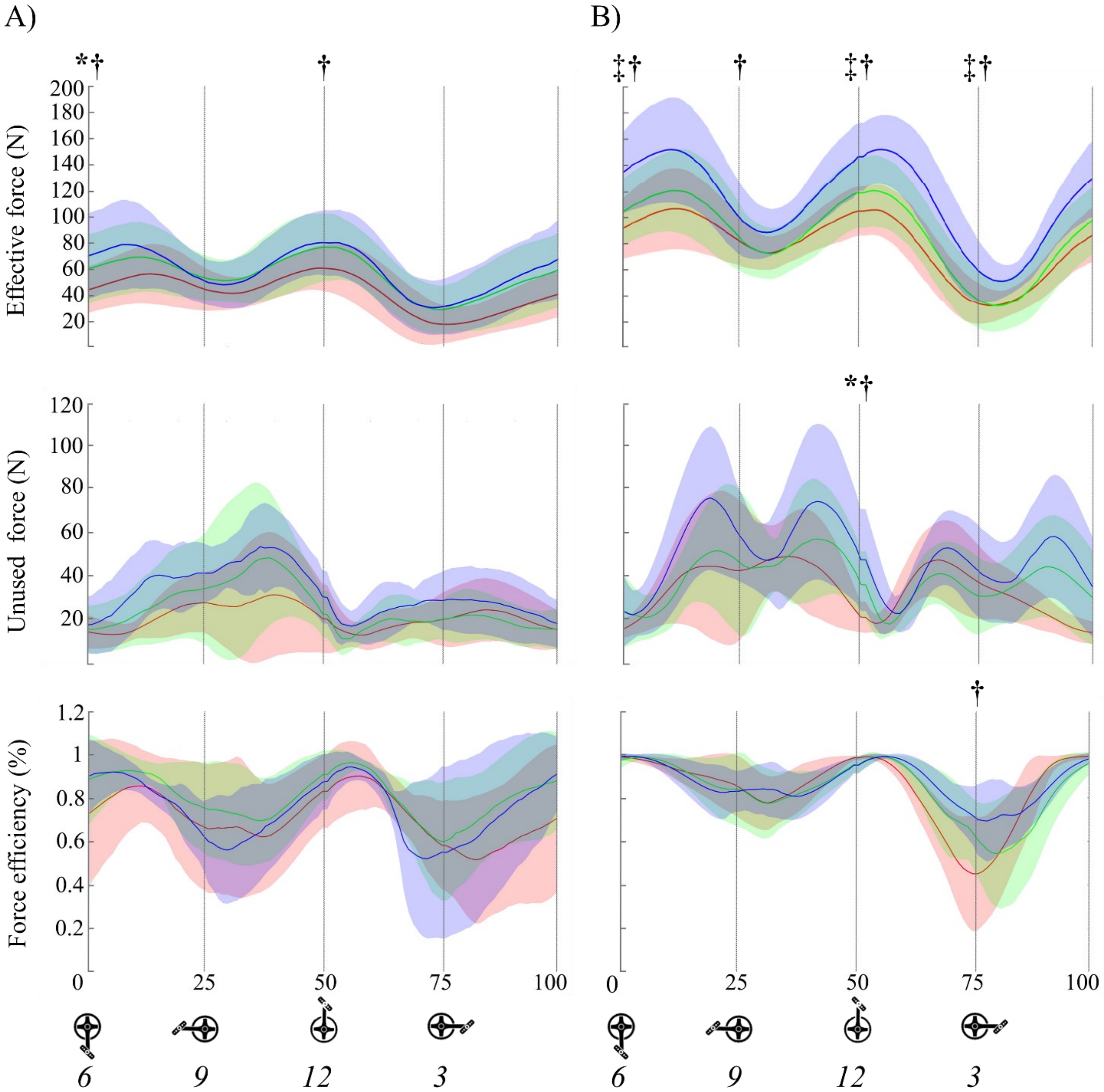

Figure 9. Effects of different loads on the crank forces for A) females and B) males. The different loads are represented in the figure as Red 3%, Green 4%, and Blue 5%. Means are shown as solid lines, and the SD are shown as the matching color shadowed areas about each mean. Significant differences are reported as ($p < .05$). Symbols represent significant differences between loads: *3 vs 4%, ‡4 vs 5%, and †3 vs 5. The X-axis represents a complete cycle from 0-100%; under the axis, the crank graphic represents the cycle positions, starting at the 6 o'clock. The Y-axis represents the force in newtons for effective and unused force; the force efficiency index represents efficiency from 0 - 1.2 (0= 0% efficiency and 1.2=120% efficiency).

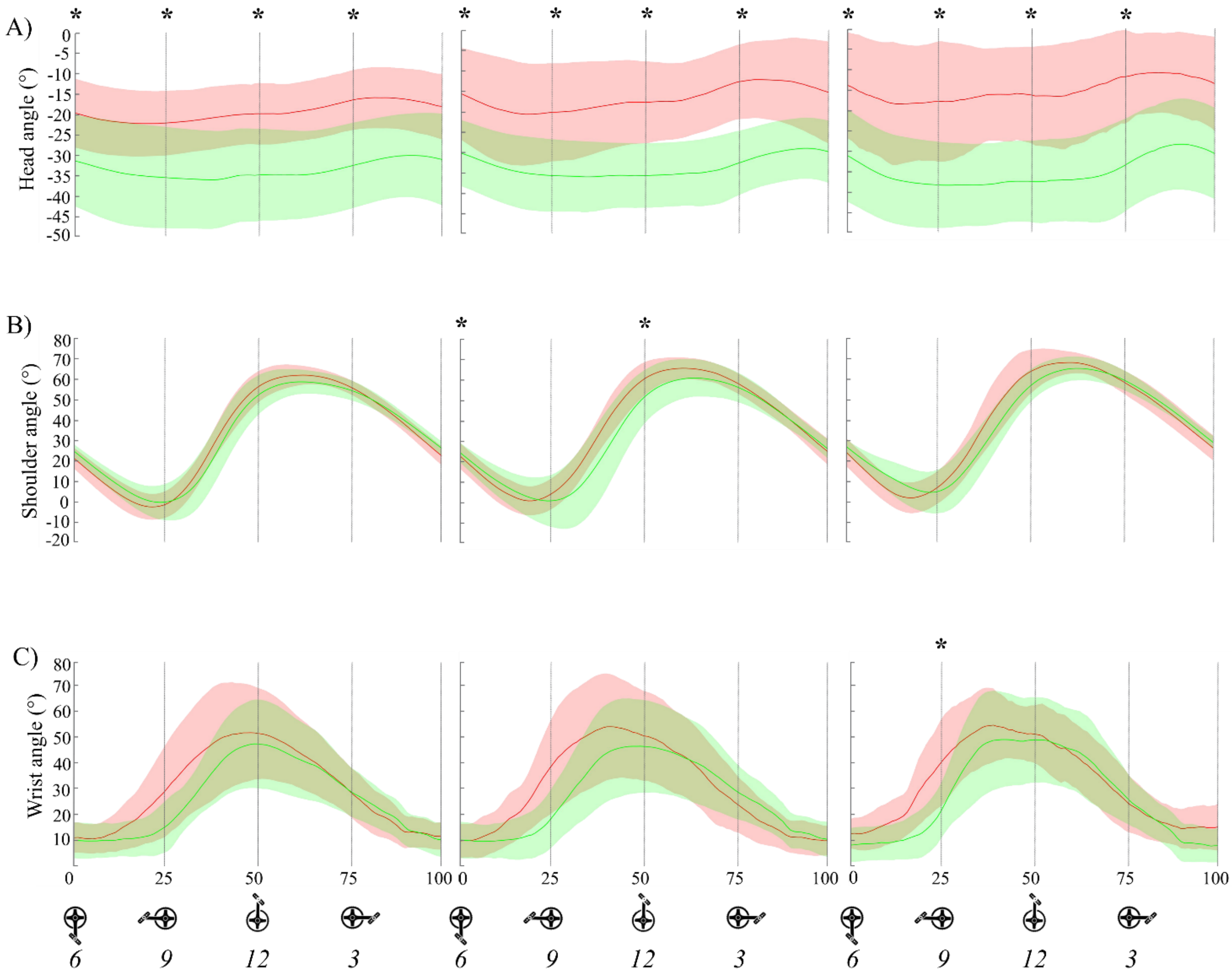

Figure 10. Sex differences in the kinematic variables for the three loads. Red= females and Green=males. The left column represents joint angle changes at 3%, the middle column at 4% and the right column at 5%. A) Neck angle: negative values represent neck flexion B) Shoulder angle: positive values represents shoulder flexion and negative values represent shoulder extension. C) Wrist angle: positive values correspond to wrist extension. Means are shown as solid lines, and the SD is shown as the matching color-shadowed areas about each mean. Symbol * represents significant differences ($p < .05$). The X-axis represents a complete cycle from 0-100%; under the axis, the crank graphic represents the cycle positions, starting at the 6 o'clock. The Y-axis represents degrees of flexion-extension.

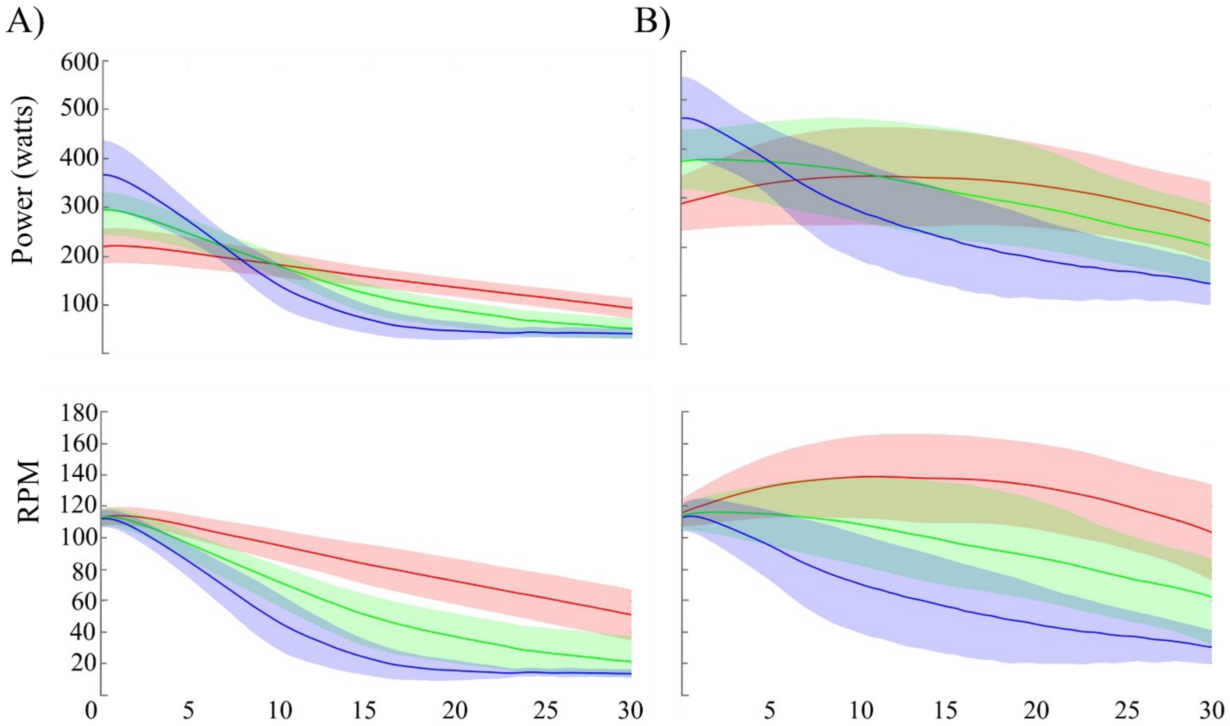

Figure 11. Power and RPM for A) females and B) males . The different loads are represented in the figure as Red 3%, Green 4%, and Blue 5%. Means are shown as solid lines, and the SD is shown as the matching color shadowed areas about each mean. The X-axis represents the duration of the Wingate test from 0 to 30 seconds. The Y-axis represents power in watts and revolutions per minute (RPM).

Appendix A: Ethical cleareance.

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:	**20230904-HK**
Approval Period:	November 17, 2022 – November 30, 2023
Funding Source:	
Responsible Faculty:	Dr. Duane Button School of Human Kinetics and Recreation
Title of Project:	*Does a bilateral deficit exist in arm cycling and is it task-dependent?*

November 17, 2022

Mr. Philip Edwards
School of Human Kinetics and Recreation
Memorial University

Dear Mr. Edwards:

Thank you for your correspondence addressing the issues raised by the Interdisciplinary Committee on Ethics in Human Research (ICEHR) for the above-named research project. ICEHR has re-examined the proposal with the clarifications and revisions submitted, and is satisfied that the concerns raised by the Committee have been adequately addressed. In accordance with the *Tri-Council Policy Statement on Ethical Conduct for Research Involving Humans (TCPS2)*, the project has been granted *full ethics clearance* for **one year**. 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 concerns, 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 **November 30, 2023**. 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 Drover, Ph.D.
Vice-Chair, Interdisciplinary Committee on
Ethics in Human Research

JD/bc

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