AN INVESTIGATION INTO THE EFFECTS OF CHANGES IN MUSCLE FORCE AND MUSCLE LENGTH ON LOCALISED MUSCLE FATIGUE

BY

MIRIAM CHRISTINA MATTISON

(ORCID – 0000-0002-6133-5328)

DISSERTATION

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Department of Human Kinetics and Ergonomics

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ABSTRACT

Introduction: Localised muscle fatigue poses significant challenges to human well-being and performance. However, despite decades of research into muscle fatigue, numerous questions remain unanswered about how such fatigue develops and whether its development and recovery differ between submaximal static and dynamic exertions. The task-dependency principle is known to influence the development of muscular fatigue, with force, range of motion and movement speeds influencing motor unit recruitment strategies. There is, however, a paucity in the literature on how these task-related factors interact with one another.

Purpose: This dissertation investigated muscle fatigue under different muscle exertions by applying a novel system for classifying static and dynamic exertions consisting of combinations of muscle length and muscle force criteria. It was hypothesised that the magnitude of localised muscle fatigue is subject to variations (or lack thereof) in the force generated by the muscle, as well as variations in muscle length as a limb moves through a range of motion.

Methods: Thirty-six (36) healthy student volunteers participated in this empirical laboratory-based study by performing an elbow flexion/extension exercise protocol designed to fatigue the elbow flexor muscles under three different exertion types on different days:

1) *"Varying Length"* - participants moved a constant external load (25% of the force generated by their isokinetic maximum voluntary exertions (MVE)) through a range of motion of 80 degrees and at a set takt of 2.6 seconds per cycle,
2) *"Pure Static"* - this condition entailed no changes in muscle length or force. The elbow flexors had to resist a constant external force corresponding to 25% of the isometric MVE while the elbow flexion angle was set at 90^{o},
3) *"Varying Force"* - the elbow joint angle was fixed at 90^{o}, but the force generated by the elbow flexors fluctuated between 15% and 35% of their isometric MVEs and according to the same takt as the "Varying Length" condition.

Numerous variables acting as fatigue indicators were recorded before, during, and after the fatigue protocol. Peak torque, time to peak torque, work and power were recorded via isokinetic dynamometry during maximum voluntary exertions before the fatigue protocol, just after its termination, and every minute for the next five minutes during the subsequent recovery period. During the submaximal fatigue protocol itself, EMG of biceps brachii, as well as performance accuracy and variability for cycle time, force, joint angles, and accelerations, were measured during the submaximal exertions at the start of and just before termination of the submaximal fatigue protocol. The exercise protocol was terminated when participants reached a Rating of Perceived Exertion of 9 on the CR-10 scale, and the time to termination was recorded.

Analyses using General Linear Models (GLM) were performed to determine the effects of time, as well as exertion type (i.e., “Varying Length”, “Pure Static”, and “Varying Force”) on muscle fatigue, as well as the recovery thereof. The “Pure Static” condition was used as a baseline measurement against which responses of the “Varying Length” and “Pure Static” conditions were compared. Significant differences were identified at $p<0.05$, and Tukey post hoc analyses determined differences within the main effects (time and condition) and their interactions.

Results: The statistical analyses revealed that:

1) The exercise protocol induced localised muscle fatigue of the elbow flexors. This is supported by significant time-related changes in most fatigue parameters calculated during the maximum force exertions before and after the protocol (peak torque, Fatigue Index, work, power), as well as during the submaximal exercise protocol (EMG amplitude, median frequency, and performance accuracy). Conversely, significant improvements were found during the recovery period, although not all variables had returned to pre-fatigue levels after five minutes of rest. Unexpected findings included an unchanged time to peak torque throughout the exercise protocol but significant improvements after rest, as well as greater joint steadiness.

2) Variations in muscle length significantly affected fatigue development. The “Varying Length” condition largely showed greater fatigue for endurance time, number of cycles, peak torque, the Fatigue Index, EMG median frequency, and force accuracy, but not for the time to peak torque, EMG amplitude and cycle-to-cycle variability for force under the “Varying Length” condition, which was lower than under the “Pure Static condition. Similar outcomes were observed during the recovery period. Separation of the concentric from the eccentric movement

phases of the “Varying Length” condition revealed a more significant contribution of the concentric muscle actions to fatigue compared to the eccentric exertion phase. All variables that demonstrated significant decrements over the duration of the fatigue protocol showed improvements over the recovery period. The “Varying Length-Concentric” exertion also experienced a significantly larger proportional increase in the time to peak torque over the recovery period.

3) Variations in muscle force had no statistically significant impact on the fatigue responses (endurance time, number of cycles, peak torque, time to peak torque, Fatigue Index, EMG amplitude and median frequency, and cycle-to-cycle variability). Only performance accuracy under the “Varying Force” condition revealed significantly greater deviations from the target joint angle compared to the “Pure Static” condition.

Discussion: It was anticipated that dynamic movements, either via changes in muscle length or muscle force, would actively promote blood flow to the working muscles and, therefore, enhance fatigue resistance. The data, however, revealed significantly greater fatigue for the "Varying Length” exertion, compared to the “Pure Static” condition, and no differences in the fatigue responses between the “Pure Static” and “Varying Force” exertions. These unexpected findings can likely be attributed to a variety of physiological and methodological factors. Firstly, the mean loading of 25% of maximum voluntary exertion may have been too low to induce effective occlusion of blood flow, so even though the variations in muscle length and force may have had a peristaltic effect, the impact on muscle fatigue was negligible. Furthermore, it is plausible that even though the external loads manipulated had been relativised to each exertion type’s maximum strength capacity, the true peak torque was not obtained during the maximum exertions prior to the protocol, thereby creating a larger proportional workload for the “Varying Length-Concentric” exertion, but a lower workload for the “Varying Length-Eccentric) exertion. Additionally, as per the length-tension relationship, the internal effort that had to be generated by the elbow flexor muscles under the “Varying Length” condition fluctuated as the forearm moved through its range of motion, thereby placing a greater demand on the metabolic and energetic processes during the concentric movement phases and less under the eccentric and isometric exertions. One important methodological consideration relates to the challenges of comparing different exertion types with one another since they

exhibit very different characteristics with regard to their motor strategies, particularly as fatigue develops, and may, therefore, be likened to comparing “apples with pears”.

Conclusion: The study concluded that tasks exhibiting variations in muscle length and muscle force do not necessarily enhance fatigue resilience. Given the different mechanical and physiological mechanisms between exertion types and the associated methodological challenges of matching workloads, further research into the effects of task-dependent factors on muscle fatigue is recommended.

Keywords: *Localized muscle fatigue, isometric, isotonic, concentric, eccentric, muscle actions*

DEDICATION

This dissertation is dedicated to my late mother, Gabriele Renz, … you always knew this day would come, and I know you are proud …

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DECLARATION

I hereby declare that the work presented in this dissertation:

i) is submitted for examination for the degree of Doctor of Philosophy;

ii) is an original report of my research composed during my registration at Rhodes University;

iii) has not been submitted for examination at any other university or for any other degree or professional qualification;

iv) is entirely my own, except where stated otherwise by reference or acknowledgement;

v) encountered no conflicts of interest;

vi) was conducted according to the highest ethical standards governing research.

Signed: Miriam Mattison.

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CHAPTER 1: INTRODUCTION

1.1 BACKGROUND TO THE STUDY

It is well-established that fatigue, in general, is a complex phenomenon with a multitude of causal factors, interactions, and outcomes (Hunter *et al.*, 2004; Barry & Enoka, 2007; Tornero-Aguilera *et al.*, 2022; Behrens *et al.*, 2023). While fatigue can manifest as a general whole-body phenomenon and include sleepiness, cardiovascular and cognitive tiredness, or exhaustion (Ream & Richardson, 1996; Skau *et al.*, 2021), localised muscle fatigue (LMF) differs from general fatigue in that it affects muscle functioning, resulting in a reduced force production (Chaffin, 1973; Fitts, 1994; Enoka & Duchateau, 2008), an increased perception of the effort required to produce the required force (Enoka & Stuart, 1992), as well as alterations in spatial and temporal movement characteristics (Côté *et al.*, 2002). LMF has long been acknowledged to consist of a multitude of central and peripheral processes (Vøllestad, 1997; Noakes, 2000; Seghers & Spaepen, 2004; Marcora *et al.*, 2009; Tornero-Aguilera *et al.*, 2022) and already begins at the start of the activity (Bigland-Ritchie & Woods, 1984; Jensen *et al.*, 2000; Barry & Enoka, 2007), is magnified throughout the activity (Enoka & Duchateau, 2008), and can eventually lead to failure of the contractile mechanism (Fitts, 1994).

With the focus of this dissertation being on localised muscle fatigue, it is important to mention that LMF is not only a concern amongst sports physiologists because of its effects on athletic performance (Noakes, 2000; Dambroz *et al.*, 2022); muscle fatigue also has relevance in occupational settings, daily activities, and medicine and rehabilitation (Chaffin, 1973; Behrens *et al.*, 2023) due to its impact on people's productivity and well-being. Well-cited short-term effects of localised muscle fatigue on the worker include increased perceived effort of work and localised discomfort (Radwin *et al.*, 2002), loss of motor control (Kumar, 2001), reduced proprioception (Voight *et al.*, 1996), and lower strength output (Seghers & Spaepen, 2004), while long-term effects may include reduced perceptions of work satisfaction and quality of life (Konz, 1998; Radwin *et al.*, 2002; Gorelick *et al.*, 2003; Rashedi & Nussbaum, 2015). For the employer, LMF may result in poor quality of work from their employees (Konz, 1998; Rashedi & Nussbaum, 2015) and increase the risk of accidents and injuries (Kumar, 2001; Gorelick *et al.*, 2003; Rashedi & Nussbaum, 2015). Localised fatigue has even been hypothesised to be causally related to the development of musculoskeletal disorders (Armstrong *et al.*, 1993; Radwin *et al.*, 2002; Seghers & Spaepen, 2004; Rashedi & Nussbaum, 2015; Gallagher & Schall, 2017). Therefore, it is in the interest of both employers and employees that localised muscle fatigue is well understood so that suitable work practices and routines can be developed (Seghers & Spaepen, 2004).

1.2 LIMITATIONS IN CURRENT FATIGUE RESEARCH

In 2016, fatigue researchers Enoka and Duchateau highlighted that despite the vast amount of academic literature on human fatigue, relatively little was still known about the intricacies of localised muscle fatigue. Decades of research into the topic of LMF have resulted in a considerable amount of knowledge about the adverse effects of muscle fatigue on workers' well-being and work performance, its causal mechanisms, as well as influencing factors. Despite this, little of this knowledge has translated into effective practical advice to reduce muscle fatigue (Enoka & Duchateau, 2016).

One possible reason for the lack of practical impact may be that the complexity of different force exertions has limited the research conducted on muscle fatigue, since the physiological pathways that result in muscle fatigue are influenced by a large variety of individual and task-specific factors. From a mechanical point of view, the purpose of skeletal muscles is to convert chemical energy into mechanical energy, thereby generating force, which is transferred to the bones to which they are attached (McCuller *et al.*, 2023), and in doing so either stabilising or moving a limb or body segment. Motor units within the muscle are responsible for generating the tension, but must be recruited in a specific manner to generate the appropriate amount of force for the activity. The characteristics of the task executed play a significant role in determining how motor units are recruited and, therefore, how a muscle adapts to a fatiguing activity; this is referred to as the principle of "task-dependency" (Enoka & Stuart, 1992; Enoka, 1995).

Task-related factors such as the intensity of the exertion, its duration, the speed of movement, and the type of muscle exertion (e.g., isometric, concentric or eccentric) influence central and peripheral mechanisms involved in the motor unit recruitment and the excitation-contraction coupling mechanisms (Enoka & Stuart, 1992; Bigland-

Ritchie *et al.*, 1995; Enoka, 1995; Iridiastadi & Nussbaum, 2006). Although it is well-known that muscle fatigue develops sooner in static / isometric muscle exertions (i.e., exertions that are devoid of changes in muscle length and force) than in dynamic (non-isometric) ones (Masuda *et al.*, 1999), surprisingly little research has focussed on more complex tasks, such as those involving dynamic muscle actions and/or intermittent exertions (James *et al.*, 1995; Nussbaum, 2001; Iridiastadi & Nussbaum, 2006; Xia & Frey Law, 2008). Even though most activities, including industrial tasks, are dynamic, rather than static in nature (Smith, 2001), the majority of studies have investigated muscle fatigue in simplified scenarios such as during static muscle exertions, and with strict control over muscle length and force magnitude (Jonsson, 1988; Potvin, 1997; Potvin & Bent, 1997; Kroemer, 1999; Masuda *et al.*, 1999; Krüger *et al.*, 2019). This may be because dynamic exertions are characterized by a multitude of factors which could influence the mechanical and physiological processes leading to muscle fatigue. As a result, the current understanding of muscular fatigue is predominantly based on studies that have focussed on isometric exertions (Kroemer, 1999; Krüger *et al.*, 2019), and most fatigue models hence do not factor in the complexities associated with dynamic muscle actions, such as the length-tension and force-velocity relationships (Potvin, 1997; Cheng & Rice, 2005; González-Izal *et al.*, 2012; Rashedi & Nussbaum, 2015; Krüger *et al.*, 2019), and which in turn influence the fatigue processes at a muscular level. Similarly, the processes that reverse the effects of fatigue also appear to depend on numerous factors, including the history of muscle loading, such as duration, intensity and modality, and the resultant contributions of central and peripheral factors to the fatigue development process (Rashedi & Nussbaum, 2015; Carroll *et al.*, 2017). It is, therefore, reasonable to assume that if the rate and magnitude of localised muscle fatigue development are influenced by different exertion types, the reversal of the effects of muscle fatigue could also differ between different muscle actions.

In addition to the relatively limited understanding of muscle fatigue under dynamic muscle actions, a further limitation of the current state of muscle fatigue research is that most exertions occur at submaximal levels, rather than at maximal or near-maximal levels. Muscle fatigue development under sub-maximal conditions requires further attention, as such exertion levels are particularly relevant in daily activities, sporting, and occupational contexts. For example, work practices in industrial contexts consist of highly repetitive motions at low to moderate efforts (Bigland-Ritchie *et al.*, 1986a; Nussbaum, 2001; Sjøgaard & Jensen, 2006; González-Izal *et al.*, 2012; Rashedi & Nussbaum, 2015). Jobs characterised by low-level dynamic exertions in the literature include, but are not limited to, office work (typewriting, mouse work), industrial sewing, letter sorting, dental work, and meat cutting (Sjøgaard *et al.*, 1988; Sjøgaard & Jensen, 2006). Despite the low force requirements of such tasks, localised muscle fatigue still develops, potentially leading to some, or all, of the aforementioned adverse outcomes (Sjøgaard & Jensen, 2006).

In conclusion, “the assessment of isometric force alone can give an incomplete or misleading interpretation of the overall consequences of fatigue” (Krüger *et al.*, 2019, p.1), and further systematic investigations of the effects of the interactions of the task-related factors on localized muscle fatigue and its recovery are required (Enoka and Stuart, 1992), in particular during low-level and dynamic exertions.

1.3 STATEMENT OF THE PROBLEM AND AIM OF THE STUDY

In various contexts, including the workplace, muscle fatigue can result in a multitude of problems for the employee and the employer, even if the activities performed are intermittent and at a low intensity. The task-dependency principle highlights that the physiological and mechanical processes that result in fatigue development depend on the nature of the activity. Since most daily tasks are dynamic (González-Izal *et al.*, 2012), it is important to understand muscle fatigue in more detail for exertions that are not static in nature and how different task-related factors interact to influence the fatigue process. However, relatively little is known about localised muscle fatigue development and its recovery under different exertion types, thus limiting the ability to suitably address this issue. The overarching aim of the study was to experimentally investigate the development of, and recovery from, localised muscle fatigue under different muscle exertions at submaximal levels. More specifically, and as detailed in subsequent chapters of this thesis, a novel classification for static and dynamic muscle exertions considering selected task-dependent factors was proposed, which was tested in this research.

CHAPTER 2: REVIEW OF LITERATURE

The literature relating to muscle fatigue is vast since systematic research into muscle physiology has been conducted since the late 19th century by Angelo Mosso, and the early 20th century by A.V. Hill (Barclay & Curtin, 2022; Di Giulio *et al.*, 2006). Scientific interest in muscle function and fatigue was rekindled in the 1980s (Gandevia *et al.*, 1995), and the bulk of research in this field has been published since then, with researchers such as De Luca, Duchateau, Enoka, Gandevia, Noakes, Marcora and Rohmert being just a few of the many researchers who have contributed to the field of muscle fatigue across the decades. In light of the extensive body of knowledge on localised muscle fatigue, the scope of this literature review has been limited to address topics most relevant to the aims of this study and which will assist the reader in understanding the research concept (Chapter 3) and associated methodology (Chapter 4). These topics entail an overview of muscle fatigue and the fundamentals of muscle functioning as they relate to muscle fatigue. Terminology used for different muscle exertions, sources of muscle fatigue and how muscle fatigue can be quantified are also discussed, followed by the effects of task-related factors on fatigue.

2.1 OVERVIEW OF MUSCLE FATIGUE

Muscle fatigue is a complex phenomenon, and despite there being a generalised consensus in the scientific community about certain aspects thereof, countless challenges remain. “Localised muscle fatigue” (LMF), also referred to as “exercise-induced muscle fatigue”, “neuromuscular fatigue”, or “motor performance fatigue”, is a transient event that interferes with the voluntary activation of a skeletal muscle or muscle group (Chaffin, 1973; Gandevia, 2001; Enoka & Duchateau, 2008; Behrens *et al.*, 2023), and results in reductions of a muscle’s maximal force-generating capacity (Iridiastadi & Nussbaum, 2006). More specifically, it is defined as an exercise-induced reduction, or even failure, in the ability of a muscle to maintain a required force or power (Edwards, 1981; Hagberg, 1981; Bigland-Ritchie & Woods, 1984; Søgaard *et al.*, 2006), or as an increased effort to maintain the desired force level (e.g. Gandevia, 1999; Marcora *et al.*, 2009; Boyas & Guével, 2011). Furthermore, it is important to highlight that these decrements are reversible with rest and are thus not the result of pathological muscle weakness or injury (ACSM, 2006; Zwarts *et al.*, 2008; Wan *et al.*, 2017). While earlier definitions highlighted fatigue as the point of task failure, it is now accepted that localised muscle fatigue is a dynamic and time-dependent process of the neuromuscular system that develops as soon as an activity is started and is, therefore, progressive (Bigland-Ritchie, 1984; Jørgensen *et al.*, 1988; Merletti & Parker, 2014; Søgaard *et al.*, 2006; Barry & Enoka, 2007; Enoka & Duchateau, 2008; Abd-Elfattah *et al.*, 2015).

Despite there being some agreement on what characterised muscle fatigue, "[w]e are [still] largely unable to state with certainty why an individual becomes fatigued under various conditions" (Enoka & Duchateau, 2008, p.11). Researchers agree that localised muscle fatigue does not occur due to a single mechanism but that it is the result of complex, dynamic and multi-factorial interactions, believed to involve physiological, biomechanical and psychological components (Fitts, 1994; Enoka, 1995; Seghers & Spaepen, 2004; Baptista *et al.*, 2009; Rashedi & Nussbaum, 2015; Behrens *et al.*, 2023). These mechanisms lead to a decline in the desired performance during diverse activities involving voluntary muscle force generation (Vøllestad, 1997). It was even suggested by Abd-Elfattah *et al.* (2015) that the mechanisms that cause fatigue change during fatigue development are the result of cascading physiological processes and various compensatory responses employed in an attempt to maintain a specific performance output.

2.2 MUSCLE FUNCTIONING

To understand the fatiguing processes within a muscle and the factors influencing these processes, it is important first to understand muscle function under unfatigued conditions. The purpose of muscular exertion is to produce tension within the muscle, which in turn is transferred to the bones to which the muscle attaches, thereby either stabilizing or moving an external resistance (Kroemer, 1999). A muscle's contractile process is, however, not restricted to the structural part of the muscle but is the result of several sequential processes that begin in the brain's motor cortex, where neural impulses are generated and sent via the motor neurons of the spinal cord to the contractile units of the relevant muscle(s) (Edwards, 1975; Bigland-Ritchie, 1984; Enoka, 1995; Gandevia, 2001; Zwarts *et al.*, 2008; Tortora & Derrickson, 2009). The *Cross-bridge* and *Sliding Filament* theories state that the action potentials sent from the central nervous system elicit muscle action potentials in all the skeletal muscle fibres with which it forms synapses, and in response to these action potentials, various physiological events within the muscle fibres (e.g., excitation-contraction coupling and cross-bridge cycling processes) cause the so-called z-discs to pull on neighbouring sarcomeres (Huxley, 2004; Konrad, 2005; Tortora & Derrickson, 2009). More specifically, the actin and myosin filaments that constitute the myofibrils within muscle fibres are pulled past one another in a process called the *powerstroke* (Tortora & Derrickson, 2009). The resultant tension is initially “absorbed” due to the stretching of the muscle’s elastic structures, such as the connective tissues surrounding the muscles and tendinous structures (Kapitaniak, 2001b; Faulkner, 2003). However, with increasing tension, this force is eventually transferred to the muscle’s attachment points on the respective bones, thus creating a counterforce to an externally applied force. Depending on the magnitude of the tension produced by the muscle, this force will ultimately lead to the movement of a limb or external resistance by accelerating and decelerating the load, or merely maintaining a limb’s position (Hall, 2007).

2.2.1 Motor Unit Recruitment

The process of generating a force within a muscle does not occur concurrently in all muscle fibres. Achieving a specific force via smooth muscle exertion requires selective stimulation and relaxation of the different motor units within a muscle (Tortora & Derrickson, 2009). Motor units are functional units within skeletal muscles, and each motor unit consists of a collection of muscle fibres that are all innervated by the same motor neuron (Floeter, 2001). Motor unit recruitment is a complex process and requires selective stimulation and relaxation of the different motor units within a muscle to achieve the desired force output (Floeter, 2001; Tortora & Derrickson, 2009). Recruitment of the muscle fibres, and therefore the muscle’s ability to produce force, depends on numerous characteristics, including their respective motor neurons’ cell body sizes, number of dendrites, surface area, and depolarization thresholds, amongst others (ACSM, 2006), but also the motor unit’s firing rate, the muscle fibre length and the movement velocity (Tsianos & Loeb, 2013). In short, the central nervous system must integrate various structural and functional factors with the continuous feedback received via the afferent neural pathways (Schmidt & Wrisberg, 2008; Madeleine, 2010). Motor control necessitates fine-tuning the motor programme, i.e. the set of ‘commands’ sent from the CNS to a muscle, or group of muscles, detailing the motor units to be recruited, the recruitment order, and the duration and frequency of recruitment to achieve the intended force (Schmidt & Wrisberg, 2008; Gardiner, 2011; Taylor *et al.*, 2016). Producing the appropriate force level is ultimately achieved via intricate patterns of two main control strategies, namely the spatial and temporal recruitment patterns of motor units (Kroemer, 2001; Konrad, 2005; Srinivasan & Mathiassen, 2012).

Spatial Recruitment

Spatial recruitment refers to the number of motor units activated to produce a force (Chaffin *et al.*, 2006; Gamet & Fokapu, 2008). The more motor units that are recruited simultaneously across the entire muscle, the higher the contractile strength exerted by that muscle (Kroemer, 2001). Larger motor units innervate more muscle fibres (Tsianos & Loeb, 2013) and are, therefore, able to produce greater force, such as those used for powerful movements. Conversely, smaller motor units contain fewer muscle fibres and are thus better for precise movement control (Tortora & Derrickson, 2009; Fukuda *et al.*, 2010). Within a particular motor unit, the skeletal muscle fibres tend to be of the same type (Tsianos & Loeb, 2013). Generally, small motor units are generally composed of slow-twitch fibres, meaning they are the least powerful, but also the most fatigue-resistant fibres due to their high mitochondrial enzyme content (Edwards, 1975; Fitts, 1994; Ma *et al.*, 2012). In contrast, larger motor units are stronger due to their high proportion of fast-twitch fibres (ACSM, 2006; Fitts, 1994). These fibres contain most myofibrils, have the largest action potential amplitude, can hydrolyse ATP quickly (Edwards, 1975; Fitts, 1994; Gamet & Fokapu, 2008; Tortora & Derrickson, 2009), and hence produce greater strength and power (ACSM, 2006); however, they also fatigue quickly (Hall, 2007).

Typically, not all motor units are recruited in unison, not even during maximal voluntary exertions, as they have different recruitment thresholds (Sjøgaard & Jensen, 2006). The different motor units in a muscle are recruited in a specific order, depending on the force requirements. According to *Henneman's Size Principle*, during an increasing voluntary muscle exertion, particularly an isometric one, smaller (slow-twitch) motor units are recruited first, with progressively larger motor units being added if the task requires more force (Henneman *et al.*, 1965; Milner-Brown *et al.*, 1973; Linnamo *et al.*, 2003). However, given the varying electrophysiological and morphological characteristics of motor units, the relationship between the number of motor units recruited and the resultant force outputs is not completely linear (ACSM, 2006). It is also acknowledged that under some circumstances, this order of motor unit recruitment may be altered, depending on the characteristics of the task (Gardiner, 2011), particularly in muscle actions with very rapid force exertions (ACSM, 2006). Nevertheless, such patterns of motor unit activity allow for more prolonged force exertions, and thus delay the performance decrements resulting from fatigue development within a muscle.

Temporal Recruitment

Temporal motor unit recruitment, also known as “rate coding”, “motor unit firing rate”, or “discharge rate”, refers to the number of action potentials arriving at the neuromuscular junction within a given timeframe (Gamet & Fokapu, 2008). A single impulse arriving at the motor unit creates a twitch contraction, while several consecutive action potentials create a temporal fusion of twitches, or “wave summation”, resulting in a more forceful muscle exertion until tetanus is achieved (Clamann, 1993). Enoka & Duchateau (2017) pointed out that rate coding plays a significant role in the force increase during rapid exertions, force control during dynamic exertions, and force maintenance during submaximal static exertions. Clamann (1993) also reported that a greater rate coding resulted in smoother force gradation compared with increased recruitment. In general, the motor unit firing rate of individual motor units increases with rising muscular effort until a maximum firing rate is reached (Maton, 1981). The minimum and maximum firing frequencies differ between motor units (ACSM, 2006).

Spatial and temporal recruitment are not independent of one another, and there exists a fine interplay between these two recruitment types to achieve the desired force (Clamann, 1993; Chaffin *et al.*, 2006; Duchateau & Baudry, 2014). Combinations of temporal and spatial patterns of motor unit activity also allow for more prolonged force exertions, but also delays in fatigue development in a muscle. The relative contributions of spatial and temporal recruitment to force production, as well as fatigue development, are, however, very much dependent on the muscle’s recruitment characteristics, which in turn depend on the task performed (Kukulka & Clamann, 1981).

2.3 TERMINOLOGY FOR MUSCLE EXERTIONS

Having described the processes underlying force generation in a muscle, it is important to clarify the terminology surrounding muscle exertions. Classifications of muscle actions have been in debate since the 1920s and 1930s (Faulkner, 2003), with many different terms in various disciplines having appeared in the literature since. The word "contraction" is most commonly used to denote a muscle's process of generating force. However, Knuttgen & Kraemer (1987) and Faulkner (2003) pointed out that the word "contraction" is flawed since it implies directionality and, more specifically, a reduction in muscle length. However, since muscle activations can also result in no changes or increases in muscle length, the terms "muscle exertion" or "muscle action" are increasingly replacing the term "contraction" and will thus be used as such in this dissertation.

Furthermore, the classification criteria of different muscle actions are limited and confusing, as they do not accommodate the more intricate subtleties and variations of exertions. Given that the focus in this thesis is on fatigue development during different muscle exertions, it is important to tease out the terminology relating to various muscle actions. More specifically, the principle of task-dependency highlights that the nature of the task, and therefore the corresponding characteristics of the muscle exertions, will determine the rate and extent of fatigue development as well as recovery (Barry & Enoka, 2007). To make sense of the various descriptors of muscle exertions, an investigation of the literature has revealed three broad categories of classifying muscle actions that can be described in terms of muscle length, the force produced, and movement speed.

2.3.1 Terminology based on Muscle Length

The most elementary classification of muscle actions commonly described in textbooks, are "dynamic" and "static" exertions, both of which refer to muscle length, and depend on whether the muscle actions result in an overt movement, or no movement, respectively (Knuttgen & Kraemer, 1987; McArdle *et al.*, 1996; Sjøgaard & Jensen, 2006; McLester & St Pierre, 2008; Klavora, 2010). If the tension generated by the muscle leads to limb movement, then the exertion is considered "dynamic", but if not, it is "static". However, the terms 'static' and 'dynamic' are ill-defined, which, given the complexities of muscle recruitment and their dependency on the action performed, can lead to very different outcomes relating to fatigue.

During static, or “isometric” (“iso” = same; “metro” = length) muscle exertions, there is no change in the muscle length (McArdle *et al.*, 1996; Tortora & Derrickson, 2009). The lack of change in muscle length is either because the individual is exerting a force against an immovable object, or because the moment generated by an external agent on a skeletal articulation is equalled by the counter-moment created by the muscle, with the result that the two opposing moments cancel each other out (Hall, 2007; Tortora & Derrickson, 2009; Schaefer & Bittmann, 2017). Although from a physiological perspective, actin and myosin do slide past one another during an isometric exertion and energy is still expended (Tortora & Derrickson, 2009), the resultant tension is initially “absorbed” due to the stretching of the muscle’s elastic structures, such as the connective tissues surrounding the muscles and tendinous structures (Kapitaniak, 2001b; Faulkner, 2003). However, with increasing tension, this force is transferred to the muscle’s attachment points on the respective bones, thus creating a counterforce to an externally applied force (Tortora & Derrickson, 2009).

Schaefer and Bittmann (2017) differentiate between two types of isometric exertions. During a “holding isometric muscle action” (HIMA), the external force produced by the muscle remains constant, while during a “pushing isometric muscle action” (PIMA), the muscle force can vary, despite the muscle length remaining unchanged (Schaefer & Bittmann, 2017). Both HIMA and PIMA can occur under conditions of maximal and sub-maximal effort (Knuttgen & Kraemer, 1987; Hall, 2007). In summary, during an isometric exertion, the overall length of the muscle remains constant, and no visible movement is created (Knuttgen & Kraemer, 1987; Kapitaniak, 2001b), irrespective of the force applied.

Dynamic exertions, on the other hand, produce movement and have also been referred to as “variometric” (referring to changing muscle length) by Kroemer (1999), “non-isometric” by Nussbaum (2001), or “anisometric” by Madeleine *et al.* (2001) and Madeleine (2010), where the prefix “aniso” translates to “not equal” (www.collinsdictionary.com). The prefix “auxo” could also be used but implies directionality as it means “to grow” or “to increase” (www.collinsdictionary.com), yet exertions that involve a change of muscle length can either occur via the muscle shortening (concentric exertion), but also lengthening (eccentric exertion) (Knuttgen & Kraemer, 1987; McArdle *et al.*, 1996). Concentric exertions are caused when the moment generated by the muscle exceeds the torque created by an externally applied force, resulting in a shortening of the muscle length (Tortora & Derrickson, 2009). An eccentric muscle action, on the other hand, occurs when the internal moment (i.e., the effort of the muscle) resists the external moment, but cannot equal or exceed it (Klavora, 2010), therefore resulting in the muscle lengthening even though a force is being generated. Despite the terms "concentric" and "eccentric" being flawed, and several other terms having been proposed to describe these two exertion types (refer to Davies & White, 1981; Faulkner, 2003), none of these alternative terms have gained wide-spread acceptance. Therefore, for the purpose of this dissertation the terms "concentric" and "eccentric" are maintained for exertions during which the muscle shortens and lengthens, respectively.

2.3.2 Terminology based on Muscle Force

During dynamic muscle exertions, in other words, those that produce movement, an external resistance is displaced by creating tension within the muscle. An isotonic exertion generates the same internal muscle tension through a range of motion ("iso"= "same"; "tonos" = "force"), regardless of whether the muscle lengthens or shortens (Knuttgen & Kraemer, 1987; Tortora & Derrickson, 2009; Klavora, 2010). It must, however, be emphasized that the external resistance produced by a muscle does not necessarily equal the internal muscle tension (Kroemer, 1970). In other words, even though the magnitude of a load moved remains constant through a range of motion, the internal tension and effort developed by the muscle can vary, as the tension needing to be produced is affected by joint angle (leverage) and resultant angle of pull of the muscle (McArdle *et al.*, 1996; Smith, 2001; Hall, 2007; Klavora, 2010). In fact, exertions that produce a consistent internal tension during a dynamic exertion do not, or at least very rarely, occur in typical human performance (Kroemer, 1970; Knuttgen & Kraemer, 1987; Smith 2001) due to mechanisms such as the length-tension and force-velocity relationships. Isoinertial exertions ("iso" = "same"; "inertia" = "resistance"), on the other hand, are those muscle actions that move a constant external load, while the internal muscle tension varies through the range of motion (Smith, 2001). They are hence now also more commonly known as "dynamic constant external resistance" (Yamaguchi *et al.*, 2006; Costa *et al.*, 2016), during which the muscle continuously adapts its motor recruitment to produce the appropriate tension (Klavora, 2010). Klavora (2010) thus proposed the term “auxotonic” to describe muscle exertions during which internal tension varies, explaining that even though the term “auxotonic” directly translated refers to “increased tension”, it can also be used in the sense of “variable tension”.

2.3.3 Terminology based on Movement Speed

Motions resulting from either concentric or eccentric muscle actions can also vary in terms of their movement speed, which in turn influences the internal tension and effort developed by the muscle (McArdle *et al.*, 1996; Smith, 2001; Klavora, 2010). The term used to refer to a movement at a constant velocity is termed “isokinetic” (Knuttgen & Kraemer, 1987). An individual tested on an isokinetic device would be instructed to exert a force as fast and hard as possible against the device’s attachment, while the isokinetic dynamometer controls the speed of movement, while measuring the force exerted by the muscle (Smith, 2001). This means that during isokinetic exertions, the force produced by the muscle may either be isotonic or auxotonic. Similar to the terms relating to muscle length and muscle force, a muscle action that varies in movement speed throughout the range of motion should thus be called “auxokinetic”.

2.3.4 Terminology combining Muscle Length and Force

Despite differences in opinions on the terminology of muscle actions, it is acknowledged that while all muscle exertions produce force, they can do so either by maintaining a constant length (thus creating no movement) or by changing muscle length (Faulkner, 2003), as well as displaying variations in movement speed. Although the terminology for each classification of muscle actions has been discussed in isolation above, these terms can occur in conjunction with one another. Therefore, an accurate description of a muscle exertion should consist of a combination of the previously discussed terms. However, since running through all possible scenarios would make for a lengthy discussion, and because much of the terminology is not conventional, for the purpose of this dissertation, combination terms used are restricted to those involving muscle length and force.

A completely static exertion, i.e., one without changes in muscle length or changes in force, can be labelled an “isometric-isotonic” exertion (Potvin & Bent, 1997). However, given the discussion around the differences between isotonic and isoinertial exertions, the correct terminology for a purely static muscle action without changes in the external load applied could, therefore, also be "isometric constant-force" (Farina *et al.*, 2002; Contessa *et al.*, 2009), "holding isometric muscle action" (Schaefer & Bittmann, 2017) or simply "isometric-isoinertial". If a constant external force is, however, produced in a moving limb, this could then be termed an "auxometric-isoinertial" exertion. Conversely, if the muscle force varies despite lack of length change, such as when pushing with varying levels of effort against an immovable object, the varying force levels may be described as "auxotonic" (Klavora, 2010), non-isotonic (Iridiastadi, 2003), or even "anisotonic", and could thus be referred to as an "isometric-auxoinertial" exertion. Finally, an exertion that consists of both changes in muscle length and generates variations in force can thus best be described as an "auxometric-auxoinertial" exertion.

2.4 SOURCES OF MUSCLE FATIGUE

With the fundamentals of muscle functioning explained, as well as how different muscle exertions are defined and categorised according to the task characteristics they speak to, the attention in this narrative literature review now turns to the physiological processes leading to muscle fatigue. It is generally accepted that muscle fatigue can occur due to impairments at a multitude of locations along the pathways between the central nervous system and the motor units, thereby contributing to the reduction in force output (Enoka & Duchateau, 2008; Rashedi & Nussbaum, 2015). As a general classification, these causal factors are commonly divided into 'centrally' and 'peripherally' mediated factors (ACSM, 2006). While this is a convenient classification, it is well accepted that there is no single cause of muscle fatigue; rather, several central and peripheral mechanisms are involved concurrently in the development of localised muscle fatigue and may even change as fatigue progresses (Fitts, 1994; Vøllestad, 1997; Potvin & Bent, 1997; Noakes, 2000; Enoka & Duchateau, 2008; Fitts, 2008; Abd-Elfattah *et al.*, 2015).

2.4.1 Central factors

Central fatigue involves all mechanisms located at the higher motor centres (i.e., the central nervous system) that activate the lower motor neurons, including conscious efforts, motivational aspects and integration of sensory information (Enoka, 1995;

Davis & Bailey, 1997). It has been hypothesized that central fatigue is a protective mechanism since the CNS controls muscle force by modifying the activity of motor neurons. Altering the muscle activation strategy can purportedly prevent the muscle from being overloaded and ultimately avoid complete contractile failure (Jensen *et al.*, 2000; Tortora & Derrickson, 2009).

Localised muscle fatigue of central origin is attributed to a reduced motor drive; in other words, a reduction in the number of activated motor units (MUs) and their discharge rates, which in turn fail to maintain muscle activation (Bigland-Ritchie & Woods, 1984; Abd-Elfattah *et al.*, 2015). McNeil *et al.* (2009) point to increases in intracortical inhibition as fatigue progresses, particularly at higher exertion levels, which, as a review by Sahlin *et al.* (1998) suggests, is the result of various metabolic factors, such as hypoglycaemia, hyperammonaemia and altered plasma amino acid composition. It is also believed that brain concentration of serotonin, and perhaps other neurotransmitters, including dopamine and acetylcholine, alter the frequency of the neural impulses reaching the working muscles, i.e. the temporal recruitment, thus affecting the rate of fatigue development (Davis & Bailey, 1997). Alternatively, there may be inhibitory reflexes arising from the exercising muscles, which feed back to the spinal cord, reducing skeletal muscle recruitment at the level of the α-motoneurons, i.e., the efferent nerves innervating the muscle fibres responsible for generating force (Davis & Bailey, 1997; Tortora & Derrickson, 2009).

2.4.2 Peripheral factors

The mechanisms that cause muscle fatigue at the peripheral level involve a complex series of several interrelated metabolic and mechanical factors (Sjøgaard *et al.*, 1988). Peripheral factors of localised muscle fatigue involve mechanisms that interfere with the transmission of the action potential, particularly at the neuromuscular junction, the neurotransmitter availability and secretion in the synaptic gap, excitability of the sarcolemma, sensitivity of the receptors in the post-synaptic membrane, the excitation-contraction coupling process, the contractile elements within muscles, as well as the supply of metabolic energy and accumulation of metabolites (Fitts, 1994; Vøllestad, 1997; Gandevia, 2001; Boyas & Guével, 2011; Abd-Elfattah *et al.*, 2015). Some prominent and well-cited metabolic factors include the inadequate release of calcium ions (Ca^{2+}) from the sarcoplasmic reticulum, depletion of creatine phosphate, as well as exhaustion of glycogen supplies and other nutrients (Sahlin *et al.*, 1998; Westerblad *et al.*, 1998). The resultant decreased rate of ATP hydrolysis, increased concentrations of phosphate ions (due to the break-down of phosphocreatine), increased acidosis due to the build-up of lactic acid and H^+ ions, and failure of action potentials in the motor neuron to release enough acetylcholine (ACh) (Fitts, 1994; Sahlin *et al.*, 1998; Boyas & Guével, 2011; Wan *et al.*, 2017) are thought to ultimately interfere with the cross-bridges' ability to generate force, either by inhibiting the formation of cross-bridges or by preventing the detachment of the myosin head from actin after the powerstroke (Sjøgaard & Jensen, 2006), although there are still some disputes in the literature about their exact effects on force production (Westerblad *et al.*, 1998).

Although muscle fatigue can be attributed to a combination of central and peripheral factors, localised muscle fatigue, particularly during prolonged submaximal intermittent exertions, is thought to be more influenced by peripheral mechanisms (Bigland-Ritchie *et al.* 1986b; Enoka, 1995; Rashedi & Nussbaum, 2015). Furthermore, these processes that occur distal to the neuromuscular junction and that interfere with the muscle's contractile mechanism (Bigland-Ritchie *et al.*, 1978; Gandevia, 2001) may also be the reason for a longer recovery duration than centrally induced fatigue (Rashedi & Nussbaum, 2015).

2.4.3 Adaptations to Muscle Fatigue

Fatigue does not only manifest in an "externally measurable impairment", but also in non-observable 'internal' adaptations (Behm, 2004). Throughout the physical activity, the neuromuscular system responds to fatigue-induced limitations by adopting various neuromuscular strategies to maintain the required force output and, therefore, "resist" the effects of fatigue (Behm, 2004). According to Boyas and Guével (2011), various electrical and mechanical mechanisms initiated at the start of an activity enable the muscle to maintain, or even increase, the force levels required, even though other mechanisms of fatigue have started setting in. As fatigue sets in during submaximal force production, an increasing number of motor units are recruited at a central level to maintain a certain force output (Enoka & Fuglevand, 2001; Contessa *et al.*, 2009). This, however, only occurs up to a point, after which further increases in force can only be achieved with increased firing rates (De Luca *et al.*, 1982; Moritani *et al.*, 1981; Boyas and Guével, 2011). Clamann (1993) reported that greater rate coding resulted in smoother force gradation compared with increased recruitment. The adjustment in the discharge rate does, however, change depending on the type of muscle exertion and intensity of the activity (Boyas and Guével, 2011). Furthermore, varying patterns of motor unit activity mean that some motor units are generating tension, while others are inactive until the active motor units drop out and are replaced by the previously inactive units (Sjøgaard & Jensen, 2006; Tortora & Derrickson, 2009) - a phenomenon called "motor unit rotation" (Bawa *et al.*, 2006). As fatigue progresses, more motor units are recruited, which is also evident from the increases in EMG (Potvin & Bent, 1997). According to the same authors, it appears, however, that at some point between 50% and 85% of maximum voluntary effort (MVE), i.e., the amount of force that can be voluntarily generated, the entire motor pool has been recruited and, consequently, this explanation may only be valid for relatively low force levels (Potvin & Bent, 1997).

Furthermore, it has been proposed that with fatigue, the motor units start firing more synchronously. Synchronization of motor units refers to the simultaneous discharge of motor neurons, meaning that motor neurons reach their action potential threshold at the same time (Yao *et al.*, 2000). Although the functional significance of motor unit synchronization is still not completely understood, Semmler (2002) speculated that it is a deliberate movement control strategy to increase some aspect of motor performance, even though it does not necessarily increase overall force output. Nonetheless, there does appear to be consensus that it contributes to increases in EMG amplitude (Naeije & Zorn, 1982; Yao *et al.*, 2000; Semmler, 2002).

In addition to the above theories, other mechanisms have been proposed that enable the maintenance of the force as fatigue progresses, or that purposefully downregulate the neural drive. "Neural potentiation" is an enhancement of a muscle's force because of previous exertions (contractile history), although this depends on the muscle in question (Robbins, 2005). Similarly, force production can be temporarily enhanced via the insertion of additional neural impulses into the existing motor program – a phenomenon referred to as 'catch-like properties' (Behm, 2004). Conversely, 'muscle wisdom' refers to variations in the motor unit discharge rate and results in an adjustment in the muscle's contractile speed. This is considered a defence mechanism, protecting the muscle from harm induced by the activity by limiting the reduction in the membrane excitation and calcium release. Muscle wisdom is, however, dependent on the task characteristics and the muscle itself (Boyas & Guével, 2011).

2.4.4 Recovery from Muscle Fatigue

While it is important to understand the physiological processes leading to muscular fatigue during static and dynamic exertions, it is arguably equally important to understand how the muscle recovers from fatigue. Schwendner *et al.* (1995) define recovery from fatigue as "the return of the neuromuscular force generating potential following fatiguing exercise" (p.186), or within 80% thereof. Restoration of a muscle's maximal force production does, however, have very different time courses, depending on the manner in which fatigue was induced (Allen *et al.*, 2008). While there is limited understanding regarding the details of the recovery process(es) (Rashedi & Nussbaum, 2015; Carroll *et al.*, 2017), it is accepted that the processes that cause localised muscle fatigue to develop are reversible with time, and that muscles can regain their full strength and endurance capacity so long as the rate of fatigue build-up does not exceed the rate of recovery, in which case the end-result would be a state of exhaustion (Liu *et al.*, 2002; Sjøgaard & Jensen, 2006).

Recovery from muscle fatigue is a continuous process that not only occurs after cessation of the fatiguing activity but may also occur concurrently with that of fatigue accumulation (Frey Law *et al.*, 2012), even though most studies that have investigated recovery have done so in a non-activated muscle after cessation of a fatiguing protocol (e.g., Yates *et al.*, 1987; Linnamo *et al.*, 2000; Cheng & Rice, 2005; Oranchuk *et al.*, 2020). Such studies have shown that recovery of a fatigued muscle generally follows a logarithmic curve, meaning the force generated during a maximum voluntary exertion improves rapidly over the first few minutes following the end of the fatiguing exercise, after which the recovery rate slows down and can require hours, and sometimes days, to achieve full recovery (Lind, 1959; Rohmert, 1960; Linnamo *et al.*, 2000; ACSM, 2006; Carroll *et al.*, 2017). A statistical model by Frey Law *et al.* (2012) suggested that recovery after sustained isometric exercise takes 10-15 times longer than the fatigue process itself. Lind (1959) found that for the initial phase of rest (i.e., the first 10 minutes after cessation of the fatigue protocol), the muscle recovered by more than 50%, followed by a further 20-25% recovery during the following 30 minutes. The effectiveness and rate of recovery from muscle fatigue are, however, dependent on the causal mechanisms of fatigue, which, in turn are influenced by the task characteristics and which are detailed in Section 2.6.

What can be concluded from the information summarised in this section is that muscle fatigue is multifaceted with a variety of central and peripheral causal factors, as well as mechanisms that attempt to maintain the required motor performance despite ever-increasing decrements. Furthermore, muscle fatigue is reversible, yet the time course of recovery is variable. Before interrogating the factors influencing the variability in muscle fatigue development and recovery, it is first necessary to detail how muscle fatigue is quantified.

2.5 INDICATORS OF MUSCLE FATIGUE

Muscle fatigue is the outcome of complex interactions of central and peripheral factors, each of which is the result of a different central or peripheral physiological mechanism (Allen *et al.*, 2008; Enoka & Duchateau, 2008). Since fatigue is a multidimensional concept (Merletti *et al.*, 1991), it can be quantified via a variety of biomechanical, physiological and biochemical measures (De Luca, 1984; Gonzáles-Izal *et al.*, 2012). Kumar *et al.* (2002) agree that even though there is no direct, valid, and reliable quantitative measure of localised muscle fatigue, there are a variety of subjective and objective indicators of fatigue. Limiting an assessment method for fatigue to just a single variable may, hence, run the risk of missing changes in one or other variable that come with muscle fatigue (Seghers & Spaepen, 2004). Seghers and Spaepen (2004) and Madeleine (2010) thus suggest that the combined use of several measurement methods might yield a more complete picture of muscle fatigue. A review by Al-Mulla *et al.* (2011) highlighted several sensors and signal acquisition methods that could be used for the study of muscle fatigue. However, careful selection of such assessment methods is required when more complex muscle exertions are involved since recruitment patterns can change, depending on the properties of the task (Gardiner, 2011). Furthermore, De Luca (1984) and Seghers and Spaepen (2004) pointed out that monitoring temporal changes in the physiological variables throughout an activity would be necessary since muscle fatigue is an ongoing process over time rather than failure at a point in time. Some of the most commonly used fatigue indicators relevant to this thesis include: reduced strength and endurance, a greater perceived exertion, or sense of effort invested, increased discomfort, diminished neuromuscular control, muscle tremors, longer reaction times, decrements in proprioceptive ability, and altered electromyographic (EMG) signals (Madeleine, 2010; Rashedi & Nussbaum, 2015).

2.5.1 Maximum Voluntary Force Exertions

Probably the most commonly used objective indicator for the development of localised muscle fatigue is the failure to maintain a maximal or sub-maximal force output (Edwards, 1981; Hagberg, 1981; Bigland-Ritchie & Woods, 1984; Iridiastadi & Nussbaum, 2006; Søgaard *et al.*, 2006). The use of maximum voluntary exertions (MVEs) is a popular method to quantify the relative decrease in force-generating capacity of the total neuromuscular system (Bigland-Ritchie & Woods, 1984; Gandevia, 2001; Kapitaniak, 2001b; Carroll *et al.*, 2017) given the simplicity of recording this variable (Krüger *et al.*, 2019). However, there is no consensus yet on the magnitude of strength decline that would confirm fatigue. Rainoldi *et al.* (2001) suggested that a 10% day-by-day variation was considered a ‘normal’ level of maximum isometric muscle strength variability. Similarly, studies cited by Wyse *et al.* (1994) mentioned maximum isokinetic strength variability of 8.5% to 13.1% for intra-day and inter-day trials, respectively, while the fatigue protocol in a study by Dickerson *et al.* (2015) was terminated once static strength had reduced to less than 70% of an individual’s unfatigued strength capacity.

Related to maximum strength is the concept of peak power, which is calculated as the rate of force development, or the amount of work performed within a given timeframe. Work, on the other hand, is the product of the force generated over a certain distance, meaning that work and power can only be computed for exertions that exhibit a change in position (Hall, 2007). Peak power declines due to decrements in force produced (and consequently work), as well as velocity (ACSM, 2006). The force-power relationship states that with increasing levels of fatigue, peak power declines and is maintained at progressively lower shortening velocities, resulting in slower movements (ACSM, 2006). A study by Krüger *et al.* (2019) confirmed that during a cycling protocol, peak torque, peak power, and maximal velocity all decreased with fatigue development, with greater decrements witnessed with higher levels of exercise intensity.

One more variable associated with maximum strength that has gained recent interest as a fatigue indicator is that of the rate of force development, which has been suggested to be a more sensitive indicator of neuromuscular fatigue than maximum voluntary force (D'Emanuele *et al.*, 2021). It reflects how quickly a muscle can increase the force it generates during an explosive and voluntary exertion (Maffiuletti *et al.*, 2016). Therefore, it is usually indicated by the time it takes the muscle to reach peak torque, with a higher rate of force development being reflected by a shorter time to peak torque. Neuromuscular fatigue impairs the rate of force development, resulting in an increase in the time required to reach peak torque (D'Emanuele *et al.*, 2021).

2.5.2 Endurance Time

Instead of measuring the decrement in maximum force, fatigue can also be quantified as the duration to the point of task failure (Gandevia, 2001). The time at which an individual expresses the inability to continue with a certain task due to lack of physical capacity (Dickerson *et al.*, 2015) is also referred to as the endurance time. The limits of physical capacity could include maintaining a specific force output (Kumar, 2006; Enoka & Duchateau, 2008), or keeping up with specific movement requirements, such as speed, timing, or movement accuracy (Cheng & Rice, 2005; Gates & Dingwell, 2011; Cowley *et al.*, 2014). While there are numerous peripheral factors that influence maximum endurance time, Mauger (2013) argues that the consideration of fatigue as being the point of task failure is limited in physical activities that are self-paced, and that the ability to exert a specific force level or movement rhythm is subjectively modifiable. Motivation, mood, perceived effort, and discomfort are some such factors influencing muscular endurance capacity (Kumar, 2006; Marcora, 2010).

2.5.3 Perceptual Responses

While localised muscle fatigue can be assessed in terms of performance decrements, or the inability to maintain a particular performance, it can also be determined by the perceived effort that arises from maintaining a given work output (Bigland-Ritchie & Woods, 1984; Gandevia, 1999; Marcora *et al.*, 2009; Boyas & Guével, 2011). Traditional explanations for exercise termination relate to physical/physiological processes and have neuromuscular, bioenergetic and resource considerations (Behrens *et al.*, 2023; Noakes, 2000). The evidence pointing to a relationship between perceived muscle fatigue and objective fatigue indicators is overwhelming. For example, a study by Marzouk *et al.* (2023) showed clear associations between a variety of neuromuscular and biomechanical measures and self-reported fatigue. Similarly, Zhao *et al*. (2022) and Zhao *et al*. (2023) reported moderate to strong correlations between ratings of perceived exertion and the spectral fatigue index during knee flexion, elbow flexion and a back squat exercise. However, others, such as proponents of the psychobiological model of exercise tolerance, have argued that fatigue is highly dependent on one's perceived ability as well as motivation to succeed and complete the task (Hartman *et al.*, 2019; Marcora & Staiano, 2010; Pageaux, 2016). For example, Marcora (2009) proposed that the sense of effort is centrally generated and positively correlates with markers of metabolic muscle stress and increased central command, although the exact cause-and-effect relationship is not clear. Furthermore, during low-to-moderate exercise intensities, it seems that it is not only psychological factors that influence perceptions of effort but also social influences (Marcora, 2010). These factors can thus affect the validity of such subjective feelings as indicators for fatigue. Therefore, it is advisable to use perceived effort as an adjunct to other fatigue measures (Marcora, 2009).

While fatigue development might be indicated by high ratings of perceived discomfort and could thus warrant the use of a pain scale (Iridiastadi & Nussbaum, 2006), arguably the most commonly used rating scales to subjectively quantify fatigue are the Visual Analogue Scale (VAS) (Capodaglio, 2001), Likert Scale (Grant *et al.*, 1999) and Ratings of Perceived Exertion (RPE) Scale (Borg, 1982). Of these, the use of RPE is a valid indicator of localised muscle fatigue, as demonstrated by a variety of studies that found significant correlations of RPE with muscle activity (e.g., Capodaglio, 2001; Jakobson *et al.*, 2014; Williams, 2017; Ahmad & Kim, 2018). A comparative study of rating scales by Grant *et al.* (1999) found results obtained from the VAS during a running protocol to be more reproducible compared to the Likert and Borg scales, yet it was the Borg scale that was more sensitive to changes in general fatigue. Capodaglio (2001) found that during an arm cranking exercise, the Borg Category Ratio (CR10) scale results significantly correlated with those of the VAS. These researchers concluded that the CR10 scale showed a good level of reliability and reproducibility, and it is thus suitable in dynamic application settings such as sports.

2.5.4 Electromyography

Electromyography (EMG) is the method of measuring and analysing myoelectrical signals (Konrad, 2005) and has long been considered a reliable tool for indicating the development of localised muscle fatigue (Kumar, 2006; Al-Mulla *et al.*, 2011). Changes in physiological responses, such as changes in motor unit firing rates, can be picked up by EMG (Troiano *et al.*, 2008), and this myoelectrical activity is in turn associated with a muscle's mechanical activity (Kapitaniak, 2001a). Surface EMG (sEMG) makes use of surface electrodes overlying a muscle to record the action potentials at the muscle fibre membrane underlying the electrodes (Konrad, 2005; Zwarts *et al.*, 2008; Tortora & Derrickson, 2009; Fukuda *et al.*, 2010), as opposed to intramuscular electrodes, which are needle electrodes that are inserted into the muscle and are thus more invasive and therefore less commonly used. The raw electrical signals picked up by either EMG electrodes are amplified and filtered to eliminate noise (Konrad, 2005), after which they can be processed.

With the development of localised muscle fatigue, certain changes in neuromuscular coordination occur; hence, the use of EMG is widespread for inferring and quantifying fatigue (Zwarts *et al.*, 2008; Al-Mulla *et al.*, 2011; Marco *et al.*, 2017). However, determining localised muscle fatigue from EMG is less explicit than maximum force or perceived discomfort ratings. For one, the interpretation of changes in muscle activity is influenced by the quality of the EMG signal (De Luca *et al.*, 2010), as the signal is dependent on various factors, including tissue characteristics, crosstalk from neighbouring muscles, electrode type and placement, movement between the electrode site and muscle belly, external noise from electrical hum, or the electrodes and amplifiers themselves (Merletti, 1999; Farina *et al.*, 2004; Konrad, 2005; Raez *et al.*, 2006; González-Izal *et al.*, 2012). Furthermore, the surface EMG signals tend to show random waveforms, due to the many signals received simultaneously from different motor units and the superposition of these action potentials (De Luca, 1997). These signals, therefore, need to be processed first before inferring fatigue (Marco *et al.*, 2017).

The challenges experienced with EMG signal recording and interpretation occur during all exertion types; however, they are amplified during dynamic exertions due to their non-stationarity (Knaflitz & Bonato, 1999; Farina *et al.*, 2002; Cifrek *et al.*, 2009). This complexity added during dynamic muscle actions is the result of the alterations of recruitment and de-recruitment of motor units, the shifting of the electrodes relative to the origin of the action potential, and the changes in conductivity properties of the tissues as the muscle changes its fibre diameter, length, and orientation as the joint angle changes (Farina *et al.*, 2004; Farina, 2006; Merletti *et al.*, 2004). The EMG signal also changes with the recruitment of different numbers of motor units, adjustments in force or power through the range of motion, variations in fibre and muscle length, as well as the muscle fibre conduction velocity due to muscle fatigue (Merletti *et al.*, 2004; Farina, 2006; Cifrek *et al.*, 2009). Despite this, several authors have successfully made use of EMG to assess muscle fatigue during dynamic exertions (e.g., Komi & Tesch, 1979; Farina *et al.*, 2004; Yassierli & Nussbaum, 2007; Gonzáles-Izal *et al.*, 2010).

EMG feature analyses can be divided into three categories: the time domain analysis, the frequency domain analysis, and the time-frequency domain analysis (Rainoldi *et al.*, 2001; Farina *et al.*, 2002; Phinyomark *et al.*, 2012; Yousif *et al.*, 2019). While Phinyomark *et al.* (2012) proposed 37 features of EMG signal classifications, for this review, only the most prevalent methods relevant to fatigue identification and relating to the time domain and frequency domain are considered.

Time Domain Analysis

The EMG's temporal, or time, domain makes use of the EMG signal's amplitude, which indicates the neural drive to the muscle and the number of motor units activated, in other words, muscle excitation (Kamen & Gabriel, 2010; Vigotsky *et al.*, 2018). It does this by either calculating the root mean square (RMS) of the signal, the integrated EMG (iEMG), or the average rectified value (ARV) of an interval, amongst others (Merletti *et al.*, 1991; Rainoldi *et al.*, 2001; Zwarts *et al.*, 2008; Nazmi *et al.*, 2016). Calculating the root mean square (RMS) can quantify the electric signal, as it reflects the physiological activity in the motor unit during muscle exertions (Fukuda *et al.*, 2010). Fatigue is detected during isometric exertions if the EMG amplitude (i.e., either the RMS, iEMG or ARV) increases significantly over time while maintaining a constant force output (Hagberg, 1981; Bigland-Ritchie & Woods, 1984; Rainoldi *et al.*, 1999; Seghers & Spaepen, 2004; Kumar, 2006; Kamen & Gabriel, 2010).

During prolonged submaximal exertion, the active muscle fibres lose their capacity to produce tension due to metabolic changes, which necessitates the recruitment of additional motor units to maintain the tension required by the task or activity (Bigland-Ritchie & Woods, 1984). The recruitment of more motor units, as well as their increased excitation rate and synchronous firing, are responsible for the signal amplitude increases (Enoka & Duchateau, 2008; Al-Mulla *et al.*, 2011). Since the central nervous system facilitates the recruitment of these additional units, the increase in electrical activity is picked up by the EMG system (Chaffin, 1973). However, during maximal isometric exertions, the pool of inactive motor units to draw on is limited, which results in the fatigued motor units dropping out without being replaced. The motor unit firing rate and muscle fibre conduction velocity decrease, thus resulting in a decline in the recorded EMG amplitude (Bigland-Ritchie *et al.*, 1983; Kamen & Gabriel, 2010). One limitation of the time-domain analysis is that it is not possible to differentiate motor unit recruitment from rate coding using sEMG amplitude, as the recruitment characteristics vary from muscle to muscle and also between force exertions (Vigotsky *et al.*, 2018).

While fatigue-related changes in EMG amplitude are well-established for isometric exertions, they are less clear for analysis of fatigue during dynamic exertions since the difficulties in interpreting sEMG signals in static contractions are amplified in dynamic cases. Apart from the factors mentioned earlier in this review that influence the sEMG signals recorded during static exertions, there are other factors that affect sEMG signals during dynamic tasks and that differ from those recorded during static conditions (Farina, 2006). For one, EMG analyses of dynamic muscle actions undermine the assumptions of stationarity (González-Izal *et al.*, 2012), therefore complicating the analysis of the EMG signal obtained during a movement. During static muscle actions, the joint angle remains constant; however, during dynamic exertions, the joint angle changes, which causes a shift of the underlying muscle fibres with respect to the recording electrodes (Roy *et al.*, 1998). In addition, during a dynamic exertion, the rapid changes in the recruitment and de-recruitment of motor units and changes in muscle force cause a faster change in the sEMG signal properties than in an isometric exertion (González-Izal *et al.*, 2012). Therefore, since the sEMG signal during a dynamic contraction can be assumed to be non-stationary, the traditional frequency techniques may not be appropriate for extracting information, and more complex techniques are needed (González-Izal *et al.* (2012).

Despite the challenges faced by the nonstationarity of dynamic exertions, Tesch *et al.* (1990) found that fatigue-induced amplitude-based changes during dynamic exertions follow the same trend as those obtained during isometric muscle actions. In other words, the amplitude increases under submaximal exertions of a constant force and decreases under maximal exertions. Furthermore, findings by Yassierli and Nussbaum (2009) suggest that during dynamic trunk exertions, RMS may be a better fatigue measure to use than the mean power frequency; however, the same was not found for isokinetic shoulder abductions (Yassierli & Nussbaum, 2007).

A common procedure for analysing EMG amplitude is to normalize the working EMG signal to the signal obtained during a maximum voluntary isometric exertion (MVIE). Such normalization procedures were found to be suitable during static and dynamic exertions (Albertus-Kajee *et al.*, 2011). However, the same authors point out that the use of the MVIEs as a standardization method does come with limitations, namely, a) maximum exertions assume that the maximum voluntary effort does represent 100% of muscle activity, and b) dynamic exertions are limited by the force-velocity relationship as well as factors such as the shifting of motor units under the electrode site during movement (Soderberg & Knutson, 2000). On the other hand, Mirka (1991) considered dynamic methods of normalization to be more appropriate than using isometric exertions, as dynamic normalization methods consider the effects of muscle activation, muscle length, and angular velocity on EMG. Using maximum voluntary isometric exertions for normalization of dynamic exertions would require interrupting the dynamic activity to perform the MVIE (Potvin & Bent, 1997; Roy *et al.*, 1998). In addition, motor recruitment strategies during isometric exertions may be different compared to dynamic exertions (Nakawaza *et al.*, 1993; Nazmi *et al.*, 2016).

Frequency Domain Analysis

Another approach to analysing the raw EMG signal is via frequency domain analysis (Seghers & Spaepen, 2004; Zwarts *et al.*, 2008). This analysis method makes use of the EMG signal's spectral frequencies, also known as the 'power spectrum'. Generally, the raw signal undergoes a Fast Fourier transformation (González-Izal *et al.*, 2012), after which the time course of the power spectrum's mean or median frequency is calculated (Rainoldi *et al.*, 2001), although numerous other transformation techniques exist (Knaflitz & Bonato, 1999). Frequency domain analysis can provide physiological information such as muscle fibre conduction velocity, and possibly even motor unit firing rates (Kamen & Gabriel, 2010).

Amongst all spectral parameters of the EMG signal (refer to Phinyomark *et al.*, 2012 and Nazmi *et al.*, 2016), the mean power frequency (MnPF) and the median power frequency (MdPF) are among the most common parameters used in fatigue studies since they are sensitive to changes to the physiological manifestations of fatigue (Iridiastadi & Nussbaum, 2006; Zwarts *et al.*, 2008) and are less affected by noise (De Luca, 1997). Zwarts *et al.* (1987) found that MdPF is directly proportional to the speed of muscle fibre conduction velocity, with correlation coefficients ranging between 0.79 and 0.98. It must however be pointed out that Naeije and Zorn (1982) found selective increases in muscle fibre action potential conduction velocity and spectral shifts, indicating that spectral shifts are not solely dependent on MFCV but also other mechanisms. Nonetheless, together with the mean power frequency (MnPF), the dynamics of muscular recruitment can be observed. Furthermore, Iridiastadi and Nussbaum (2006) found that changes in spectral frequency were moderately correlated with muscle strength and perceived discomfort, thus confirming a certain level of usefulness of these measures as an indirect measure of fatigue. However, De Luca (1997) also pointed out that median frequency signals are more variable due to the less consistent EMG signal at lower frequencies, but that this could be rectified via appropriate filtering.

As muscle fatigue progresses, changes in temporal recruitment occur, hence the rate of decline of the mean and median frequency during sustained exertions has been commonly used as an indicator of localised muscle fatigue (Zwarts *et al.*, 1987; Kumar & Narayan, 1998). A study by Öberg *et al.* (1990), however, concluded that to confidently infer fatigue development, the change in the mean frequency had to be 8% or more relative to the starting value. The shift to the lower frequencies of the frequency spectrum observed during fatigue is largely attributable to diminished conduction along the sarcolemma, although central changes in motor-unit firings, particularly synchronisation, also play a role (De Luca, 1984; Zwarts *et al.*, 1987; Farina *et al.*, 2002; Liu *et al.*, 2021). Muscle fibre conduction velocity decreases (Zwarts *et al.*, 1987; Masuda *et al.*, 1999; Farina *et al.*, 2002), which is thought to be the result of a change in the energy metabolism (such as depletion of substrates) (Hagberg, 1981), as well as an accumulation of metabolic by-products such as lactic acid, reducing intracellular pH and decreasing the excitability of the muscle fibre membrane (Masuda *et al.*, 1999). The number of active motor units also decreases (Bigland-Ritchie and Woods, 1984), motor units fire more slowly and become more synchronized (Liu et al., 2021). Surface EMG shows these effects via the power spectrum shifting linearly to lower frequencies during both maximal and submaximal isometric exertions (Viitasalo & Komi, 1977; Bigland-Ritchie and Woods, 1984; Zwarts *et al.*, 1987; Masuda *et al.*, 1999; Rainoldi *et al.*, 1999; Nussbaum, 2001; Kamen & Gabriel, 2010). Possible explanations for this lie with central and peripheral factors, of which peripheral factors are thought to be caused by changes in the shape of motor unit potentials and a decrease in muscle fibre conduction velocity (MFCV) (Zwarts *et al.*, 1987). This shift to the left in the power spectrum also correlates with the reduced muscle pH due to biochemical changes (Gamet & Fokapu, 2008) and reductions in force output (Mannion & Dolan, 1996).

While decreases in EMG mean and median power frequencies have been widely used as fatigue indicators, they have been generally used during sustained isometric contractions at submaximal effort levels (De Luca, 1997). However, as with the EMG amplitude measures, there is concern regarding the validity of such measures for dynamic exertions (Roy *et al.*, 1998; Nussbaum, 2001; Kamen & Gabriel, 2010). It is quite likely that the patterns of neural activation are different during dynamic and static muscle actions (González-Izal *et al.*, 2010); hence, extracting information from the EMG signal obtained during a static exertion to infer fatigue during dynamic exertions may not be accurate. Furthermore, the changes in force/power and the movement of a limb through a range of motion result in a shift of the muscle underneath the electrodes and may undermine the assumption of non-stationarity of the EMG signal needed for the Fast-Fourier Transforms when calculating the EMG power spectrum (Farina *et al.*, 2002). Mean and median frequencies change with varying joint angles (Öberg *et al.*, 1990), as well as alterations in the number of active motor units, especially for exertions of less than 20-30% of maximum voluntary exertion (Kamen & Gabriel, 2010), and which may be the reason why there may not necessarily be changes in the spectral frequency during dynamic exertions (e.g., Ament *et al.*, 1996). In such cases, it is recommended to analyse time intervals of between 0.5-1s for analysis and to report the frequency results in those intervals (Kamen & Gabriel, 2010) or to obtain EMG amplitudes and frequency measures at the same muscle length during dynamic exertions.

In contrast, other studies have found that, despite its limitations, standard spectral analyses are still appropriate and reliable indicators of fatigue for EMG taken from dynamic cyclic efforts (Christensen *et al.*, 1995; Potvin & Bent, 1997; MacIsaac *et al.*, 2001; Nussbaum, 2001). They have even been argued to be less variable than the EMG-RMS measures and to be more sensitive to changes in task-related parameters (Iridiastadi & Nussbaum, 2006; Yassierli & Nussbaum, 2007). Similarly, the study by Potvin and Bent (1997) supports the use of median power frequency recorded during dynamic exertions to quantify fatigue of the biceps brachii muscle. In their study, the average decreases in the MdPF were consistent across measurements taken during a dynamic activity and during isometric exertions. Mathematical modelling approaches by MacIsaac *et al.* (2001) supported the notion that compression of the power spectrum to lower frequencies also applied to dynamic exertions. Therefore, even though some authors caution against the use of EMG spectral analyses to infer fatigue during dynamic exertions due to the effects of changing muscle force and lengths and changes in underlying motor recruitment patterns, the mean and median power frequency remain popular variables in fatigue studies (Nussbaum, 2001; Farina, 2006).

Time-frequency Analysis

Time-frequency domain analysis is a combination of time and frequency (Nazmi et al, 2016). It is considered to be a more suitable analysis for EMG measurements obtained under non-stationary conditions (Knaflitz & Bonato, 1999; Raez *et al.*, 2006), and makes use of features such as discrete and continuous wavelet transforms, empirical mode decomposition, and wavelet packet transform (Knaflitz & Bonato, 1999; Nazmi *et al.*, 2016). Compared to the time- or the frequency domain analyses, relatively few studies have, however, used the time-frequency analysis method.

2.5.5 Performance Accuracy and Precision

Two indicators of successful task performance are its accuracy and its precision (Kumar *et al.*, 2017). Although these terms are often used interchangeably, there is a distinct difference between them. In the human movement sciences, ‘accuracy’ refers to the “degree of closeness” to an intended movement outcome; in other words, how close a motor performance came to achieving a set goal, while ‘precision’ is the ability to consistently reproduce a particular motor performance outcome, i.e., several repetitions of a movement with the same performance objectives (Hamm, 2016). These objectives can relate to: 1) the positioning of a limb, either at a set joint angle during isometric exertions, but more generally at the movement endpoints during dynamic exertions, 2) generating a specific target force, or 3) achieving the correct movement timing.

Neuromuscular motor control is the ability of the central nervous system to execute “purposeful, coordinated movements” (Latash *et al.*, 2010, p.382). This is a complex process that relies on the central nervous system appropriately firing action potentials to relevant muscles, and the suitable adjustment thereof in response to multisensory feedback from mechanoreceptors located around the joint and within the muscles themselves (Honeybourne, 2006; Vafadar *et al.*, 2012), as well as other sensory modalities such as vision, audition, and tactility (King et al., 2009; Ladda *et al.*, 2020). The accuracy of motor control is subject to numerous influencing factors, including age (e.g., Lindberg *et al.*, 2009), movement speed (so-called “Fitts Law”; Schmidt & Wrisberg, 2008), and visual feedback (van Leeuwen, 1999). Additionally, fatigue has also been hypothesized to impact performance accuracy since it negatively influences proprioception (Voight *et al.*, 1996), which in turn can lead to loss of motor control (Kumar, 2001) since it distorts the important feedback information needed to adjust the motor program. However, while some authors found decreases in movement endpoint accuracy with localised fatigue development (Missenard *et al.*, 2009), others found no change in motor accuracy under fatigued conditions (Selen *et al.*, 2007; Cowley *et al.*, 2014). The latter authors concluded that under fatigued conditions, the central nervous system may adopt different control strategies, such as increased coactivation of muscles, to still achieve the intended performance outcome.

Motor precision, on the other hand, is the ability to repeatedly achieve the same motor outcome, although it may not necessarily be accurate (Hamm, 2016). Any temporal-spatial variations observed in movement precision, particularly over time, are thus referred to as ‘movement variability’, ‘motor variability’, ‘coordination variability’ or ‘movement consistency’ (Monster *et al.*, 1978; Stergiou *et al.*, 2006; Srinivasan & Mathiassen, 2012; Lockhart & Stergiou, 2013). While these terms are also often used interchangeably, it is important to point out the subtle differences between them. For example, Hamill *et al.* (2012) differentiate between “coordinative variability”, i.e., the interactions between different body segments, and “endpoint variability”, the outcome of the movement or goal-directed task. Similarly, Preatoni *et al.* (2013) mention the term “performance variability”, namely, the physical adjustments made between repetitions of the same movement by means of adjusting certain kinematic variables, and “outcome variability”, which refers to the variations in the final product of that movement. In addition to performance and outcome variability, a further category was proposed by Cowin *et al.* (2022), whose framework included “strategic variability”, and which refers to the differences in the approach selected at a cognitive level for the execution of a movement.

Since execution and outcome variability occur at the motor execution level, they can be quantified by means of kinematic variables (Cowin *et al.*, 2022; Preatoni *et al.*, 2013). Movement variability can be quantified using temporal and spatial performance measures, such as cycle-to-cycle deviations in movement accuracy and precision (e.g., joint angles at the endpoints of a range of motion), movement time (e.g., spike timing and movement duration), movement velocities and accelerations (Enoka *et al.*, 2003; Faisal *et al.*, 2008; Srinivasan & Mathiassen, 2012). Motor variability can also be inferred from muscle activity and recruitment patterns, and force exertions (Faisal *et al.*, 2008; Srinivasan & Mathiassen, 2012; Vafadar *et al.*, 2012; Cowley *et al.*, 2014).

Various factors can influence movement variability, including individual and task-related factors (Srinivasan & Mathiassen, 2012; Gaudez *et al.*, 2016), with muscle fatigue being one contentious factor impacting movement variability (e.g., Cortes *et al.*, 2014; Gates & Dingwell, 2011). Under unfatigued conditions, natural variations occur in forces when performing an isometric exertion (Enoka *et al.*, 2003). It is also well-accepted that exercise-induced fatigue affects motor control due to changes in neuromuscular coordination, including the timing and duration of muscle activation and the pattern of muscle activity, and this could affect an individual’s ability to perform smooth and controlled movement patterns (Cortes *et al.*, 2014; Gates & Dingwell, 2011).

The effects of muscle fatigue on movement variability are, however, contested. Some studies found increases in isometric force fluctuations, postural tremors and altered dynamics of limb motion (Vafadar *et al.*, 2012; Cowley *et al.*, 2014), which in turn induce modifications in movement performance (D'hooge *et al.*, 2012; Cortes *et al.*, 2014; Abd-Elfattah *et al.*, 2015). For example, an experiment by Contessa *et al.* (2009), which required participants to maintain a steady force at 20% MVIC, showed significant increases in the variability of force recordings as fatigue set in. This increased force variability did not correlate with motor unit firing rates but rather was attributed to the additional recruitment of motor units. On the other hand, an increasing body of evidence indicates that movement variability decreases with fatigue. Allen and Proske (2006) found that even though muscle fatigue increased errors in position-matching as well as the direction of the position error, fatigue did not significantly affect movement-tracking error. Similarly, Cortes *et al.* (2014) concluded that fatigue in the lower extremities had a differential effect on movement variability of kinetic and kinematic markers, as, during a side-stepping protocol, they found significantly reduced variability of time-dependent and amplitude-dependent measures, such as ground reaction forces, when fatigued. A further study revealed significant increases in the superior-inferior accuracy of reaching a target with muscle fatigue, but no differences in the mediolateral direction (Vafadar *et al.*, 2012).

The exact reason(s) for the variations in motor control are not yet fully understood and Gaudez *et al.* (2016) indicated that the relationship between fatigue, performance accuracy and motor variability is complex and conflicting. Since fatigue decreases a muscle's force-generating capacity (Enoka & Duchateau, 2008), but also alters afferent proprioceptive feedback, it is plausible that, as muscle fatigue progresses, there is an increased need for activation to maintain joint stability (Solomonow *et al.*, 2003), as well as overall motor performance. This would necessitate the recruitment of larger motor neurons or change the levels of co-activation, which in turn would produce a more variable force output (Missenard *et al.*, 2009). Faisal *et al.* (2008) also speculated that localised muscle fatigue influences the transitions of action potentials at the neuromuscular level, where the transition of action potential twitches into smooth movement is inhibited, resulting in jerky movements. Furthermore, it is unknown whether fatigue-induced changes in motor control occur due to decrements in proprioception (Abd-Elfattah *et al.*, 2015) or whether muscle fatigue causes a re-organization in motor strategies aimed at maintaining task performance (Madeleine, 2010; Srinivasan & Mathiassen, 2012). The question, therefore, remains whether movement variability is a lack of motor control resulting from muscle fatigue, or whether changes in movement variability occur proactively prior to the onset of fatigue to prevent certain negative fatigue-induced effects (Srinivasan & Mathiassen, 2012).

Numerous variability analysis techniques exist, which can be classified into several domains of analysis (for comprehensive reviews of analysis techniques, refer to Bravi *et al.*, 2011 and Preatoni *et al.*, 2013). Technologically advanced three-dimensional motion analysis techniques, such as 3D array analysis, have not only shown to provide high levels of accuracy, but are also useful in research where the movements of individual are dynamic, cover large physical areas and in environments prone to interferences, such as sports performance on the sports field (e.g., Bonnechère *et al.*, 2018; van der Kruk & Reijne, 2018). Such equipment is, however, costly and provides vast volumes of data to be analysed. In the absence of such technology, more traditional statistical measures of calculating variability itself include measures such as the standard deviation, coefficient of variation, or the range around a central point from cycle to cycle, as well as over time (Enoka *et al.*, 2003; Cortes *et al.*, 2014; Bravi *et al.*, 2011; Stergiou & Decker, 2011); in other words, assessments of variability should include amplitude- and time-dependent analyses (Cortes *et al.*, 2014). Due to intra- and inter-individual differences, a measure of “relative variability”, defined as variability normalised by the range of motion, should also be considered in the analysis of movement variability (Srinivasan & Mathiassen, 2012).

2.6 TASK-DEPENDENT FACTORS

Given the understanding that interferences to the metabolic and neurological mechanisms during muscle exertions can occur anywhere along the central and peripheral pathways, it is also important to highlight that the processes leading to overt fatigue responses are heavily influenced by a muscle’s task-dependency (Enoka, 1995; Rashedi & Nussbaum, 2015). It is also known that recovery depends on the history of muscle loading (Rashedi & Nussbaum, 2015) and that the rate of muscle recovery is determined by the interplay between exercise duration, intensity, and modality due to the varying contributions of central and peripheral factors to the fatigue process (Carroll *et al.*, 2017). The task-related factors that mediate the muscle’s force production capability and therefore contribute to the development of fatigue, as well as its recovery, include the intensity and duration of force exertion, the exertion type, and the speed of movement (Enoka, 1995; Rashedi & Nussbaum, 2015).

2.6.1 Intensity and Duration of Force Exertion

The intensity of the force exertion and the endurance time of the exertion have an inverse relationship; very high or maximal static exertions have a limited endurance time, while sufficiently low force exertions with small displacements can be maintained for longer periods of time (Sjøgaard & Jensen, 2006). Furthermore, the level of an isometric exertion and a muscle's endurance time have a curvilinear relationship, with an exponential decrease being observed in endurance time with increasing force levels (Rohmert, 1960; Moritani *et al.*, 1981). At the same time, the effort invested increases (Marcora, 2010) until task failure eventually occurs. The manifestations of muscle fatigue under different exertion levels are, however, variable. For example, an experiment by Björksten and Jonsson (1977) found that an isometric force produced by the elbow flexors of 8% of MVE could be maintained for 60 minutes, while an exertion of 15% of maximum, the force proposed by Rohmert (1960) to have indefinite endurance time, could only be sustained for 10 minutes. Furthermore, a study by Sjøgaard *et al.* (1986) revealed that at very low force exertion levels of less than 10%MVE, no biochemical changes, increases in muscle temperature, or restrictions in blood flow could be detected, yet perceptions of effort still increased. A study by Kumar (2001) found that despite strong correlations between the median frequency of EMG and the force of a sustained maximum voluntary exertion, blood oxygenation and blood volume were only moderately correlated with force. This can be explained by the fact that the mechanical pressure generated by the muscle fibres during prolonged exertions has haemodynamic consequences, such as increased systolic and diastolic blood pressure as well as heart rate (Fallentin & Jørgensen, 1992; Kapitaniak, 2001b).

Regarding recovery from fatigue, Yates *et al.* (1987) found that after a dynamic fatigue protocol of 17% MVE, the biceps brachii force production recovered by 35% after the first 30 seconds, but then slowed down to 70% after 7 minutes and just less than 90% after 20 minutes. Furthermore, Sjøgaard & Jensen (2006) reported that fatigue that developed during exercise at high and sustained force levels over short periods of time recovered faster than fatigue induced by prolonged low-load levels, probably due to different biochemical changes in the muscle and reperfusion rates (Carroll *et al.*, 2017). Allen *et al.* (2008) also highlighted that high-frequency fatigue, i.e., the fatigue resulting from muscle stimulations at near-maximal levels, recovered very rapidly, often within a few seconds. This, too, is most likely due to the greater impact of reduced blood flow on the excitation-contraction coupling process during high-force-low-duration activities. On the other hand, low-intensity activities over longer durations are more influenced by central factors, which, due to unknown reasons, take longer to recover (Carroll *et al.*, 2017). Experiments by Kuorinka (1988) revealed that although endurance time was different between the fatigue protocols at low (15%MVE and 30%MVE) and high (60%MVE) exertions, no significant differences were found in the restitution of the EMG spectrum if recovery started from the point of exhaustion.

Force Level and Motor Unit Recruitment

In addition to blood flow to the working muscles (Boyas & Guével, 2011), varying endurance times are also attributed to different recruitment strategies and metabolic processes between high/maximal and low/submaximal static exertions (Fallentin *et al.*, 1993; Potvin & Fuglevand, 2017), meaning that the onset of fatigue also differs between different levels of force exertion. During slow ramp isometric exertions, in other words, exertions that slowly increase in force from rest to maximum, the smaller, and therefore weaker, motor units are recruited first, followed by the appropriate rate coding (Gardiner, 2011). Further increases in force that cannot be achieved with the already recruited motor units then necessitate further recruitment; this time of larger motor units with concurrent adjustments in rate coding (Clamann, 1993; Gardiner, 2011). A maximum voluntary exertion (MVE) is reached when all motor units are recruited, and firing frequencies are at, or near, their maximum (Gardiner, 2011). However, during continuous maximum voluntary isometric muscle actions, motor unit recruitment and firing rates are at their highest at the beginning of the exertion, after which they decline, as evidenced by the reductions in EMG activity (Zwarts *et al.*, 2008; Abd-Elfattah *et al.*, 2015). As a result, the torque produced is at its greatest at the beginning of the exertion and then decreases exponentially over the duration of the exertion (Rohmert, 1960; Allen *et al.*, 2008), despite unchanged or maybe even increased efforts invested by the individual (Gamet & Fokapu 2008).

Tasks performed at sub-maximal levels, however, do not require activation of all motor units, meaning activation of muscles can occur differentially to generate force (Enoka, 1995). To maintain a constant force, new motor units are recruited, while others are de-recruited/drop out (Seghers & Spaepen, 2004; Bawa *et al.*, 2006; Taylor *et al.*, 2016). Such a recruitment strategy allows for short recovery breaks for individual motor units, thus enabling the muscles to prolong the time of force exertion (Enoka, 1995). During sustained submaximal isometric exertions at lower forces, spatial recruitment is preferred over rate coding until the required force level is achieved, while the converse is applicable for higher force levels. Nonetheless, the electromyographical activity appears to increase linearly with force production, although there is some uncertainty about the impact of force level on this linear EMG-force curve (Kamen & Gabriel, 2010). An experiment by Fallentin *et al.* (1993) recorded motor unit recruitment of biceps brachii via an isometric endurance protocol at workloads of 10% and 40% of maximum voluntary exertion. Findings indicated that during low-level exertions of 10% MVE, only one or two motor units with a low firing frequency (10Hz) were active at the start of the protocol. However, as time progressed, new units were recruited to maintain a constant torque, as evidenced by a significantly higher firing frequency and increases in spike amplitudes (Fallentin *et al.*, 1993). An experiment by Zwarts *et al.* (1987) showed that during sustained isometric exertions of 40%MVE the muscle fibre conduction velocity (MFCV) decreased, as did the median power frequency. Evidence of motor unit rotation was also observed, and this was corroborated by Bawa *et al.* (2006). However, the experiment by Fallentin *et al.* (1993) also found that no recruitment of new motor units was detected for protocols with a 40% MVE workload. In addition, the number of action potential spikes with high amplitudes decreased. This seemed to indicate that for muscle exertions at higher workloads, the same motor units were recruited under fatigued conditions as at the start of the protocol, and that adjustments in action potential shapes and durations occurred to maintain the isometric task performance.

The Role of Blood Flow in Muscle Functioning

Arguably, the most cited reason for the effects that force intensity has on muscle fatigue development, and therefore endurance time, is the restriction in blood flow (Hietanen, 1984; Sjøgaard *et al.*, 1988; Murthy *et al.*, 2001; Boyas & Guével, 2011). Force development in a muscle relies on the conversion of chemically bound energy to mechanical energy (Tortora & Derrickson, 2009), and while some chemicals are found within the muscle, such as limited amounts of ATP-CP or glycogen, other substrates, including glucose and free fatty acids, have to be supplied through blood flow, particularly during prolonged and high-intensity activities (Sahlin *et al.*, 1998; Sjøgaard & Jensen, 2006). Furthermore, certain metabolic processes require the presence of oxygen to release energy from the substrates (McArdle *et al.*, 1996). The concept of the arterio-venous pressure gradient proposes that changes in intramuscular pressure influence the amount of blood flowing to the working muscles (Humphreys and Lind, 1963; Hietanen, 1984; Sjøgaard *et al.*, 1988), thereby affecting the supply of nutrients and oxygen as well as waste product removal. During exercise, afferent feedback from the muscles interacts with the cardiovascular and respiratory processes, increasing systemic blood pressure and thereby perfusion pressure, which in turn increases blood flow to the working muscles (Barcroft and Millen 1939; Petrofsky & Hendershot, 1984; Fallentin & Jørgensen, 1992; Taylor *et al.*, 2016). Adequate blood flow is essential for the optimal functioning of a muscle since it supplies oxygen and substrates, and removes heat and metabolites (e.g., hydrogen and phosphate ions) (Sjøgaard *et al.*, 1988; Westerblad *et al.*, 1998; Boyas & Guével, 2011), thereby prolonging the muscle's endurance time (Petrofsky & Hendershot, 1984; Fallentin & Jørgensen, 1992).

Despite the initial increases in perfusion, blood flow has also been shown to decrease under certain load levels and exertion types, particularly static exertions (Barcroft and Millen, 1939; Oranchuk *et al.*, 2020). Impaired circulation could explain the development of muscle fatigue, particularly during sustained high-level isometric actions, since there is a mismatch between energy supply and energy breakdown during such muscle exertions (Sjøgaard *et al.*, 1988; Noakes, 2000). This appears to occur once the intramuscular pressure exceeds the venous blood pressure, resulting in the blood flow slowing down and ultimately being occluded (Sjøgaard *et al.*, 1988).

The more muscle fibres that are activated during high-level exertions, the greater the occlusion (Hietanen, 1984). Intramuscular pressure does, however, vary considerably between and within muscles, making it difficult to determine at what levels of maximum exertion blood flow is sufficiently restricted to contribute to fatigue. The impact of intramuscular pressures on blood flow varies depending on the muscle itself, different parts within the muscle, the muscle fibres, and connective tissues (Sjøgaard *et al.*, 1988), which explains the diversity reported on in the literature. For example, Kapitaniak (2001b) found that beyond an exertion level of 25% of maximum voluntary exertion (MVE), the increases in systemic blood pressure appeared to be no longer adequate to overcome the intramuscular pressure. Similarly, Barcroft and Millen (1939) and Edwards (1975) found that blood flow was already partially occluded in the leg muscles when forces exceeded 20% MVIE, while a study by Barnes (1980) showed that forearm blood flow was maximal with contractions of 20-25% MVIE but declined thereafter. Mortimer *et al.* (1971) indicated that force values higher than 30%MVE severely restricted blood flow, while Fallentin and Jørgensen (1992) found the same at 40%MVE. Humphreys and Lind (1963) reported that blood flow to the forearm increased during static exertions of up to 50% of maximum voluntary isometric exertion (MVIE), and extrapolations of their data suggested that complete occlusion occurs at levels exceeding 70% MVIE.

Even though it was assumed that at low enough loads (i.e., 5-10% of maximum), the blood pressure would exceed the intramuscular pressure generated by the exertion, and thereby ensure sufficient blood flow to the working muscle (Jørgensen *et al.*, 1988), it was also pointed out that the complex microcirculatory regulation may become impeded at such low levels of force exertions (Sjøgaard *et al.*, 1988; Sjøgaard & Jensen, 2006). A study by Fallentin and Jørgensen (1992) revealed that physiological responses at 10% MVE were the same as those at around 40% MVE (Fallentin & Jørgensen, 1992). It has further been suggested that during low-level exertions and thus low blood flow velocities, the diastolic blood pressure is the deciding factor for maintaining blood flow, rather than the mean of systolic and diastolic blood pressures (Sjøgaard & Jensen, 2006). Perfusion was found to decrease at values of between 5% and 15% of MVIE (Sejersted *et al.*, 1984; Sjøgaard *et al.*, 1986), with complete occlusion occurring around 50%-60% of MVIE in tibialis anterior (McNeill *et al.*, 2015), and between 51% and 75% of maximum hand grip strength (Barnes, 1980). Similarly, findings by Sjøgaard *et al.* (1988) showed that blood flow in the quadriceps during knee extension, as well as during handgrip exertions, slowed down at only 10% of MVE and was almost completely occluded between 40% and 50% of MVE. These findings led Sjøgaard *et al.* (1988) to conclude that muscle exertion levels had to be below 10% of MVE to maintain homeostasis.

A further effect of high intramuscular pressure is that the muscle's water content also increases, thereby increasing the thickness of the muscle (Sjøgaard & Jensen, 2006), which in turn raises tissue pressure and further obstructs blood flow. Differences in muscle fibres and connective tissues influence intramuscular pressures, which can explain the large variations between muscles, but also within different parts of a muscle. It is, therefore, virtually impossible to determine at what contractile levels blood flow is restricted to induce fatigue, and which may be the reason for the inconsistencies in the literature (Sjøgaard *et al.*, 1988).

2.6.2 Type of Muscle Exertion

Although the effects of force/exertion level on muscle fatigue dominate the literature on task-related factors, it is also important to interrogate the role that different types of muscle actions play in fatigue development since different exertion types yield divergent responses due to varying physiological mechanisms (Krüger *et al.*, 2019). What has been well-established in the literature is that maximal force exertion and neural activation differ between static and dynamic muscle actions and that fatigue-related changes in one exertion type cannot be transferred to the other (Cheng and Rice, 2005). However, less is understood about how motor unit recruitment and force production vary under fatiguing conditions during dynamic (both concentric and eccentric) muscle actions (Enoka, 1996; Søgaard *et al.*, 1996).

In an unfatigued state, a muscle's force-producing capacity varies according to the type of muscle action it performs, as the active movement of a limb through a range of motion entails accelerations and decelerations, as well as changes in the length-tension relationship which influence the muscle's strength producing capacity (Kroemer, 2001; Freivalds, 2004; Lieber & Ward, 2011). Eccentric exertions, in particular, appear to stand apart from concentric and isometric exertions in terms of their mechanical and electromyographic responses and energy requirements (Pasquet *et al.*, 2000). When comparing static with dynamic exertions, it is well-established that under unfatigued conditions, a muscle can produce considerably greater force during eccentric exertions, and the least under concentric conditions (Doss & Karpovich, 1965; Linnamo *et al.*, 2000; 2003; Freivalds, 2004). For example, Seliger *et al.* (1980) found that during a whole-body weightlifting exercise using the knees, eccentric exertions resulted in a maximal tension three times greater compared to concentric actions, and with maximal isometric strength falling between concentric and eccentric forces. A study by Doss and Karpovich (1965) found a 23% difference in maximal strength production between the eccentric and isometric exertions, but only a 13.5% difference between maximal concentric and isometric force exertions. However, inter-individual variability in force production was greater for the eccentric exertions, compared to the concentric ones (Doss & Karpovich, 1965). The reason for a muscle's greater eccentric force-producing capacity is that breaking the protein cross-bridges during eccentric exertions requires a greater force, than merely maintaining the cross-bridges during an isometric exertion (Freivalds, 2004). During a lengthening exertion, the stretch in the muscle can resist a greater tension before the cross-bridges undergo mechanical separation. Eccentric exertions are, therefore, also less energy-demanding than concentric muscle actions (Abbott *et al.*, 1952). Maintaining the actin and myosin bonds requires less ATP hydrolysis and a lower neural input during eccentric exertions (Enoka, 1996) compared to concentric and isometric exertions, meaning the cross-bridges create greater tension up to the point when they are eventually disrupted (Pasquet *et al.*, 2000).

In addition to maximal force-producing capabilities, neural activation patterns also differ between static and dynamic exertions (Komi & Tesch, 1979; Cheng & Rice, 2005). Coordinating a movement requires careful modulation of the motor unit discharge rate to control the lengthening or shortening of the muscle, the transition from lengthening to shortening, or vice versa (Enoka & Duchateau, 2017). Isometric exertions allow for a fine-tuning of spatial and temporal recruitment of motor units to achieve a relatively consistent force output, irrespective of whether the force exerted is at maximal or sub-maximal levels (Chaffin *et al.*, 2006; Duchateau & Baudry, 2014). Eccentric exertions, however, recruit more motor units compared to other exertion types (Pasquet *et al.*, 2000), but each motor unit has a lower discharge rate (Grabiner & Owings, 2002). On the other hand, Duchateau and Enoka (2016) proposed that the recruitment order does not differ between concentric and eccentric exertions. For example, at the same level of force, eccentric muscle actions have a lower motor unit activity compared to concentric exertions (Duchateau & Enoka, 2016). There is thus still a lack of consensus regarding the detailed motor unit recruitment responses between concentric and eccentric exertions. Furthermore, Madeleine *et al.* (2001) and Enoka and Fuglevand (2001) found significantly lower EMG amplitudes during eccentric exertions compared with isometric muscle actions, but Linnamo *et al.* (2000) found no difference in the mean EMG amplitude between the two dynamic exertion types. Linnamo *et al.* (2003) did, however, also report lower mean spike frequencies for eccentric exertions at different force levels, but this difference was only significant at 40% and 60% of maximum force. The reason for this may be that motor units recruited during eccentric exertions are high thresholds units, i.e., units that need strong stimulation to be activated (Enoka and Fuglevand, 2001), and which are able to produce the greatest force. Enoka and Fuglevand (2001) also highlighted that EMG amplitude was greater and recruitment thresholds were lower during concentric exertions due to a greater number of motor units being recruited. However, the experiment by Linnamo *et al.* (2003) showed inconsistent results in the mean spike amplitude between concentric and eccentric exertions and pointed out that preactivation of a muscle influenced both mean spike frequency and mean spike amplitude.

After a fatigue-inducing exercise protocol with different muscle actions, significant changes have been identified in various neuromuscular parameters, including torque, the EMG, muscle conduction velocity, as well as blood lactate concentrations (Pasquet *et al.*, 2000; González-Izal *et al.*, 2014). With regards to force produced, the study by Pasquet *et al.* (2000) revealed a 31.6% decrement in concentric torque produced by the ankle dorsiflexor muscles over the duration of a fatiguing exercise, while eccentric torque only decreased by 23.8%. The same trend was noticed by González-Izal *et al.* (2014) using the vastus lateralis muscle during a knee extension protocol and which could be attributed to the reduced energy demand of eccentric exertions (Enoka & Stuart, 1992). Linnamo *et al.* (2000), however, found that even though the absolute decrease in average force was significantly greater during the eccentric exertions, once relativized to its unfatigued force, the difference in force decrements between concentric and eccentric exertions became non-significant (50% and 53% decrease, respectively). Similarly, maximum voluntary isometric exertions were found to decrease to a greater extent after a concentric exercise protocol (37%), compared to an eccentric protocol (34%), although the difference was not significant either (Pasquet *et al.*, 2000).

In determining the main contributing factor(s) in localised fatigue, Iridiastadi and Nussbaum (2006) claim that reductions in a muscle's force-generating capacity with fatigue progression are attributed to changes in the EMG spectra, which, in turn, are subject to different motor unit recruitment strategies; in other words, central factors. Evidence of how the electromyographic signal responds to different muscle actions is, however, contradictory. Although it is generally accepted that EMG amplitude increases with muscle fatigue development (e.g., Enoka & Duchateau, 2008; Al-Mulla *et al.*, 2011), the study by Linnamo *et al.* (2000) showed no significant changes in the average EMG amplitude for biceps brachii for both concentric and eccentric muscle actions. However, the same study found inconsistent results in the EMG amplitude for brachioradialis and triceps brachii, which could indicate some compensatory mechanisms. Other studies, however, show that concentric muscle activation values are significantly higher than the eccentric ones (Pasquet *et al.*, 2000; González-Izal, 2014), which could be explained by motor units being less readily activated during maximal concentric exertions, thus requiring a greater activation (Enoka & Stuart, 1992). Furthermore, median power frequency was also found to decrease significantly for both concentric and eccentric exertions (after concentric and eccentric exercise, respectively) (Linnamo *et al.*, 2000), as did muscle fibre conduction velocity, even though there was no difference between concentric and eccentric exercise (González-Izal, 2014).

While central factors are known to influence the EMG signal to the working muscles (Bigland-Ritchie & Woods, 1984; McNeil *et al.*, 2009), Pasquet *et al.* (2000) argued that, based on their findings of the M-wave analysis, the reductions in force with fatigue progression could not be attributed to central factors such as changes in neural activation, but pointed out that the dominant cause was related to peripheral mechanisms such as the intracellular changes controlling the excitation-contraction coupling processes. Evidence for this is, however, also contradictory. For example, the study by Linnamo *et al.* (2000) attributed the significant decreases in the median power frequency after a fatigue protocol, at least in part, to reduced muscle fibre conduction velocity as well as elevated blood lactate concentration. González-Izal *et al.* (2014) also indicated greater increases in blood lactate concentrations under concentric compared to eccentric exercise but found that muscle fibre conduction velocity did not differ between the two dynamic exertion types. It is speculated that the propagation of the neural signal is impeded by the accumulation of potassium ions in the transverse tubules, thus interfering with the excitation-contraction coupling process (Pasquet *et al.*, 2000). However, Tenan (2009) concluded that changes in potassium and lactate concentrations did not significantly relate to changes in a muscle's frequency spectrum after a fatiguing cycling task.

Given the significant role that blood flow plays in the metabolic processes of muscle exertions (Barcroft & Millen, 1939; Oranchuk *et al.*, 2020), it is plausible that variations in muscular force exertions during different exertion types can influence muscle fatigue development, due to peristaltic effects which in turn may promote blood flow (Humphreys & Lind, 1963; Tschakovsky *et al.*, 1996). For example, Barcroft and Millen (1939) highlighted that "rhythmic exertions" led to increases in blood flow, which in turn could delay fatigue development. Sjøgaard *et al.* (1988) proposed that this was the result of repeated contractions and relaxations since intramuscular pressure increased during a dynamic exertion, but decreased at rest, thus resulting in varying blood flow. For rhythmical skeletal muscle exertions, Tschakovsky *et al.* (1996) found significantly higher blood flow than if the pressure was applied via a cuff. These researchers, as well as Hamann *et al.* (2003), however, pointed out that another mechanism other than muscular force production alone was involved in muscle perfusion, such as vasodilation at the onset of the activity, and that this may be the reason for the inconsistency in research findings.

Intramuscular pressure and blood flow are, however, not only linked to variations in force, but also muscle length, as demonstrated in a study by Poole *et al.* (1997). These authors found that as a muscle was elongated while moving through a range of motion, and sarcomere length increased, capillaries became less tortuous. From a certain sarcomere length onwards, capillaries also became more stretched, and smaller luminal diameters were recorded. Additionally, capillary shapes also became more elongated, indicating the blood had to "squeeze" through the capillaries (Poole *et al.*, 1997). Poole *et al.* (1997) concluded that the mean capillary red blood cell velocity was most likely the result of increased resistance, which hampered blood flow while the muscle was stretched. Similarly, Sjøgaard *et al.* (1988) highlighted that blood vessels are prone to mechanical influences since they are flexible and pliable and can therefore distort and 'collapse'. Such changes in the blood vessels' diameter would ultimately increase the resistance within the blood vessels. Muscle length has also been reported to influence muscle fibre conduction velocity in studies by Aljure and Borrero (1968) and later Arendt-Nielsen *et al.* (1992), but no studies were found investigating any differences in muscle perfusion and, by extension, muscle fatigue development during conditions of varying muscle length.

2.6.3 Movement Speed

In addition to the magnitude of the force exerted, and the type of exertion, Enoka *et al.* (2003) highlight that the speed of the movement also influences the amount of motor unit activity during a muscle exertion, and consequently muscle fatigue development. During dynamic muscle actions, one significant factor influencing the maximum force production capacity of a muscle is the velocity at which the limb moves, as well as the direction in which it moves (Wilkie, 1950; Oatis, 2004). The so-called “force-velocity relationship” states that a faster movement speed will result in lower force output, and vice versa, although not necessarily in a linear manner (Fenn & Marsh, 1935; Perrin, 1993; Alcazar *et al.*, 2019). The physiological mechanism underlying the force-velocity relationship is that of the time it takes for cross-bridges to be formed. In fast-twitch fibres, the cross-bridge cycle speed is considerably greater compared to that of the slow-twitch fibres, which ultimately means that muscles with a high proportion of fast-twitch fibres can generate greater contractile velocities and power than muscles with more slow-twitch fibres (ACSM, 2006). Furthermore, concentric exertions require shortening of the sarcomeres, and for this to happen, energy must be liberated in proportion to the work done (Fenn & Marsh, 1935). At faster velocities, there is little time for cross-bridges to form and interact as the filaments slide past one another during a shortening exertion. As only a few cross-bridges are formed, the muscle’s force-producing ability is limited (Lieber & Ward, 2011), resulting in an exponential decrease in the force-velocity curve with increasing movement velocities during concentric exertions (Fenn & Marsh, 1935; Katz, 1939; Oatis, 2004). During slower shortening velocities, more time for cross-bridge formation is available, resulting in higher force output. However, the force-velocity relationship under eccentric exertions reveals that with increases in lengthening velocity, force production increases rapidly from the isometric baseline measure but then tapers off to reach a plateau. The reasons for this finding are poorly understood due to the complex interactions between force, velocity, and time, as well as the developing stiffness within a muscle while lengthening (ACSM, 2006).

What can be concluded from the information presented in the preceding sections is that muscle fatigue is multifaceted with a variety of central and peripheral causal factors, as well as mechanisms that attempt to maintain the required motor performance despite ever-increasing decrements. Furthermore, task-dependent factors determine the rate of muscle fatigue development and recovery, thus adding to the complexity of the topic. Another challenge encountered in the study of fatigue is how to quantify localized muscle fatigue.

2.6.4 Recovery from Muscle Fatigue

Although the logarithmic recovery curve appears to have a similar shape for a variety of variables, recovery rates do, however, seem to vary between fatigue indicators as well as different fatiguing mechanisms (Yates *et al.*, 1987). For example, Rohmert (1960) and Zwarts *et al.* (1987) found a similar trajectory for recovery of force and muscle fibre conduction velocity, while Kuorinka (1988) also showed an initial rapid restitution in the median power frequency of the EMG spectrum after exercise, which then tapered off to a plateau beyond 5 minutes after cessation of the activity. Hara (1980), however, reported that recovery time of the EMG spectrum was shorter than mechanical and physiological recovery, and McDonald *et al.* (2015) found that perceived fatigue quickly returned to pre-fatigue levels, although kinematic and muscular changes remained. Furthermore, a study by Micklewright *et al.* (2017) discovered that during rest periods following a fatiguing task, RPE quickly dropped to zero, while physiological indicators of muscle fatigue, such as blood lactate concentration and respiratory exchange ratio (VCO_2/VO_2), were still elevated, as compared to baseline values obtained prior to the fatiguing task. Recovery times have been reported to range from as brief as 5-6minutes after cessation of a physical activity (Hara, 1980), to as long as 30 minutes post-exercise (Pasquet *et al.*, 2000).

The quick recovery of muscle force during the initial phase of rest can be explained by the increased reperfusion; in other words, the return of blood to the exercised muscle (Hara, 1980; Carroll *et al.*, 2017), as well as the return to intramuscular homeostasis of calcium and potassium ions (Allen *et al.*, 2008). Since fatigue development does not occur purely due to peripheral factors, but also central / supraspinal factors (Carroll *et al.*, 2017), it could be assumed that recovery, too, is reliant on a combination of central and peripheral processes. Zwarts *et al.* (1987) found that recovery was not solely dependent on the return of blood flow to the biceps brachii muscle and speculated that central factors were involved in the recovery of a muscle after prolonged exertions. The slower recovery to full capacity thereafter seems to be largely influenced by central processes (Carroll *et al.*, 2017) and could be the reason why the relationship between recovery time and activity time is not linear, but rather exponential (Sjøgaard & Jensen, 2006).

When comparing recovery times after various exercise modalities, Kuorinka (1988) found very similar recovery trajectories after static versus dynamic exercise, although the recovery curves from the dynamic exercise were less regular (smooth) than for the isometric exercise. Yates *et al.* (1987), on the other hand, concluded that recovery of dynamic endurance is slightly more rapid than recovery of static endurance. Linnamo *et al.* (2000) pointed out that concentric and isometric exertions can take up to some hours to recover, while eccentric exertions can take days, if not weeks, to recover to full strength capacity if muscle fibres are damaged. Linnamo *et al.* (2000) also reported significant increases in serum creatine kinase levels and muscle soreness two days after eccentric exertions, but not concentric exercise. This could imply damage to muscle fibres, which in turn could explain the longer recovery time (Linnamo *et al.*, 2000). Furthermore, Pasquet *et al.* (2000) found that although peak torque after a concentric exercise protocol had decreased to a greater extent compared to the eccentric protocol, the latter condition had a slower recovery of peak torque. Additionally, even though recovery of EMG followed different time courses for the concentric and eccentric conditions, no significant difference was found between these two conditions for peak torque after 30 minutes of recovery (Pasquet *et al.*, 2000). Similarly, a study by González-Izal *et al.* (2014) found that although peak concentric torque had decreased to a greater extent than eccentric peak torque upon cessation of the protocol, they found no significant difference between the two exertion types between minute 0 and minute 5 post exercise for peak torque, conduction velocity, EMG root mean squared amplitude, and the Dimitrov fatigue index. Furthermore, even though EMG median power frequency (MdF) was significantly lower for the concentric condition immediately after the fatiguing exercise compared to the eccentric condition, after five minutes of rest, no differences were found MdF between the two exertion types. A further finding of the González-Izal *et al.* (2014) study was that blood lactate concentrations remained significantly higher for the concentric condition throughout the 5-minute recovery period.

In conclusion, muscle actions are influenced by the muscle length, muscle force and movement speeds. Each of these factors is task-dependent and governed by different mechanisms, including different recruitment strategies, levels of blood supply, secretion of neurotransmitters and levels of motivation. However, the effects of the interrelationship between task-dependent factors on localised muscle fatigue development as well as its recovery remain poorly understood. More specifically, most activities at work, sport and daily living consist of static and dynamic muscle exertions at varying levels of intensity, and this interaction therefore warrants further research.

CHAPTER 3: RESEARCH CONCEPT

Muscle fatigue is acknowledged to be a multifactorial phenomenon consisting of complex interactions between a variety of mechanisms occurring at central and peripheral levels (Rashedi & Nussbaum, 2015). The literature review in Chapter 2 highlights that decades of research using various approaches to fatigue modelling have identified a multitude of interacting circulatory, metabolic, chemical, neurological and psychological mechanisms that shape the fatigue process. Furthermore, muscle fatigue is the outcome of a unique process or combination of processes (Abd-Elfattah *et al.*, 2015), and the characteristics of a task influence its development as well as recovery (Bigland-Ritchie *et al.*, 1995). The principle of task dependency highlights that the task requirements for producing a specific force output, for example, whether an activity is static or dynamic, maximal or submaximal, continuous or intermittent, as well as its duration, frequency and force magnitude, influence the types of muscle actions performed (Iridiastadi & Nussbaum, 2006; Yassierli & Nussbaum, 2007), which in turn determine the fatigue processes (Enoka & Stuart, 1992; Barry & Enoka, 2007; Rashedi & Nussbaum, 2015). Considerations of such task characteristics have particular relevance in occupational contexts, where fatigue development under intermittent exertions and submaximal levels prevails (Rashedi & Nussbaum, 2015). The premise upon which this study is based is that the traditional way dynamic muscle actions are classified ignores the subtle differences that may characterise different muscle exertions, particularly dynamic ones. While, currently, muscle length (either the change thereof, or lack of change) has been the determining characteristic for defining and naming muscle actions, there is sufficient evidence in the literature highlighting additional task parameters (e.g., muscle force or contractile speeds) as important determinants of the circulatory, metabolic, chemical, neurological and psychological mechanisms that occur during the contractile process, and which in turn may influence the development of muscle fatigue (refer to the Review of Literature in Chapter 2). The purpose of this chapter is to outline the research concept underpinning the current study.

3.1 RESEARCH QUESTIONS

The overriding purpose of this research was to investigate whether different task-related characteristics of muscle exertions, specifically muscle length and muscle force, contribute to different outcomes relating to localised muscle fatigue under continuous and submaximal conditions. While the effects of the magnitude of muscle loading on fatigue development are well-established, it is less well known how continuous exertions at intermittent levels of force (i.e., varying force) influence fatigue development, compared to the commonly used singular force in most studies. Furthermore, the force generated by the muscle can occur during different actions, yet the prevalent terminology for the types of muscle exertions is limited to the change in muscle length (or absence thereof), without taking into consideration the concurrent force production. The assumption that combinations of muscle force and changes in muscle length would influence fatigue development was based on findings in the literature that highlight the important role of blood flow in the efficient functioning of the working muscles. Adequate blood flow can, however, be inhibited by vasoconstriction, and evidence indicates that both muscle length and muscle force can reduce the diameter of the arteries and veins that supply oxygen and nutrients and remove waste products, all of which have been suggested to interrupt the peripheral processes required for effective muscle functioning. It was anticipated that static exertions would induce greater fatigue by reducing the blood flow to the working muscles, whereas dynamic exertions, i.e., those exhibiting changes in muscle length, would be more fatigue-resistant (e.g., Masuda *et al.*, 1999; Baxi *et al.*, 2017), since the changes in muscle length could have a peristaltic effect, thereby promoting blood flow. Similarly, exertions at a set muscle force were expected to result in greater fatigue compared to exertions that involved rhythmic variations in muscle force, which may also have a peristaltic effect. However, these two factors (i.e., changes in muscle force and changes in muscle length) have never been studied in combination with one another. It was further important to determine whether the muscle fatigue developing from combinations of muscle length and force would also impact the recovery of muscle fatigue once the fatiguing activity had ceased.

The current study was therefore guided by the following questions:

- Does a muscle exertion characterized by a lack of movement (and therefore a consistent muscle length) fatigue faster than a muscle that moves a joint through a range of motion, thereby varying the muscle length?
- Does a muscle action generating a consistent force output fatigue faster than an exertion that varies between a minimum and maximum force, yet does not change in length?
- Is the recovery process influenced by the type of exertion and loading that have led to muscle fatigue?

3.2 EXPERIMENTAL FRAMEWORK

Chapter 2 discussed the shortfalls of the current classification of muscle actions and related terminology. It highlighted that muscle length, muscle force and movement speed are dominant task-dependent characteristics that should be considered in fatigue research. However, the conventional classifications only consider one parameter when describing a muscle action. The matrix in Figure 1 depicts the conceptual framework used in the current study to classify muscle exertions, acknowledging that some characteristics of muscle exertions can occur in combination with one another and that this may influence the underlying mechanisms leading to muscle fatigue during sub-maximal continuous/prolonged activity. More specifically, this research focussed on assessing the effects of two task-dependent variables on muscle fatigue and recovery, namely muscle length and muscle force, in isolation, as well as in combination with one another. This study did not aim to identify the underlying physiological mechanisms of fatigue, but rather to quantify the differences in fatigue development and recovery under different muscle exertion types using a novel approach to classifying different types of muscle exertion.

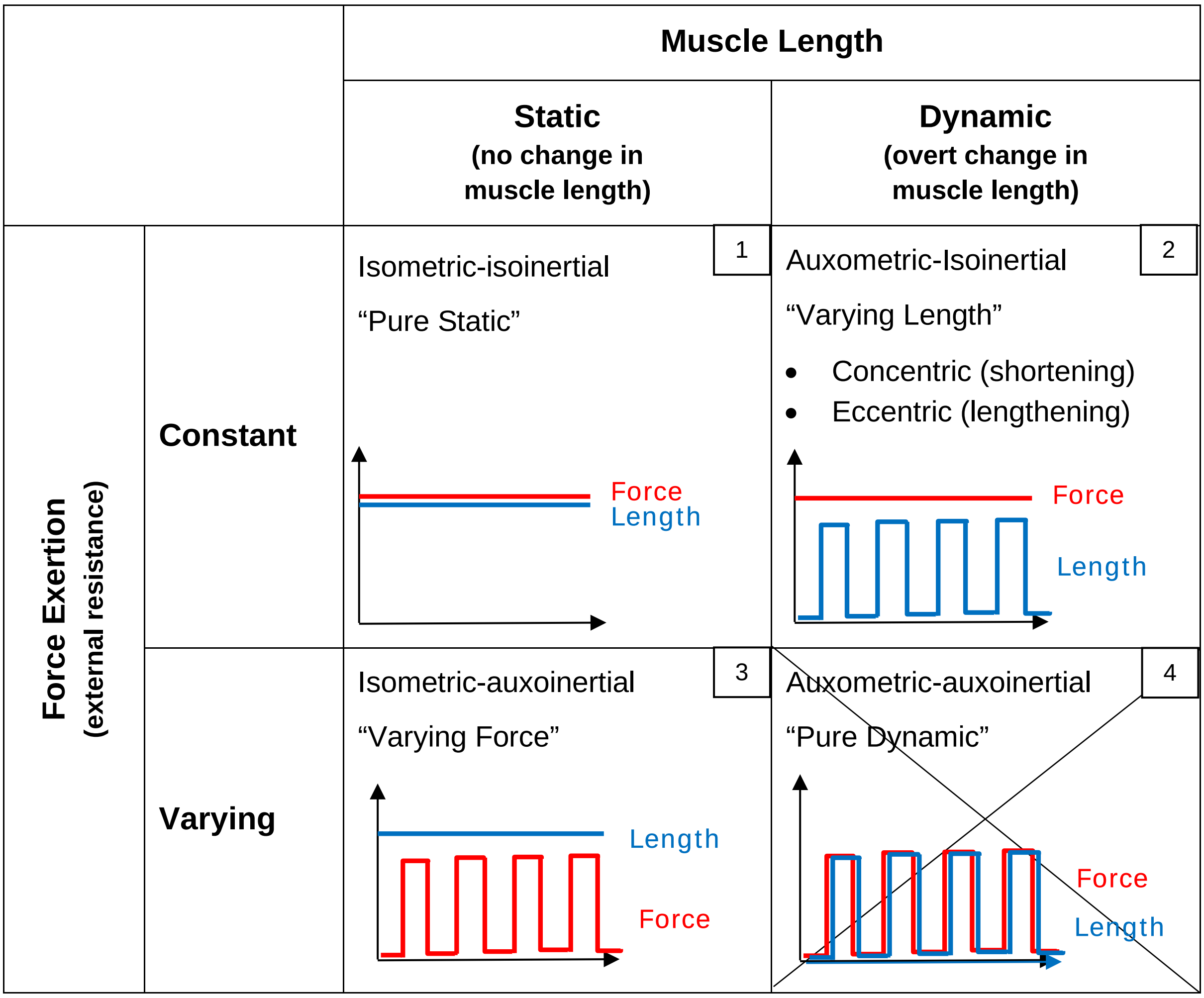


Figure 1: Conceptual matrix of types of muscle exertions using combinations of muscle length and force exertion.

Fields 1-4 depict the terminology used for the different combinations of muscle length and force exertion. Field 1 shows a typical static exertion with no variations in muscle length or force ("Pure Static"), while field 4 depicts a dynamic exertion with changes in both these parameters ("Pure Dynamic") (this type of exertion was eliminated from the current study and is hence crossed out). The remaining fields show variations in only one of the two parameters, i.e., either length (field 2 – "Varying Length") or force (field 3 – "Varying Force"), while the second factor remains constant.

In each of the fields in Figure 1, force exertion (constant vs varying) and muscle length (static vs dynamic) are tested in combination with one another. The impact of force exertion levels on the development of muscle fatigue was investigated by comparing a muscle exerting a constant force to varying levels of force under isometric conditions, in other words, with no change in the muscle's length (Figure 1 – fields 1 and 3, respectively). Conversely, the influence of muscle length on fatigue development, i.e., applying a force to a constant external resistance, in the presence and absence of movement, was determined by comparing fields 1 and 2 in Figure 1. Finally, although realistically, most real-life movements fall under the category that combines both variations of muscle length with variations of muscle force (Figure 1 – field 4), this condition was ultimately eliminated from the experimental design, since it presented methodological and analytical challenges relating to the standardization of the fatigue and testing protocols. From a biomechanical perspective, as a limb moves through a range of motion, changes occur in muscle length, movement speed and moment arm, alongside the varying force, which would have been impossible to control. This would also have presented countless options in coordinating the timing of minimum and maximum joint angles and forces. From a technical perspective, sophisticated equipment such as an isokinetic dynamometer does not allow for the real-time force adjustments necessary for a "Pure Dynamic" condition. These challenges would have resulted in significant analytical difficulties when accurately quantifying muscle fatigue. Since the "Pure Dynamic" condition was excluded, the design concept for the current study involved an incomplete two-factorial design.

To describe muscle exertions according to the muscle's length and force produced, and yet to facilitate ease of understanding, the following terminology describing the various muscle actions is used in this thesis:

- *"Pure Static" exertion (Figure 1 – field 1)*: This is the equivalent of an "isometric-isoinertial" exertion; in other words, a muscle action that varies neither in muscle length, nor the external force it resists. This exertion would be the equivalent to a "holding isometric muscle action", or "constant force isometric exertion" where the muscle length remains static and the muscle force is constant.
- *"Varying Length" exertion (Figure 1 – field 2)*: This auxometric-isoinertial exertion is a muscle action that experiences dynamic variations in muscle length due to changing joint angles, by either shortening (concentrically) or lengthening (eccentrically), while the external resistance remains constant.
- *"Varying Force" exertion (Figure 1 – field 3)*: Isometric-auxoinertial exertions maintain a constant muscle length (i.e., no voluntary movement occurs), while the

force generated by the muscle against an external resistance varies. This exertion has also been referred to as a "pushing isometric muscle action".

The above framework forms the basis of the experimental design and methods described in the following chapter. Participants were subjected to a fatigue protocol, during which they had to produce continuous sub-maximal exertions for each of the exertion types mentioned above (fields 1-3). A variety of indicators of muscle fatigue were recorded before, during and after the fatigue protocol and from which the development of localised muscle fatigue as well as its recovery could be inferred.

CHAPTER 4: METHODOLOGY

4.1 EXPERIMENTAL DESIGN

To investigate the effects of exertion type on localised muscle fatigue, this study employed a traditional empirical research approach. While the research concept in Chapter 3 depicts a two-factorial design with muscle length and muscle force as factors, the incomplete nature of this design was not feasible for a two-factorial statistical analysis due to the ‘missing’ condition (varying muscle length and varying muscle force in Figure 1; Field 4). Furthermore, given that physiological events underlying fatigue cannot be immediately detected (Bigland-Ritchie & Woods, 1984), most fatigue indicators can only infer its development, as well as recovery from fatigue, by monitoring changes over time. Time was therefore added as a factor, even though it was not included as one in the matrix depicted in the research concept (Figure 1). This meant devising an alternative version of a two-factorial design for the purposes of conducting inferential statistics (Table 1) and to address the following objectives:

1. To establish the effects of time on muscle fatigue development as well as recovery from muscle fatigue.
2. To identify differences in fatigue development and recovery between the selected exertion types.

Table 1 depicts a repeated-measures two-factorial design with time and exertion type (which combines all three combinations of muscle strength and muscle force) as factors.

Table 1: Two-factorial Design Matrix for the statistical analysis of fatigue development according to exertion type.

		Factor 2: Exertion Type		
		Varying Force	**Pure Static**	**Varying Force**
Factor 1: Time	**Pre-Fatigue**			
	Post-Fatigue-Min 0			
	Post-Fatigue-Min 5			

4.2 STATISTICAL HYPOTHESES

4.2.1 Hypothesis 1: Effects of time on muscle fatigue development and recovery

The first hypothesis considered the effects of time on the development of muscle fatigue, as well as its recovery. This hypothesis was included as a so-called "manipulation check" to confirm the effectiveness of the fatigue protocol. In other words, it first needed to be verified that the exercise protocol had indeed induced muscular fatigue before being able to compare the differences in fatigue responses between the experimental conditions (Hypothesis 2).

Hypothesis 1(a)

It was hypothesized via the alternative hypothesis (H_A) that time-dependent changes would occur in the selected fatigue parameters (as detailed in section 4.4) between the start of the fatigue protocol and the end. The null hypothesis (H_O) proposed no significant differences in the selected fatigue parameters over the duration of the fatigue protocol.

$$H_O = \mu_{\text{Pre Fatigue Protocol}} = \mu_{\text{Post Fatigue Protocol}}$$

$$H_A = \mu_{\text{Pre Fatigue Protocol}} \neq \mu_{\text{Post Fatigue Protocol}}$$

where µ refers to the participants' responses as per selected fatigue indicators before ('pre') and at the end and after the fatigue protocol ('post').

Hypothesis 1(b)

This hypothesis focused on the recovery process, where the H_A proposed time-dependent improvements in the selected fatigue parameters at the end of the fatigue protocol compared to the start, while the H_O supported no change during the recovery period.

$$H_O = \mu_{\text{Start Recovery Period}} = \mu_{\text{End Recovery Period}}$$

$$H_A = \mu_{\text{Start Recovery Period}} \neq \mu_{\text{End Recovery Period}}$$

where µ refers to the responses of the selected fatigue indicators at the start and at the end of the recovery period.

4.2.2 Hypothesis 2: Effects of exertion type on muscle fatigue

The second hypothesis proposed that the type of exertion would influence the development of fatigue and the recovery thereof.

Hypothesis 2(a)

Here, it was hypothesized that a muscle action with no changes in muscle length (i.e., "Pure Static" exertion) would result in faster muscle fatigue development compared to a muscle that had a dynamic component (i.e., movement) (H_A). The H_O predicted no effect of muscle length on the rate of fatigue development.

$$H_O = \mu_{\text{Static muscle length}} = \mu_{\text{Dynamic muscle length}}$$

$$H_A = \mu_{\text{Static muscle length}} \neq \mu_{\text{Dynamic muscle length}}$$

where µ refers to the fatigue indicators under conditions with static or dynamic muscle lengths.

Hypothesis 2(b)

The final hypothesis proposed that consistent force generation would significantly influence muscle fatigue development compared to exertions with a varying force (H_A), while acceptance of the H_O would indicate no effect of muscle force on fatigue development.

$$H_O = \mu_{\text{Constant force exertion}} = \mu_{\text{Varying force exertion}}$$

$$H_A = \mu_{\text{Constant force exertion}} \neq \mu_{\text{Varying force exertion}}$$

where µ refers to the fatigue indicators under conditions of constant vs varying force exertions.

4.3 SELECTION OF INDEPENDENT VARIABLES

In line with the research design matrix depicted in Table 1, the independent variables selected for this study were time and type of muscle exertion, the latter of which consisted of considerations of muscle length and muscle force on muscle fatigue.

4.3.1 Time

Since fatigue is a time-dependent process (Søgaard *et al.*, 2006; Barry & Enoka, 2007; Enoka & Duchateau, 2008) and can thus only be determined by monitoring temporal changes in a muscle's force-generating capacity and other performance-related outcomes (De Luca, 1984; Seghers & Spaepen, 2004), time was selected as one factor in the study's two-factorial design. Localised muscle fatigue was therefore quantified by recording various fatigue indicators at the start and the end of a fatigue protocol. The duration of the protocol was determined by each study participant's individual level of fatigue development, until they reached a level of near-exhaustion (RPE of 9). Similarly, recovery from fatigue was determined by assessing changes in the selected indicators at the point of termination of the fatigue protocol over the rest period following it. More specifically, the first five minutes of the recovery period were monitored since it is known that recovery from muscle fatigue, using maximum muscle strength as an indicator, follows an exponential curve with the first few minutes showing the greatest recovery (e.g., Sahlin & Ren, 1989), after which it was assumed that any changes in the recovery process would be minimal.

4.3.2 Type of Muscle Exertion

The second independent variable selected in this study was the type of muscle exertion. As depicted in the research concept in Chapter 3, the impact of muscle exertion on fatigue and recovery from fatigue was compared between exertions with an isometric component and/or an isoinertial component.

External Force / Load

There is no singular accepted definition of muscle workload, and meanings can be quantified in a variety of manners, e.g., via the force exerted, the effort invested in the execution of a task, or physiological responses such as heart rate or oxygen consumption (Kapitaniak, 2001a). For the purpose of this study, muscle workload was defined as the "dynamic constant external resistance"; in other words, the fixed (aka constant) external load manipulated by the muscle.

The average muscle force generated over a period of time is also known as the "mean muscle load" (Seghers & Spaepen, 2004). The mean muscle load selected in the current study for all conditions was 25% of each participant's maximum voluntary exertion (MVE), obtained either via isometric or dynamic exertions. The reason for relativising the load moved or held, was to reduce the inter-individual variability between participants with very different strength profiles. Additionally, Rohmert (1960) indicated that endurance time was independent of maximum strength and muscle group activated if the load was set at a percentage of maximum strength.

According to Iridiastadi and Nussbaum (2006), many occupations require low force production levels of around 12-28% MVE. Force levels were set at 25% of isometric MVE for the experimental conditions with an isoinertial component (i.e., “Pure Static” and “Varying Length”). This decision was informed by several studies that used mean loads ranging between 20% and 25% of MVE to successfully induce fatigue over durations lasting between 1 minute and 16 minutes (King, 2015; Nel, 2015; Patterson, 2016; Cantor, 2017; Oliver, 2017; King, 2018), while Rashedi and Nussbaum (2016) ran 1-hour long protocols using a mean muscle load of 25%MVE. This load of 25% of maximum voluntary isometric exertion (MVIE) was exerted continuously during the “Pure Static” condition; in other words, with the joint under investigation in a static position, as well as during the “Varying Length” condition, during which the joint would move consistently through a set range of motion but using a constant external load.

For the “Varying Force” condition, the forces produced by the muscle were set to vary between a minimum of 15%MVE and a maximum of 35%MVE, as it was assumed that the muscle load would also average at 25%MVE. Nel (2015) reported consistent results when minimum and maximum force levels were varied between 15% and 35%MVE in a protocol similar to the current one. Sjøgaard *et al.* (1986) found that the biceps femoris muscle was still supplied with adequate blood flow at low-level static muscle actions of 5%, whereas Rashedi and Nussbaum (2016) used minimum exertion levels of 15%MVE for index finger abduction. Furthermore, Kroemer (1970) and Kapitaniak (2001a;b) referred to static efforts of less than 20%MVE, as ‘subcritical static efforts’, and Rohmert (1960) highlighted that isometric loads of up to 15%MVE could be maintained for extended periods of time. It is important to note that, for the purpose of the current experiments, the force level variations were continuous rather than intermittent in nature, meaning there were constant force exertions of various levels and no periods of zero force exertion. The reason for this was to avoid a recovery effect of the stops (or rest periods) between efforts which would introduce an unwanted intermittent component to the protocol.

Muscle Length

Muscle length was manipulated by adjusting the joint angle. Muscle length affects force production due to the length-tension relationship, which states that a muscle's best force-producing capacity is at its resting length (Kroemer, 1999). For the biceps brachii muscle, this resting length has been reported to range between 70 degrees (Kang *et al.*, 2013) and 107.49º (Chang *et al.*, 1999) of elbow flexion. In the current study, a 90º elbow flexion angle was used as a baseline angle, as it roughly falls in the middle of this range. This joint angle has also been used in experimental protocols by Griffin (1987) and Naeije and Zorn (1982). For the conditions with an isometric component, the muscle length was thus set at this 90º elbow flexion angle.

In contrast, for the experimental condition with the dynamic (auxometric) component, the range of motion was set at 40º on either side of the 90º elbow flexion angle. The elbow would thus move from 50º to 130º elbow flexion, therefore totalling an 80º range of motion, and ensuring the same average muscle length as during the isometric exertions. These upper and lower limits of the range of movement were similar to those used in the Griffin (1987) study. Setting an upper limit for the range of motion had a practical reason, as the interference of soft tissues (i.e., the bulk of the muscle belly, particularly in well-trained males) and a poor length-tension relationship of biceps brachii when flexing the elbow beyond 130º resulted in research participants struggling to achieve the full range of motion. A more practical reason was that the isokinetic dynamometer used for the experiments prevented the eccentric mode from being initiated if the required endpoints were not reached during the dynamic exertions. Throughout the isometric and auxometric exertions, the forearm remained fully supinated, as pronation results in reduced activation of the biceps brachii (Oatis, 2004).

4.4 DEPENDENT VARIABLES

Given the complex mechanisms underlying fatigue, it is only possible to determine localised muscle fatigue through a variety of indicators (Al-Mulla *et al.*, 2011). From the numerous indicators mentioned in the literature, the following variables were selected, as they are believed to be the most reliable fatigue measures, and the relevant technologies to measure these were available to the researcher.

4.4.1 Endurance Time

In this study, endurance time was defined as the duration of the fatigue protocol from the start of the first overt movement cycle to the completion of the last cycle before termination of the protocol. Protocol durations during fatigue studies are either determined by a set time, or by the point of volitional exhaustion (Yassierli & Nussbaum, 2009; Dickerson *et al.*, 2015). Given the anticipated sizeable inter-individual variability due to various physical capacities, as well as between different experimental conditions, setting a fixed protocol duration was not considered appropriate. Instead, the protocol duration was set to the point at which each participant reached a level of near-exhaustion. Participants were asked to focus on the amount of effort invested by the working muscles and to rate this localised effort at 30-second intervals. Endurance time was recorded up to the point when participants rated the muscular effort of the involved biceps brachii as "very, very hard"; more specifically, when they reported a local Rating of Perceived Exertion (RPE) for the elbow flexor muscles of 9 on the Borg CR10 scale. A local RPE rating of 9 (out of a maximum of 10) was selected as it assured fatigue, yet avoided complete task failure, which accompanies exhaustion.

Linked to the perceived exertion and associated endurance time was the anticipation that muscle fatigue would induce changes in the number of exertion cycles that could be performed during the experimental conditions. Although it could be assumed that a set takt time per exertion cycle meant that the number of repetitions would correlate with the endurance time, changes in motor control and coordination could induce alterations in the timing of the exertions (Cortes *et al.*, 2014; Gates & Dingwell, 2011). An exertion cycle under the "Varying Length" condition was defined as the movement time through a full range of motion of elbow flexion through extension and back to maximum flexion, while under the "Varying Force" condition, cycle time was defined as the time taken from the minimum to the maximum and back to the minimum force exertion. No deliberate changes joint angle or force could be defined for the "Pure Static" condition.

4.4.2 Peak Torque, Work and Power

Arguably the most used indicator of localised muscle fatigue is a reduction in maximum force-producing capacity (Bigland-Ritchie & Woods, 1984; Gandevia, 2001; Dickerson *et al.*, 2015). It is common practice in fatigue research to interrupt the fatiguing exercise with brief maximal exertions to determine the decline in the maximal force capacity and hence quantify fatigue (Enoka & Duchateau, 2008). However, in the present study, maximum voluntary exertions were limited to before the submaximal fatigue protocol and after termination of the fatigue protocol. The reason for this was to avoid any influence that any maximum muscle actions would have on fatigue development during the sub-maximal protocol. Peak torque values were used to calculate the Fatigue Index.

In the current experiment, participants' maximum isometric and isokinetic torques produced during elbow flexion were recorded for the static and dynamic conditions, respectively, and by extension, work and power for the "Varying Length" condition. Since elbow flexion is affected by three main flexor muscles, namely biceps brachii, brachialis and brachioradialis (Tortora & Derrickson, 2009), the torque generated during elbow flexion is the combined outcome of all three elbow flexor muscles' strength. Any force decrements that may occur would, therefore, be the result of fatigue accumulation in all three muscles. The assumption was made that any reductions in force due to fatigue development would occur in similar proportions in all three muscles. Furthermore, it must be pointed out that muscle exertions are a finely tuned interaction between the agonists and antagonists to move a joint, but also to provide stability to the joint. For this experiment, the involvement of the antagonistic muscles (triceps brachii and anconeus) was assumed to be minimal, given the relatively simple uniaxial joint design of the elbow and were therefore not included as muscles of interest in this experiment.

Peak torque obtained prior to the fatigue protocol was also used to determine the external load to be manipulated during the fatigue protocol. For the experimental conditions with the isometric component, three maximum voluntary isometric exertions were each maintained for 5 seconds. Not only was the largest of the three torques produced selected as the maximum voluntary isometric exertion and used to calculate the sub-maximal load (25%MVE), but peak torques at the start and the end of the fatigue protocol, as well as during the recovery period, were used for the fatigue analyses. A similar approach was utilised for obtaining maximum dynamic strength, both concentrically and eccentrically. Three repetitions of alternating concentric and eccentric exertions were performed at a movement velocity of $60^{o}.s^{-1}$ and through an

80º range of motion (50º – 130º elbow flexion). Of the three repetitions of maximum voluntary exertions for each auxometric exertion type, the highest concentric and the highest eccentric torque produced by each participant were considered to be representative of their maximum voluntary dynamic strength.

Furthermore, since power and work are related to maximal strength (Hall, 2007), these two measures were also included for analysis, but only for the condition with overt movement (i.e., the “Varying Length” condition), since both variables are reliant on a displacement value which would not exist under the “Pure Static” and “Varying Force” conditions.

4.4.3 Time to Peak Torque

Time to peak torque has been proposed to be a reliable fatigue indicator from which inferences can be drawn for muscle fatigue development since it gives an indication of the rate with which a muscle can generate a force (D’Emanuele *et al.*, 2021). Neuromuscular fatigue results in physiological changes, influencing the muscle’s efficiency of force development, and thus the time to peak torque (Maffiuletti *et al.*, 2016). Time to peak torque was assessed during maximal exertions performed prior to and after the fatiguing exercise protocol.

4.4.4 Muscle Activity

The physiological processes of muscle activity, as quantified by electromyography (EMG), respond to different task demands. Yousif *et al.* (2019) pointed out that surface EMG (sEMG) could detect fatigue during static and dynamic muscle actions. Muscle activity of the biceps brachii was obtained through sEMG, since the biceps brachii is the only elbow flexor muscle accessible to surface EMG recordings. sEMG of biceps brachii was recorded continuously throughout the full duration of the sub-maximal fatigue protocol, i.e., from the first exertion cycle to termination of the fatiguing exercise. Furthermore, maximum voluntary exertions were obtained before the fatigue protocol, as muscle activity during maximum exertions is required for subsequent EMG normalisation (Albertus-Kajee *et al.*, 2011). Of the three traditional neuromuscular indicators of muscle fatigue, namely the EMG signal amplitude (RMS), the power spectrum, and muscle fibre conduction velocity (Arendt-Nielsen & Mills, 1988; Yassierli & Nussbaum, 2009), the RMS and median power frequency were selected to monitor the development of fatigue over time, as well as to quantify the magnitude of fatigue for comparison between the experimental conditions.

4.4.5 Performance Accuracy and Variability

It is well-known that alterations in neuromuscular coordination occur with fatigue, some of which include changes in the timing and duration of muscle activation, the pattern of muscle activity, increases in isometric force fluctuations, postural tremors and altered dynamics of limb motion (Vafadar *et al.*, 2012; Cowley *et al.*, 2014). Movement variability is defined as variations in the consistency and accuracy of repeated motor performances (Stergiou & Decker, 2011; Preatoni *et al.*, 2013); in other words, the ability to appropriately and consistently execute the required movement dynamics (Cortes *et al.*, 2014). Since not all experimental conditions involved movement per se, the term “movement” is replaced by “performance”. This study assessed temporal and spatial aspects of performance accuracy and variability through the force produced, joint angles reached or maintained, and movement velocities and accelerations attained during the fatigue protocol. Since the movement involved in the experimental protocol merely involved flexion and extension of the ulna-humeral joint, a more complex motion analysis (e.g., via 3D array analysis) was not considered necessary. Furthermore, the controlled movement protocol and environment made the elbow movements less prone to interference.

Cycle Time

Changes in neuromuscular coordination could influence the performance accuracy of individual cycles as well as the performance consistency between cycles (Côté *et al.*, 2002; Gates & Dingwell, 2008). In addition to analysing overall endurance time and number of cycles recorded, as detailed under Section 4.4.1 , exertion times for individual cycles under the “Varying Length” and “Varying Force” conditions were calculated and analysed for changes in their accuracy (i.e., being able to maintain a set takt) and cycle-to-cycle consistency over time.

Force

Force fluctuations (or steadiness of force production) have been used to evaluate the effects of fatigue and, more specifically, as a measure of task performance (Enoka & Stuart, 1992; Missenard *et al.*, 2009). This study measured how consistently participants could produce and maintain a force at 25% of their maximum voluntary isometric exertion for the experimental conditions with an isoinertial component ("Pure Static" and "Varying Length" conditions). For the experimental condition with the auxoinertial component ("Varying Force"), the accuracy and consistency of achieving the required maximum and minimum force values were assessed.

Joint angle

Muscle fatigue affects proprioception and hence the ability to move with precision (Myers *et al.*, 1999). Spatial accuracy was defined as the accuracy of achieving the required endpoint of the movement performance; in other words, the precision of repeatedly reaching or maintaining the required joint angle(s). The consistency of maintaining the required 90º joint angle was determined for the "Pure Static" and "Varying Force" conditions, as these conditions required no change in muscle length. In contrast, performance accuracy and variability for the "Varying Length" condition were determined by calculating the consistency of achieving the required minimum (elbow flexed) and maximum (elbow extended) joint angles. Similar to the force measurements, timing accuracy of reaching the required joint angles during the "Varying Length" condition was also of interest, as it was anticipated that movements would become more variable with the development of muscle fatigue as participants struggled to maintain the set movement takt.

Velocity and Acceleration

There is inconsistency in the literature regarding the effect of muscle fatigue on the consistency, or lack thereof, in movement speed and accelerations (McQuade *et al.*, 1998; Gates & Dingwell, 2008). However, given the adjustment in neuromuscular control during muscle fatigue, it is plausible that movement speeds and accelerations become more erratic as fatigue develops, which would increase the variability in movement velocities and accelerations. Since the "Pure Static" and the "Varying Force" conditions did not involve movement, variations in movement velocities and accelerations were only assessed for the "Varying Length" condition.

4.5 CONTROLLED VARIABLES

4.5.1 Joint Selection

A single joint movement was the chosen approach for this study, as this is considered a suitable manner to investigate the underlying mechanism affecting performance (Pageaux & Lepers, 2016) and was therefore also considered suitable to test the current study's research concept. Furthermore, work at prolonged low-level isometric contractions is mostly performed with the arms (Sjøgaard *et al.*, 1988). The ulna-humeral joint is responsible for elbow flexion and extension and was selected since it is a relatively simple uniaxial articulation and involves few muscles during sagittal plane movement (Wilkie, 1950). Furthermore, being a hinge joint, it has relatively few degrees of freedom, thus ensuring that the movement performed remains in the sagittal plane (Tortora & Derrickson, 2009).

4.5.2 Movement Velocity / Repetition Rate

Movement velocity was standardised as it is known that different contractile speeds affect the muscle's force-producing capacity, as defined in the force-velocity relationship (ACSM, 2006). The movement velocity during the concentric and eccentric exertions phases of the "Varying Length" condition was set at 60º per second, which was the speed recommended in the Biodex Advantage Software (V.4X) Operation Manual (Biodex Medical Systems Inc., n.d.). Such slow test velocities are also recommended by Perrin (1993) as being more suitable during testing to minimise the effect of unfamiliarity with the isokinetic dynamometry system. The movement speed of 60º per second translated to 23 movement cycles per minute (i.e., 23 flexion and 23 extension movements), meaning that each motion (i.e., either flexion or extension) was performed in 1.3 seconds. The time to complete a full cycle was, therefore, 2.6 seconds. This timing also ensured that full cycles of flexion or extension motions were performed every minute. During the baseline measurements, the isokinetic dynamometer controlled the movement velocities. However, during the fatigue protocol, the research participants were instructed to control their movement speeds so that each movement cycle was completed in 2.6 seconds. A metronome was provided to assist participants in maintaining the required cycle time.

4.5.3 Time-of-Day

Natural variability in strength expression occurs throughout the day, since the circadian rhythm follows a typical sinusoidal curve, meaning that performance output may vary throughout the course of a day (Wyse *et al.*, 1994; Knaier *et al.*, 2019). It is for this reason that across testing days, participants were tested within a 2-hour window period so that the influence that time-of-day may have had on performance variability within individuals would be minimized.

4.6 EQUIPMENT

4.6.1 Basic Demographic and Anthropometric Data

To gain insight into the general characteristics of the participant sample tested in this study, the following basic demographic and anthropometric data were recorded.

Stature was measured using a *Holtain (Ltd)* stadiometer. Participants stood barefoot and upright with the head erect, and with the heels against the baseplate of the stadiometer. Stature was measured as the distance from the ground to the apex of the skull.

Body mass measurements were obtained using a *Mettler Toledo Scale* (*model: IND231*). Participants removed shoes, excess clothing (e.g., jackets, caps), and other items (e.g., keys, mobile phone), after which they stood in the middle of the scale, in an upright posture until the value provided by the scale had stabilised.

An anthropometer from *Lafayette Instrument Company* (*model: 01290*) was used to obtain forearm length. A measure of forearm length was required to convert the maximum torque produced (measured by the isokinetic dynamometer in Nm) to kilograms (Perrin, 1993). This value, in turn, was used to determine the subsequent external load participants would be manipulating during the experimental sessions. Since the force exerted during the experimental protocol would occur with the hand in a power grip position, forearm length was measured as the distance from the olecranon process of the elbow to the third (middle) knuckle of the hand.

4.6.2 Isokinetic Dynamometry

Dynamic strength tests are difficult to perform due to the variety of factors that must be controlled, such as movement speed, range of motion and body posture (Enoka & Duchateau, 2008). An isokinetic dynamometer is a sophisticated piece of equipment that measures torque produced by a muscle, either at a pre-determined joint angle or as the limb moves through a set range of motion at a constant velocity (Smith, 2001). The isokinetic dynamometer used for this study was manufactured by *Biodex Medical Systems Inc. (model: Biodex System 4 Pro*) and was used to measure both isometric and isokinetic maximum voluntary torque production (Figure 2). The system allows for gravity correction, and data were sampled at a frequency of 100Hz.

Figure 2: Example of the Biodex setup for testing of biceps brachii.

4.6.3 Pulley System

A pulley system, using the mechanical parts of the Enraf-Nonius EN-Tree system (Figure 3), was mounted against the wall, as close as possible to the Biodex Isokinetic Dynamometer. This pulley system was used for the sub-maximal fatigue protocol, during which the appropriate external load was set, and participants had to perform the relevant exertions against this load until fatigued. The system underwent several mechanical modifications, the first of which included adding horizontal bars to the top-most weight, to which additional weights could be added (Figure 3). This alteration allowed for fine-tuning of the external load that participants were required to hold or move since the weights of the pulley system only came in full kilogram intervals. The second adjustment included adding a load cell and an analogue scale to the cable attached to the system's handle (Figure 3). These two devices were included to measure force and are further explained below. The wiring of the system was also modified so that participants had to pull the handle in a vertical direction. The entire setup, encompassing the isokinetic dynamometer and pulley system, was arranged in a manner that the chair of the isokinetic dynamometer could quickly and easily be pushed between the Biodex and the pulley system for the different phases of the experimental protocol. Therefore, participants could remain seated in the chair of the isokinetic dynamometer for the entire duration of each condition. The chair was horizontally rotated by 45º to the pulley system, so that participants remained sitting upright and pulled the handle along the scapular plane, as they did on the isokinetic dynamometer.

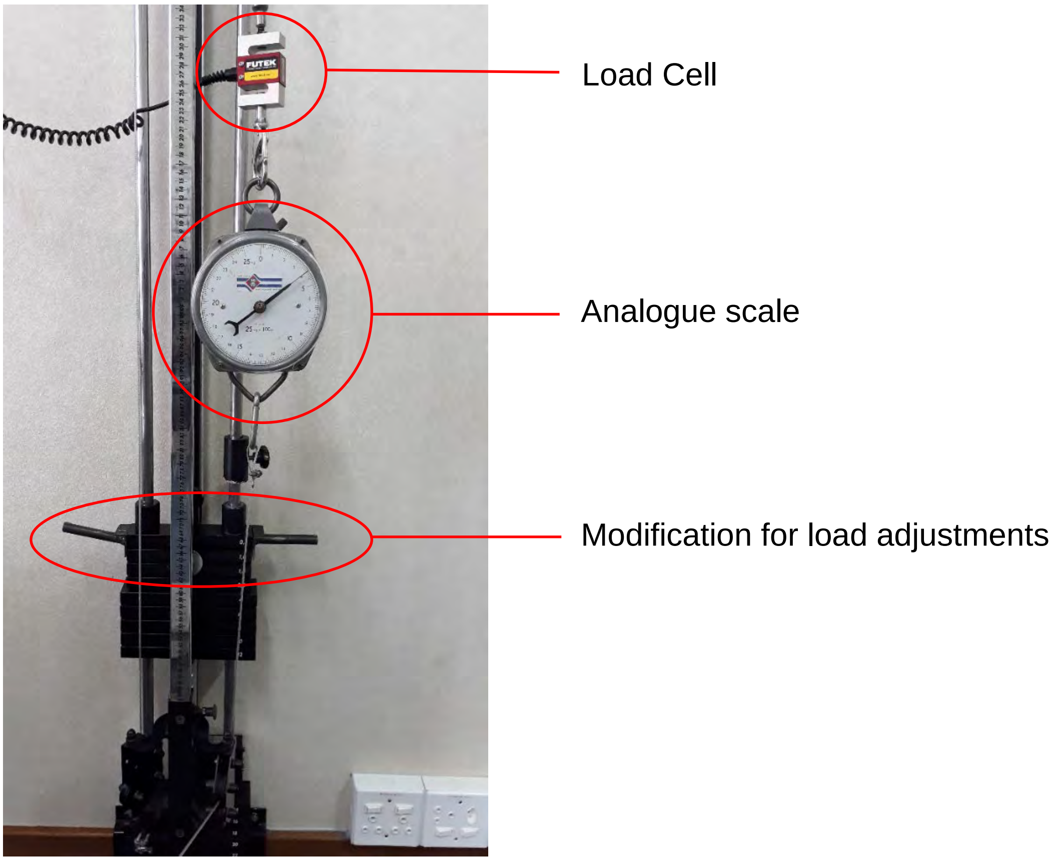


Figure 3: Image of the modified Enraf-Nonius EN-Tree system.

4.6.4 Analogue Scale and Load Cell

Two types of strain gauges were used to control and measure the force exerted by participants against the handle of the pulley system (both depicted in Figure 3). The purpose of the analogue scale from *Mass Measuring Scales (MMS)* (*25kg capacity)* was to provide visual feedback to participants so they could try to maintain/achieve the required force exertion(s), depending on the condition they performed. Before each testing session, it was ensured that the scale was set to zero.

The Futek load cell *(Model: LSB350; 1000lb capacity)* was used to assess force generated continuously throughout the fatigue protocol, with the *Sensit 2.3* software recording the force data. Baseline measurements were recorded prior to the start of the fatigue protocol, and the sampling frequency was set at 15Hz.

4.6.5 Ratings of Perceived Exertion

Since increases in perceived effort to maintain a specific performance output have been reported to indicate fatigue development (Borg, 1998), Ratings of Perceived Exertion were used to determine the level of near-voluntary exhaustion. Borg's 10-point category-ratio (CR10) scale (Table 2) was deemed suitable for this, as it can indicate central (i.e., cardio-respiratory) and local (i.e., muscular) effort. The scale starts with a value of 0, reflecting no exertion at all (using the verbal anchor of "Nothing at All"), and ends with an RPE of 10, indicating an inability to maintain the task requirements (verbal anchor: "Impossible"). Participants rated the localised effort invested by the elbow flexor muscles in 30-second intervals up to a rating of 9 (classified as "Very, very hard"), at which point the fatigue protocol was terminated.

Table 2: Ratings of Perceived Exertion CR10 Scale (Borg, 1998).

Borg CR10 Ratings of Perceived Exertion	
Rating	**Definition**
0	Nothing at all
0.5	Very, very easy
1	Very easy
2	Easy
3	Moderate
4	Somewhat hard
5	Hard
6	
7	Very hard
8	
9	Very, very hard
10	Impossible

4.6.6 Biometrics Data Logger

The *Biometrics Ltd. DataLOG* system *(Type: MWX8; Software version 8.0)* is a portable data acquisition system, which allows real-time data collection of up to 8 channels simultaneously (Biometrics Ltd DataLOG Bluetooth® & MicroSD Memory Card Operating manual, 2010). The data logger was used during this study to concurrently record electromyography, goniometry and accelerometry. The data were recorded in real-time and transmitted to the researcher's laptop via Bluetooth.

Electromyography

Even though elbow flexion is performed by three muscles, namely the biceps brachii, brachialis and brachioradialis muscles, the biceps brachii was the muscle selected for this study because it is a superficial muscle that enables surface EMG recording (Tortora & Derrickson, 2009), and this muscle shows motor unit recruitment throughout most of its contractile force range (Kukulka & Clamann, 1981), thus making it a suitable muscle for investigating the current research concept.

To record surface electromyography (sEMG), two round self-adhesive silver/silver-chloride foam electrodes with conductive hydrogel (*Kendall™/Covidien Medi-Trace® 200 Series; ref 31050522; electrode size: 10mm*) were placed next to one another on the skin overlying biceps brachii and in parallel to the muscle fibres, ensuring an inter-electrode distance of 36mm. The muscle activity was picked up via a pre-amplifier (*sensor type no.: SX230FW)*, which incorporated a high-pass filter, followed by a low-pass filter to remove unwanted frequencies (i.e., ambient and transducer noise) (refer to *Biometrics Ltd. EMG Sensor Operating Manual, 2010)*. A reference electrode (*KILLSTAT®; Part number CA4AADB; with 4mm stud*) was placed on an inactive body part (details to be found under procedures). Sampling frequency was set at 1000Hz, as recommended by Konrad (2005).

Principal detection, as proposed by Merletti (1999), included the following attributes (taken from http://www.nexgenergo.com/ergonomics/biodataemg.html):

- Bipolar / Differential
- Bandswidth: 20Hz-460Hz
- Noise: < 5µV
- Input Impedance: > 10,000,000 MOhm
- Common Mode Rejection Ratio (CMRR) at 60Hz (dB): > 96dB (typically 110dB)
- Gains: 1000
- High-pass filter: third-order filter (18dB / octave) (Butterworth filter)
- Low-pass filter: 450Hz (eight-order elliptic filter)

Goniometry

An electrogoniometer was used to record joint angle and, therefore, also the range of motion. During the analysis phase, this information would allow for the identification of different movement phases for the dynamic ("Varying Length") condition, but also allow for calculations of variations in the joint angle (i.e., determination of performance variability). It was also anticipated that the electrogoniometer would pick up on movement tremors during the "Pure Static" and "Varying Force" conditions, even though these conditions included no purposeful movement. The electrogoniometer used during the current study was a twin-axis goniometer from *Biometrics Ltd. (sensor type: SG110,* as specified in the *Biometrics Ltd Goniometer and Torsiometer Operating Manual, 2010).* Only measurements from a single axis were recorded due to the uniaxial nature of the elbow flexion and extension motion. The transducer type was a strain gauge and had a 150° measurement range. The sampling frequency was set at 50Hz.

Accelerometry

A uni-axial accelerometer sensor *(Crossbow Technology Inc.; model: CXL-LP; 2g span; 500mV.g^{-1} sensitivity)* provided information on movement variability by recording accelerations around the transverse axis, i.e., during elbow flexion and extension, as the handle of the pulley system was lifted and lowered. This accelerometer was compatible with the Biometrics data logger system and was thus able to record accelerations concurrently with the EMG and goniometry data. The sampling frequency for the accelerometer was 1000Hz.

4.7 PARTICIPATION CRITERIA AND PARTICIPANT RECRUITMENT

Participants of both sexes had to be between the ages of 18-25 years. A population group within this age range tends to be at their physical peak and is therefore at less risk of injury during the maximum force exertions (McArdle *et al.*, 1996; Kwan, 2013). Furthermore, the lower age range was limited due to ethical considerations when dealing with minors, while restricting the upper age range was considered necessary due to age-related changes that occur in strength expression (Hunter *et al.*, 2005) and cardiovascular responses to exercise (Smolander *et al.*, 1998).

Furthermore, participants had to be in good physical condition, without a history of recent (past 12 months) musculoskeletal injury or disorders of the upper extremities, neck, or upper back. Absence of injury was emphasised since the maximal exertions required during this protocol could exacerbate any injury participants may have had, and that had not been entirely resolved. Recent injury may also alter muscle action lines and tissue tolerances such that participants could be unknowingly at risk of harm (Marras, 2012; Vila-Chã *et al.*, 2012). Regular participation in physical activity was considered important since the protocol required maximal exertions, and individuals who exercise regularly are generally comfortable with giving maximal effort during physical activities. Regular weight training of 2-3 times a week was suggested by Hongu *et al.* (2015), while in studies by Iridiastadi and Nussbaum (2006) and Rashedi and Nussbaum (2016), participants engaged in moderate-intensity activities between 1-3 times weekly.

Given the above criteria, all participants recruited came from the Rhodes University student cohort, with most students belonging to the Human Kinetics and Ergonomics (HKE) Department, since they were most readily available and stood to gain educational benefits from participating in this study. However, non-students who met the inclusion criteria could also participate. The research participants were recruited by sending emails out via the Human Kinetics and Ergonomics Department's student distribution lists, as well as through brief verbal addresses in class, and word-of-mouth. Interested individuals who contacted the researcher were emailed the letter of information (Appendix A.1), and a date and time were set to meet for an individual information and habituation session.

4.8 ETHICAL APPROVAL

The experimental protocol was reviewed and approved by the Rhodes University Ethical Standards Committee – Human Ethics Sub-committee (tracking number 0150; Appendix A.3). Due to the use of student participants, gatekeeper permission was obtained from the Rhodes University Registrar (Appendix A.4).

4.9 EXPERIMENTAL SETUP AND PROCEDURES

Once ethical approval had been obtained and potential participants had been recruited, they were invited to attend an initial information, familiarisation, and habituation session. Testing of the experimental conditions occurred during three subsequent sessions. Figure 4 provides a schematic overview of the procedures.

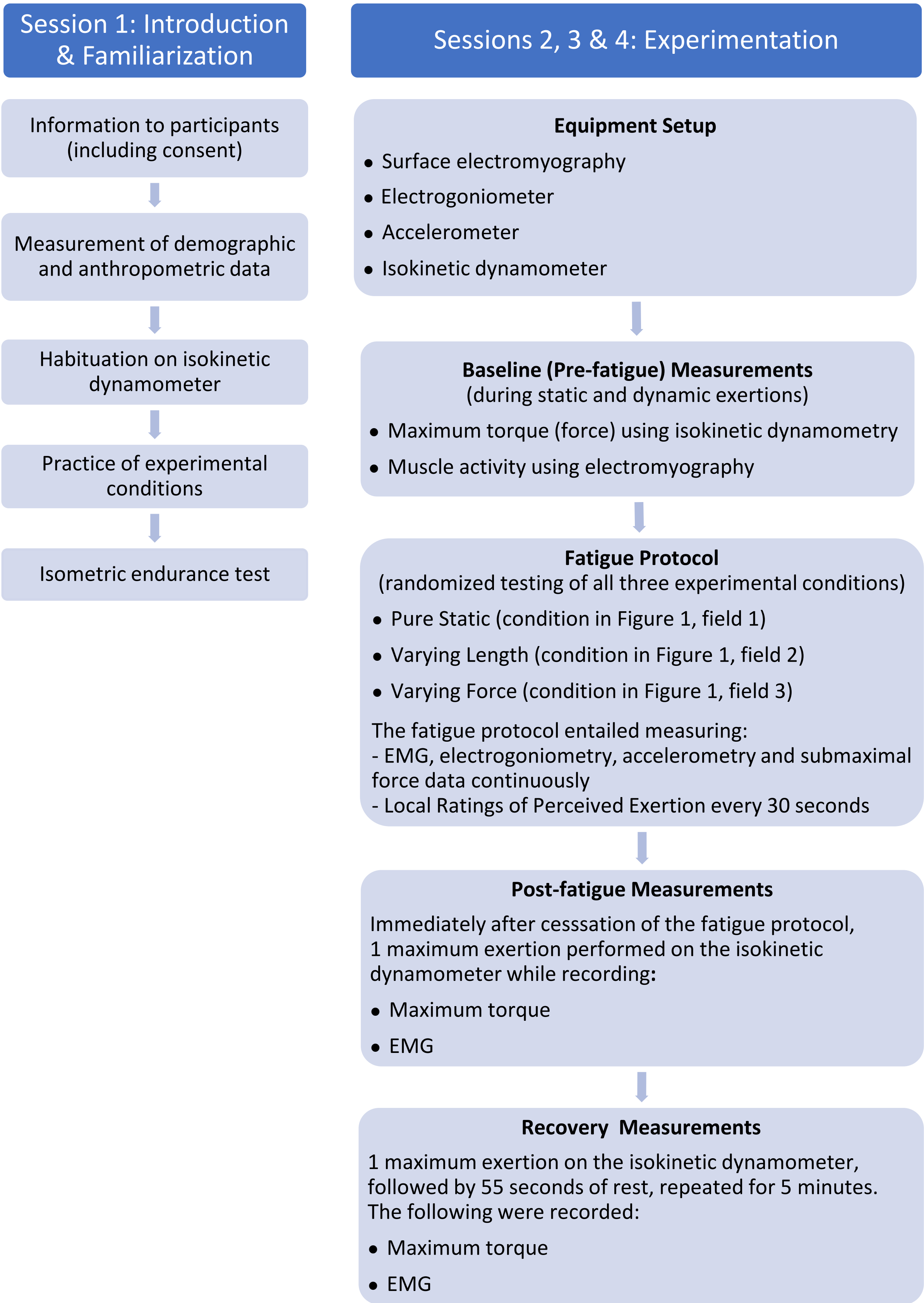


Figure 4: Overview of experimental procedures.

4.9.1 Session 1: Information and Familiarization Session

This session lasted about one hour and consisted of five parts:

i) Information and Consent

Participants were provided with a verbal explanation of the purpose of the study (as detailed in the Letter of Information – Appendix A.1), details of the experimental procedures, as well as their rights and obligations. A demonstration of the equipment was also provided. Any questions participants might have had were answered to their satisfaction, and, once they had agreed to participate, the informed consent form was signed (Appendix A.2). All participants received a copy of the letter of information and informed consent form to take home.

ii) Demographic and Anthropometric Data

Information on age, sex, and hand dominance was obtained verbally, while stature, body mass, and forearm length were recorded using the appropriate equipment, as explained in Section 4.6 ("Equipment" section) of this thesis.

iii) Habituation

Proper setup of the Biodex Isokinetic Dynamometer, particularly the chair in which the participants were seated for the testing protocol, not only ensured participants' comfort but also optimal force-producing capacity. According to the manufacturer's instructions, the dynamometer was rotated to horizontally by 30^{o}, with a tilt angle of 0^{o}. This setup ensured that elbow flexion and extension occurred along the scapular plane. Participants were seated, with the trunk upright in the chair of the dynamometer and secured via shoulder and thigh straps. Then, the seat and handle of the Biodex were adjusted so that the elbow joint was aligned with the axis of the dynamometer, and the handle fit comfortably in the hand. Contrary to the manufacturer's instructions, the shoulder was aligned with the trunk as much as possible. A setup with a flexed shoulder, as per the manufacturer's instructions, would have incurred the risk that, during extreme elbow flexion, the effects of gravity on the forearm would negatively affect biceps brachii activation. Keeping the shoulder as neutral as possible (within the constraints of the system's adjustability) ensured that biceps brachii would always be activated during the dynamic exertions, either concentrically or eccentrically. This postural alignment corresponded to that adopted during the subsequent fatigue protocol. Setup parameters for the isokinetic dynamometry system were recorded for each participant during the initial familiarisation session and standardised across all subsequent testing sessions.

Participants were habituated on the isokinetic dynamometer once the setup had been completed and its parameters recorded. The familiarisation of the task requirements entailed performing the maximum voluntary isometric, concentric and eccentric exertions of the forearm muscles, which they would need to perform during the experimental sessions. Given the unfamiliarity of the activity, particularly the concentric and eccentric exertions when performed on an isokinetic dynamometer, emphasis was placed on ensuring that detailed instructions were provided to participants and that the execution of these movements was correct.

iv) Practice of Experimental Conditions

Following the habituation on the isokinetic dynamometer, participants practised the various exertions they would be required to perform during the fatigue protocol on the pulley system. The sub-maximal loads to be manipulated during this practice session were calculated from the maximum isometric, concentric and eccentric torques obtained during the preceding maximum exertions on the isokinetic dynamometer (calculation explained under Section 4.9.2). The emphasis during this practice component of the session was placed on ensuring that participants achieved the correct ranges of motion, forces, and timing of movement cycles. It was important that participants performed smooth moves, as ballistic actions are known to produce different motor unit discharge rates (Duchateau & Baudry, 2014). It was also stressed that participants should maintain a neutral wrist angle to avoid the wrist flexor muscles from becoming unnecessarily activated, particularly during the fatigue protocol.

v) Endurance Test

At the end of the information and familiarization session, participants performed an isometric endurance test, during which they exerted a force equivalent to 25% of their isometric MVE at an elbow of 90^{o} against the pulley system. The purpose of this isometric endurance test was to expose participants to the RPE scale, as well as to obtain an endurance time estimate to use for setting the time limits required by the Futek load cell software used during the fatigue protocols in the subsequent experimental sessions.

4.9.2 Sessions 2-4: Experimental Testing Sessions

Each of the three experimental conditions ("Pure Static", "Varying Force", and "Varying Length") was tested on separate days to negate the effects of possible residual or cumulative fatigue from influencing the results. Each session lasted approximately 45 minutes. Although the order of the experimental conditions was aimed to be fully randomised, recording challenges resulted in several testing sessions having to be repeated, and therefore the final allocation of experimental conditions was not completely random. Recording challenges involved rolling blackouts, which are experienced regularly in South Africa due to an inconsistent electricity supply by the national energy provider. The absence of electricity resulted in having to reschedule several testing sessions, which influenced the permutation. The individual testing sessions were spaced at least two days apart to minimise the effect that any possible delayed onset of muscle soreness (DOMS) from the previous testing session may have had on muscle activity during subsequent sessions. This decision corresponded with the protocol of a study by Yassierli and Nussbaum (2009), who also ensured a minimum of two days' rest between testing sessions. Participants who still experienced some DOMS on the day of a subsequent testing session were rescheduled until such a day that the DOMS had subsided entirely.

At the start of each experimental session, participants were informed of the condition they would be performing that day and were reminded of the relevant procedures. Each experimental session was made up of the following four parts, using the protocol suggested by Perrin (1993).

i) Equipment Setup

For surface electromyography (sEMG) measurements, the skin overlying the biceps brachii muscle on the dominant arm, where the surface electrodes would be placed, was thoroughly cleaned with alcohol, and, where necessary, hair removed by shaving the area (Konrad, 2005; Gamet & Fokapu, 2008). This procedure would ensure proper adhesion of the EMG foam electrodes and clear signal detection. The electrode placement site was standardised by having all participants push isometrically against a static resistance with the elbow flexed at a right angle. This exertion highlighted the muscle belly at the 90^{o} position, and the electrodes were placed in line with the muscle fibres in the centre of the muscle belly (Kamen & Gabriel, 2010; SENIAM.org). This process prevented the placement of the electrodes too close to the innervation zone or to other muscles, which may have resulted in crosstalk and, therefore, negatively affect signal recording quality (González-Izal *et al.*, 2012). Furthermore, given that the muscle belly moves under the skin during dynamic exertions, it was important to place the electrodes in a central position over the muscle belly (Konrad, 2005). This placement was considered the central position, as the movement performed and the resultant migration of the muscle under the skin would occur at 40º on either side of the 90º elbow flexion angle. The reference EMG electrode could not be placed on the wrist of the working arm, as recommended by SENIAM, or anywhere else on the dominant arm, as it would interfere with the other sensors used in this study. It was therefore placed on the olecranon process of the elbow of the non-dominant arm, as this was a region without electrophysiological activity linked to the ground (Gamet & Fokapu, 2008). All electrodes and cables were secured with medical tape.

EMG setup was followed by the attachment of an electrogoniometer to the lateral aspect of the elbow. With the forearm supinated and the elbow extended to 180º, one end of the goniometer was aligned with the shaft of the humerus, while the other end was aligned with the shaft of the radius – both equal distances from the lateral condyle of the elbow (Figure 5). The goniometer was set to zero with the elbow flexed at a 90º angle.

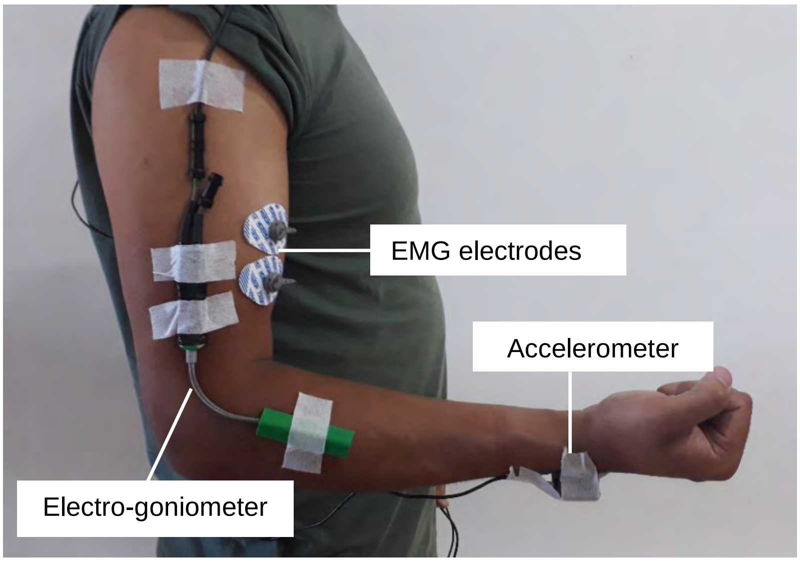


Figure 5: Photograph showing the setup of the sEMG, electrogoniometry, and accelerometry sensors attached to the arm (EMG ground electrode not visible).

The accelerometer was secured on the posterior aspect of the wrist and at the level of the radiocarpal joint (Figure 5). The shape of the pulley system's handle was such that it would have interfered with the accelerometer had the accelerometer been placed on the posterior aspect of the hand itself. Other placement options on the hand and wrist were considered equally awkward and uncomfortable.

Once all sensors had been attached to the participant, their cables were connected to the Biometrics measurement unit, which was secured around the waist via a belt.

ii) Baseline Measurements

Baseline measurements involved obtaining torque and EMG recordings during maximum voluntary exertions. Before the commencement of the baseline measurements, participants were seated upright and securely strapped to the Biodex chair to ensure a stable base of support and to avoid compensatory movements. Since limb positioning can influence a muscle's strength-producing capability (Kroemer, 1999), standardising the posture across conditions and participants was crucial. In this study, the upper arm was aligned as close as possible to the trunk, while the olecranon process rested on a padded attachment, and the elbow joint was flexed to 90^o. The exact shoulder flexion angle varied slightly between individuals and depended on each participant's anthropometry and comfort while maintaining an upright trunk. However, this posture remained consistent across experimental sessions with each participant. This postural alignment was similar to the one used by Seghers and Spaepen (2004), whose participants' upper arms were positioned vertically next to the trunk, while the forearms were positioned horizontally and supinated during maximum isometric exertions. The handle of the Biodex Isokinetic Dynamometer attachment was comfortably positioned in the dominant hand. Participants then engaged in a light warm-up exercise by performing several repetitions of elbow flexion and extension against the handle while the dynamometer was in a neutral mode (refer to Figure 2).

Once participants had warmed up and were willing to proceed, gravity correction was performed on the dynamometer. The required function of the isokinetic dynamometer was then initiated (i.e., either isometric or isokinetic), and participants were required to exert as much force as possible against the handle of the dynamometer. Since MVEs need to reflect the muscle action tested due to different muscle recruitment (Albertus-Kajee *et al.*, 2010; 2011), MVE protocols differed between conditions – either using maximal voluntary isometric exertions (MVIE) or maximal voluntary dynamic exertions (MVDE). Furthermore, since it takes a while for the force produced in the sarcomeres to be transferred to the bones and thus reach a final value (Wilkie, 1950), Chaffin (1973) recommended that maximum voluntary isometric force should be obtained from strength trials lasting between 4-6 seconds. This duration would allow for at least a 3-second long steady reading to be obtained. Iridiastadi and Nussbaum (2006) also recommended four 5-second-long MVEs. In the current study, sessions that had been allocated a condition with an isometric component (i.e., the “Pure Static” or “Varying Force” conditions), three repetitions of maximum voluntary isometric exertions were performed at an elbow flexion angle of 90^o and for five seconds each, with ten seconds rest between exertions. Preliminary work revealed that the highest MVIE was achieved within the first three exertions; hence, only three MVIEs were performed. The highest peak torque obtained by the three exertions was recorded as the maximum voluntary exertion (Søgaard *et al.*, 1996) and used for the subsequent sub-maximal load calculations.

For the dynamic (“Varying Length”) condition, maximum isokinetic torques were recorded. Albertus-Kajee *et al.* (2011) found that using maximal dynamic movement, in their case a running activity, was as repeatable as the maximum voluntary isometric exertion method. They considered this approach more appropriate, as it represented the same muscle movement. In the current experiment, the MVDE method involved three repetitions of alternating concentric and eccentric exertions through a range of motion of 80^o, and at a speed of $60^o.s^{-1}$. Throughout all maximum exertions (both isometric and isokinetic) muscle activity was recorded, and verbal motivation was provided.

iii) Fatigue Protocol

The fatigue protocol followed the baseline measurements. Several researchers referred to the fact that a rest period of between two and ten minutes should intersperse the MVIE and the exercise protocols (Chaffin, 1975; Potvin & Bent, 1997; Kumar *et al.*, 2002; Yassierli & Nussbaum, 2009). The rest period during the current study occurred as participants were moved from the isokinetic dynamometer to the EN-Tree pulley system, while remaining strapped to the Biodex chair. The seat was rotated so that the pulley system was positioned at a 45^o angle to the participants' scapular plane. In other words, the postural setup remained similar to that of the preceding maximum exertions and the rest period provided by the changeover aligned with the recommendations.

The torque obtained from the maximum voluntary exertions on the isokinetic dynamometer was used to calculate the sub-maximal load participants had to manipulate during the fatigue protocol. For the conditions with an isometric component ("Pure Static" and "Varying Force"), the peak isometric torques obtained during the baseline measurements were used. For the condition with the isoinertial component ("Varying Length"), the mean of the concentric and eccentric maximum voluntary torques was calculated to obtain a maximum voluntary dynamic torque. This was done since concentric and eccentric exertions have different maximum force-generating capacities (Hortobágyi & Katch, 1990); however, it was impossible to actively change the external load during the fatigue protocol. The external load handled during the fatigue conditions was calculated as follows:

1) Converting the maximum torque to maximum force:

$$Maximum\ Force\ (N) = \frac{peak\ torque\ (Nm)}{forearm\ length\ (m)}$$

2) Converting the maximum force to kilograms by dividing it by the gravity constant of $9.81 m.s^{-2}$.

3) Calculating the sub-maximal loads: either 25% (for the "Pure Static", and "Varying Length" conditions), or 15% and 35% (for the "Varying Force" condition) of the maximum force.

The calculated sub-maximal weights were adjusted on the pulley system by adding or removing weights and were checked using the analogue scale.

The fatigue protocol itself involved manipulating the required sub-maximal load(s) depending on the experimental condition performed. It is important to note that the force applications under all experimental conditions were continuous rather than intermittent in nature. The reason for this was to avoid a recovery effect that any stops that could have occurred between efforts and would therefore have introduced an unwanted intermitted component to the protocol.

Experimental Condition "Pure Static" (Figure 6): For this experimental condition, participants had to exert a consistent, static pulling force against the handle of the pulley system while maintaining the elbow angle at 90º flexion. Participants were instructed to keep the required joint angle and force as stable as possible. Visual feedback from the analogue scale allowed participants to consciously control the force exerted, and verbal feedback from the researcher was provided if the force dropped below the target level. A marker placed against the pulley system showed participants what level the weights should be held at when the elbow was flexed at 90º.

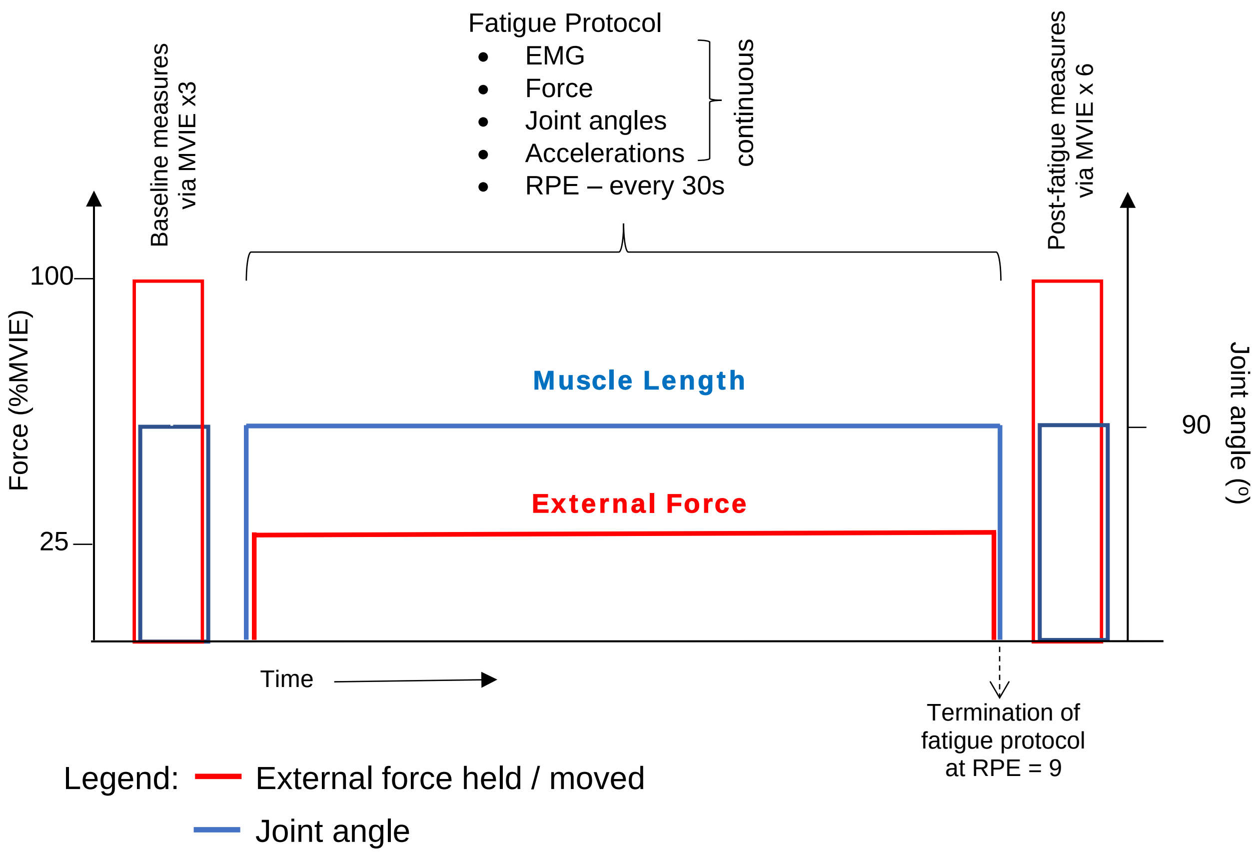


Figure 6: Graphical presentation of the experimental procedures for the "Pure Static" condition.

Experimental Condition "Varying Length" (Figure 7): For this condition, participants moved the external resistance (25% of maximum voluntary dynamic exertion) through the required 80º range of motion by continuously flexing and extending the elbow between 50º and 130º. Visual markers placed against the pulley system provided visual feedback for the upper and lower movement limits. The speed of the movement mimicked that of the baseline measurement, i.e., $60^{o}.s^{-1}$, which meant moving from one endpoint of the range of motion to the other in approximately 1.3 seconds. A metronome dictated the takt, which was set at 46 Hz. Any deviations from the required takt were pointed out to the participant by the researcher, encouraging the participant to fall back into the required movement rhythm. It was emphasised to participants to move at a relatively consistent speed and not to stop and hold the weight at the ends of the range of motion. The reason for this was to minimise an isometric addition to the isoinertial exertions.

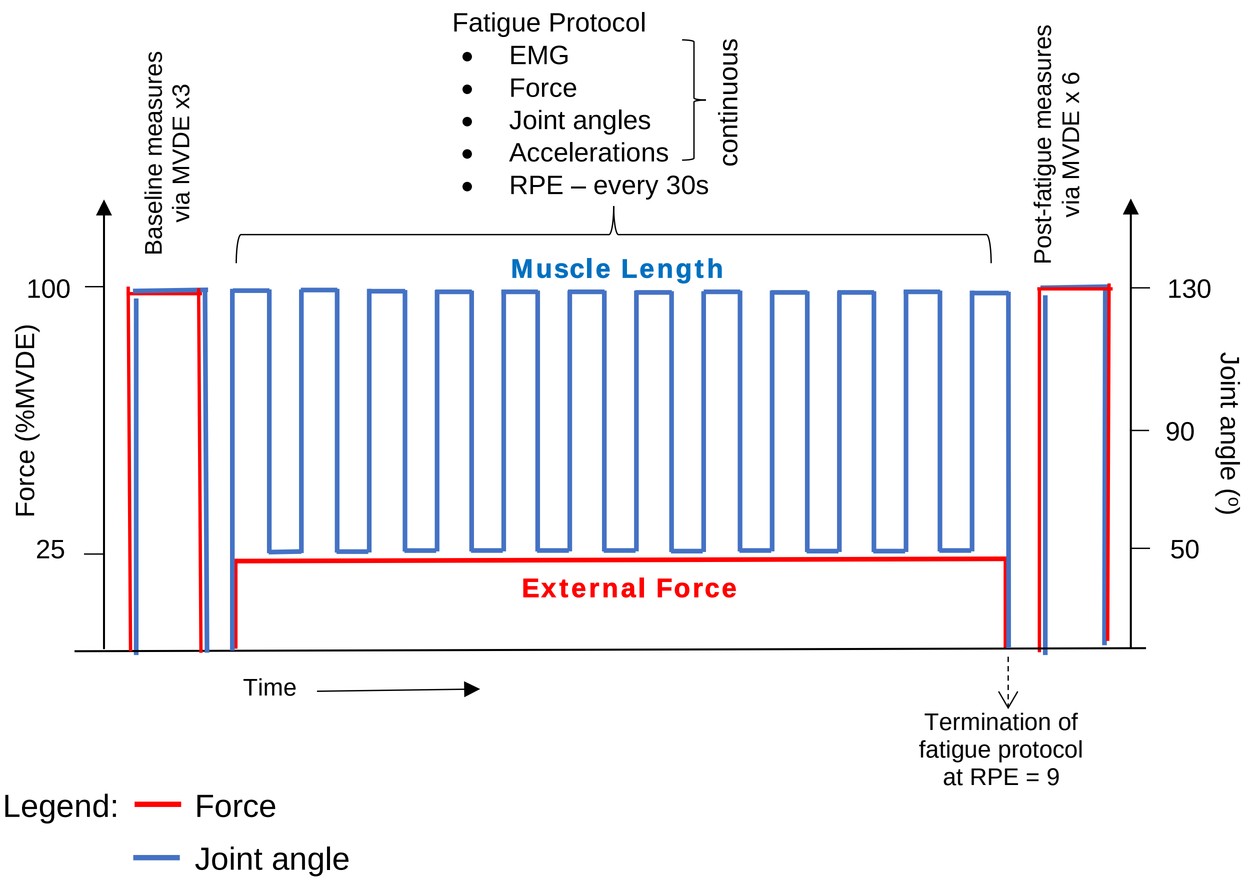


Figure 7: Graphical presentation of the experimental procedures for the "Varying Length" condition.

Experimental Condition "Varying Force" (Figure 8 - field 3): For this condition, a 'stopper' was applied to the pulley system, preventing movement of the load. The height of the stopper was set at the level where the elbow would concurrently be at a 90º flexion angle. Again, participants maintained an elbow flexion angle of 90º while pulling on the handle, but varying the force applied to the immovable resistance of the pulley system between 15% and 30% of the isometric maximum. If exerted for equal amounts of time (i.e., at the same timing of 1.3 seconds per force exertion), the mean muscle force produced was assumed to equal 25% of the maximum isometric force, thus ensuring a similar workload throughout all experimental conditions. The analogue force gauge attached to the pulley system provided visual feedback for participants to regulate their force application. The metronome again dictated the takt to which force had to vary, and smooth force application was emphasised to participants. Where necessary, participants were reminded of the required force levels.

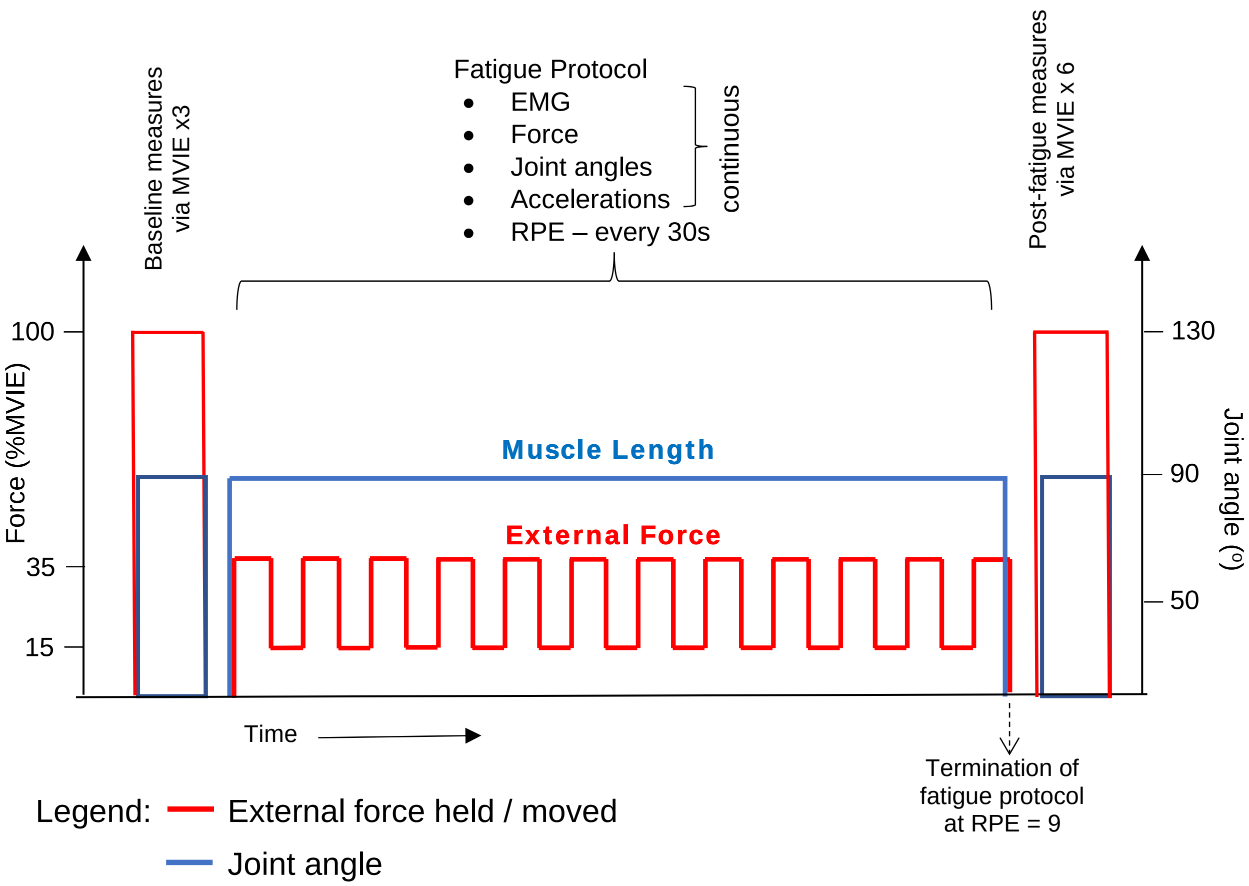


Figure 8: Graphical presentation of the experimental procedures for the "Varying Force" condition.

Throughout all three experimental conditions, the following variables were continuously recorded: force levels, muscle activity, joint angles, and movement velocities and accelerations using the load cell, EMG, electrogoniometer, and accelerometer sensors, respectively. Local RPE were obtained at 30-second intervals. Once participants reached an RPE of 9, the fatigue protocol was terminated, and the post-fatigue and recovery measurements initiated.

iv) Post-fatigue and Recovery Measurements

Immediately, upon cessation of the fatigue protocol, participants were moved back to the isokinetic dynamometer where they had to perform either one maximum isometric or one maximum isokinetic (one concentric and one eccentric) exertion, depending on what condition they had been assigned to. This exertion was followed by 55 seconds of rest and then another maximum voluntary exertion for a total duration of 5 minutes since Pasquet *et al.* (2000) found that most recovery in maximum isometric torque occurred in the first 5 minutes of the rest period. A total of six post-fatigue maximum exertions were performed, during which maximum torque and time to peak torque were recorded, as well as work and power for the “Varying Length” condition. While it is acknowledged that interjecting the recovery period with additional maximal exertions may influence the recovery process, this was considered the most feasible option to determine whether recovery differed between the three experimental conditions. Movement speed and range of motion remained the same as during the baseline measurements. During the rest periods between exertions, participants relaxed their forearms against the handle of the dynamometer to avoid activation of the biceps brachii muscle.

Once participants had completed a session, all pieces of equipment were removed, and it was ensured that participants were feeling well. Individuals were thanked for their participation, and a new date and time were set for the following session, if applicable.

For ease of reference, Figure 9 indicates the points along the experimental protocol during which the various measurements were taken to determine whether muscle fatigue had occurred as a result of the exercise protocol. The terms “pre” and “post” protocol refer to the maximal exertions just prior to and immediately upon termination of the fatiguing exercise, respectively, as well as the 5-minute recovery period. The terms “first four cycles” and “last four cycles” refer to the measurements of the sub-maximal exertions at the start and the end of the fatigue protocol.

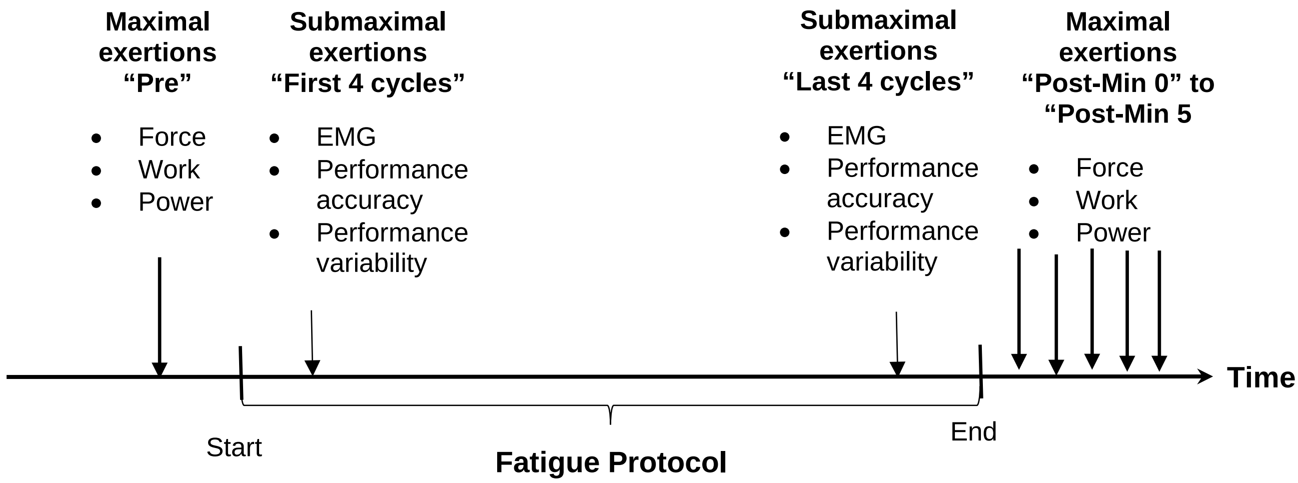


Figure 9: Schematic diagram of the points of fatigue measurement.

4.10 DATA REDUCTION AND PROCESSING

4.10.1 Data Reduction

No data reduction was necessary for the basic demographic and anthropometric data, nor for endurance time (time to RPE = 9) or number of cycles. Similarly, measures of peak torque, time to peak torque, work and power that were obtained during the maximal exertions prior to and after termination of the exercise protocol (including the recovery period) did not require any data reduction as the isokinetic dynamometer software provided those as part of the issued report. The Fatigue Index was calculated using the peak torque data. The data recorded were visually screened for any potential outliers.

However, other measures, such as EMG (which was rectified), cycle time, force, joint angles and accelerations that were recorded during the submaximal exercise protocol, needed to be reduced to manageable values since these measurement sensors had sampling frequencies ranging from 15Hz for the load cell, to 1000Hz for the EMG and accelerometer. It was therefore decided to extract data from the beginning as well as from the end of the fatigue protocol to represent the extremes of muscle fatigue development (i.e., no fatigue at the start vs the greatest fatigue at the end of the exercise protocol). Data were analysed by identifying the movement and force cycles produced during the fatigue protocol. Under the “Varying Length” condition, a cycle was defined as the movement from maximal flexion to maximal extension and back to maximal flexion again. Similarly, under the “Varying Force” condition, a cycle ranged from the point of maximal force production to minimum force and back to maximum force again. As a first step, the first three cycles from the fatigue protocol were deleted as participants needed about three cycles at the start of the protocol to settle into the required rhythm for the “Varying Length” and “Varying Force”. The following four cycles (i.e., cycles 4-7) were considered suitable for the analyses. Four cycles were chosen, as this resulted in an approximately 10.4-second time interval, which was deemed long enough to obtain sufficiently consistent and reliable data to represent the unfatigued state but short enough to avoid the impact that the onset of fatigue could have had on responses. Figure 10 depicts the start of a sample trace obtained from the Biometrics datalogger, highlighting the initial four intervals eliminated from analysis (Interval ‘A’), followed by the next four individual cycles (referred to as interval ‘B’) using for the fatigue analysis. Similarly, the last three cycles of the fatigue protocol were ignored as participants displayed a final surge in performance, i.e., a type of “end-spurt” just prior to terminating the activity and which would have changed the response characteristics. The preceding four cycles were thus extracted for analysis.

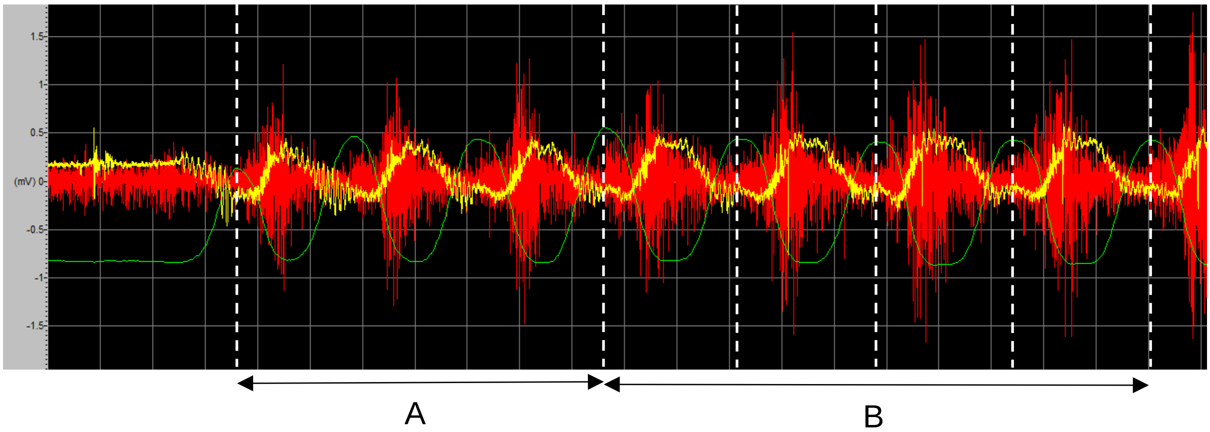


Figure 10: Sample of a measurement trace from the Biometrics Datalogger during the “Varying Length” condition at the start of the fatigue protocol.

The trace shows the joint angles in green, the EMG activity in red and the force measurements in yellow. Interval ‘A’ reflects the first three cycles that were eliminated, while interval ‘B’ shows the following four intervals (each interval indicated by the white dashed lines) used for the analysis.

While extraction of individual exertion cycles under the “Varying Length” condition utilized the goniometry data, since it allowed for clear identification and isolation of each movement cycle, for the “Varying Force” condition, it was the force data from the Sensit load cell that allowed each force cycle (varying from maximum to minimum and back to maximum force) to be identified (Figure 11). Again, the first three movement cycles were eliminated, while the following four cycles were included in the fatigue analysis. The time markers obtained from the Sensit load cell (shown on the x-axis) were used to identify the appropriate intervals in the Biometrics trace, thus identifying the corresponding EMG and movement variability data. The cycles at the end of the activity for analysis were identified by locating the point in time where the joint angle or force changed dramatically (due to letting go of the pulley’s handle). Using this point in time, the preceding three cycles were eliminated and the four cycles prior to these were extracted for analysis.

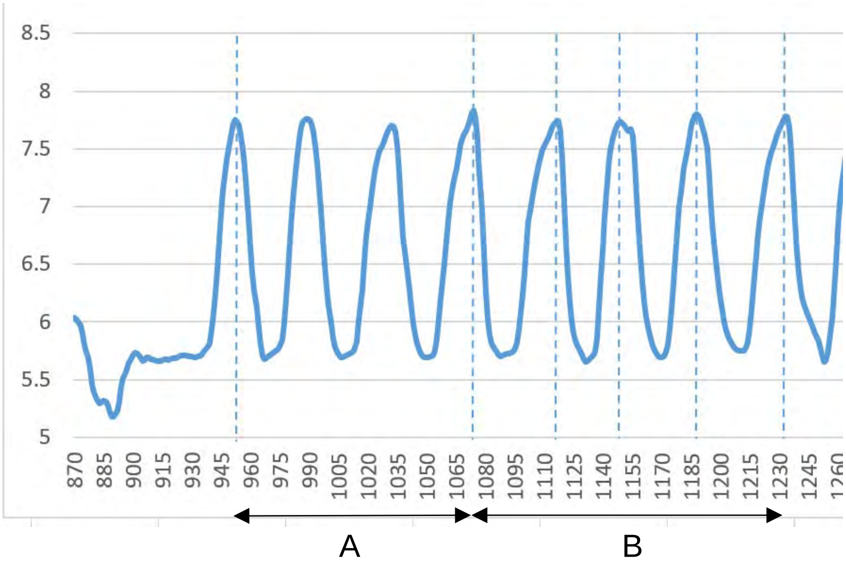


Figure 11: Sample of the Sensit (force) data, converted via Excel to a trace. It shows the first four intervals eliminated from analysis (‘A’), as well as the following three intervals used for the fatigue analysis(‘B’).

Since no force or movement cycles existed for the “Pure Static” condition, the starting point of the activity was identified at the point where the goniometry and force curves stabilised, after which the first 7.8 seconds were eliminated (the equivalent of three 2.6-second intervals, as dictated by the movement frequency of 23 cycles per minute). The following 10.4 seconds (equivalent of four movement cycles) were used for the analysis. Similarly, 7.8 seconds prior to a marked change in the force and joint angle data were deleted and the 10.4 seconds before that used for analysis. Spreadsheets in Microsoft Excel were used to calculate times, changes over time, as well as deviations and variability between cycles for the various fatigue parameters before exporting the data to the statistics software.

4.10.2 Descriptive Statistics

The StatSoft Statistica Software (version 13.4.0.14 by *TIBCO Software Inc.*) was used to calculate basic descriptive summary data (means, standards deviations and coefficients of variation) for the study sample's fatigue parameters, namely endurance time, number of cycles, peak torque, time to peak torque, the Fatigue Index, work, power, EMG amplitude (RMS), EMG median frequency (MdPF), performance accuracy and cycle-to-cycle variability for all experimental conditions and at the various time intervals. While most data are presented as absolute values, some (e.g., peak torque, work and power) were relativised to minimise the impact of individual characteristics, such as body weight or strength, on any changes or differences in fatigue responses. For performance accuracy and cycle-to-cycle variability, the variables' mean deviations and their variability (range, standard deviation, and coefficient of variance) during the first four and the last four viable movement cycles of the fatigue protocol were calculated for cycle time, forces exerted, range of motion and joint angles, as well as acceleration. Given the different characteristics of each of the muscle exertions, not all parameters were processed in the same manner. The details of how performance accuracy and cycle-to-cycle variability are presented can be found in the relevant sections of the results chapter of this thesis. Finally, associations between some variables (e.g., sex and peak torque) were assessed using the Pearson Product-Moment correlation analysis. Significant correlations were identified at $p<0.05$, with correlation coefficients with absolute magnitudes of less than 0.1 being classified as 'negligible', between 0.1 and 0.39 as 'weak', 0.4-0.69 as 'moderate', 0.7-0.89 as 'strong' and 0.9-1.00 as 'very strong' associations (Schober *et al.*, 2018).

4.10.3 Inferential Statistics

Inferential statistics were first used to establish whether the exercise protocol indeed caused muscle fatigue, and then, whether the fatigue responses differed between the three experimental conditions. Fatigue was established by considering any time-related changes in the measured variables. Peak torque, time to peak torque, work, and power data obtained during the maximum exertions prior to the fatigue protocol ("Pre") were compared to those recorded immediately afterwards ("Post-Min 0") and at the end of the recovery period ("Post-Min 5"). Similarly, the EMG, performance accuracy and cycle-to-cycle variability data (i.e., cycle time, force, joint angles, and accelerations) extracted during the first four viable cycles of the fatiguing exercise protocol were compared to those of the four cycles at the end. To identify any differences in fatigue between the muscle exertions, i.e., the "Varying Length" (auxometric-isoinertial), "Pure Static" (isometric-isoinertial) and "Varying Force" (isometric-auxoinertial) conditions, the same variables were compared but between these three exertion types. To determine the time and condition (exertion) effects on fatigue responses, a within-subject, repeated measures two-factorial ANOVA was conducted. For variables that had relativized the post-fatigue values to the pre-fatigue ones, a one-factorial analysis of variance approach was used. For this, a General Linear Model (GLM) analysis was used with a 95% confidence interval, meaning that a p-value of less than 0.05 would indicate a significant effect. Any significant effects identified were followed up with a Tukey post hoc test.

The selection of the statistical tests to determine the changes over time and between muscle exertions was based on the Central Limit Theorem. According to this theorem, samples can be treated as being normally distributed if they are large enough (Kwak & Kim, 2017; Rosner, 2000). The definition for 'large enough', however, varies, but according to Kwak & Kim (2017), samples larger than 30 can be considered to be normally distributed and, therefore, make use of parametric tests to determine condition effects. Furthermore, since all variables recorded in the current study are physiological responses, it is reasonable to assume that they should follow a normal distribution. The General Linear Model (GLM) analysis was therefore applied given that it is a robust test and encompasses a range of comparative statistical tests, including T-tests, ANOVA, regression, and Analysis of Covariance (ANCOVA). It thus allows for the use of a linear model even if the requirements for normality and homogeneity of variances are not always satisfied (Brown & Prescott, 2014). It is also applicable to both categorical and continuous variables (Brown & Prescott, 2014).

To separate the influence of the two factors under investigation within exertion types (i.e., muscle length and muscle force), no comparisons were made between the

“Varying Length” and “Varying Force” conditions, with the exception of cycle time. Furthermore, it is important to mention that the comparable conditions were not only limited to the “Varying Length”, “Pure Static”, and “Varying Force” conditions. For example, peak torque, time to peak torque, Fatigue Index, work, and power under the “Varying Length” were further divided into their concentric and eccentric phases. Similarly, under the “Varying Length” condition, the joint angles and accelerations for the flexion motion were analysed independently of the data for extension. These nuances are further detailed under the relevant sections in the Results chapter (Chapter 5).

CHAPTER 5: RESULTS

The findings presented in this chapter commence with the participants' general characteristics, followed by the recordings of the various fatigue indicators, which include endurance time, number of repetitions, strength measures (e.g., peak torque, time to peak torque, work, power, and the fatigue index), electromyography (amplitude and median frequency), as well as movement accuracy and cycle-to-cycle variability for cycle duration, force, range of motion / joint positions and accelerations. To establish the effects of time (i.e., the development of fatigue and its recovery) on these variables, recordings before/at the beginning of the fatigue protocol were compared to recordings towards the end of the protocol/upon its termination, as well as for five minutes thereafter. To determine the effects of exertion type on fatigue responses, the three muscle actions were compared with one another. However, some variables did not allow for a direct comparison between exertion types due to different characteristics, thus necessitating the separate analysis of the effects of muscle length and muscle force on fatigue responses. Details of these analyses are described under the relevant variables.

5.1 PARTICIPANT CHARACTERISTICS

39 student volunteers, all right-handed, participated in the study; however, due to incomplete data sets, three participants were excluded, resulting in a sample size of 36 (18 females and 18 males). Table 3 shows their summarized descriptive data, together with the p-value when dividing the sample by sex. Since strength expression is known to be significantly influenced by sex, selected demographic and anthropometric data were compared between males and females. Appendix B.1 (Table 19 to Table 22) provides details of these analyses.

Table 3: Summary data of demographic and anthropometric data (mean ±standard deviation; coefficient of variation).

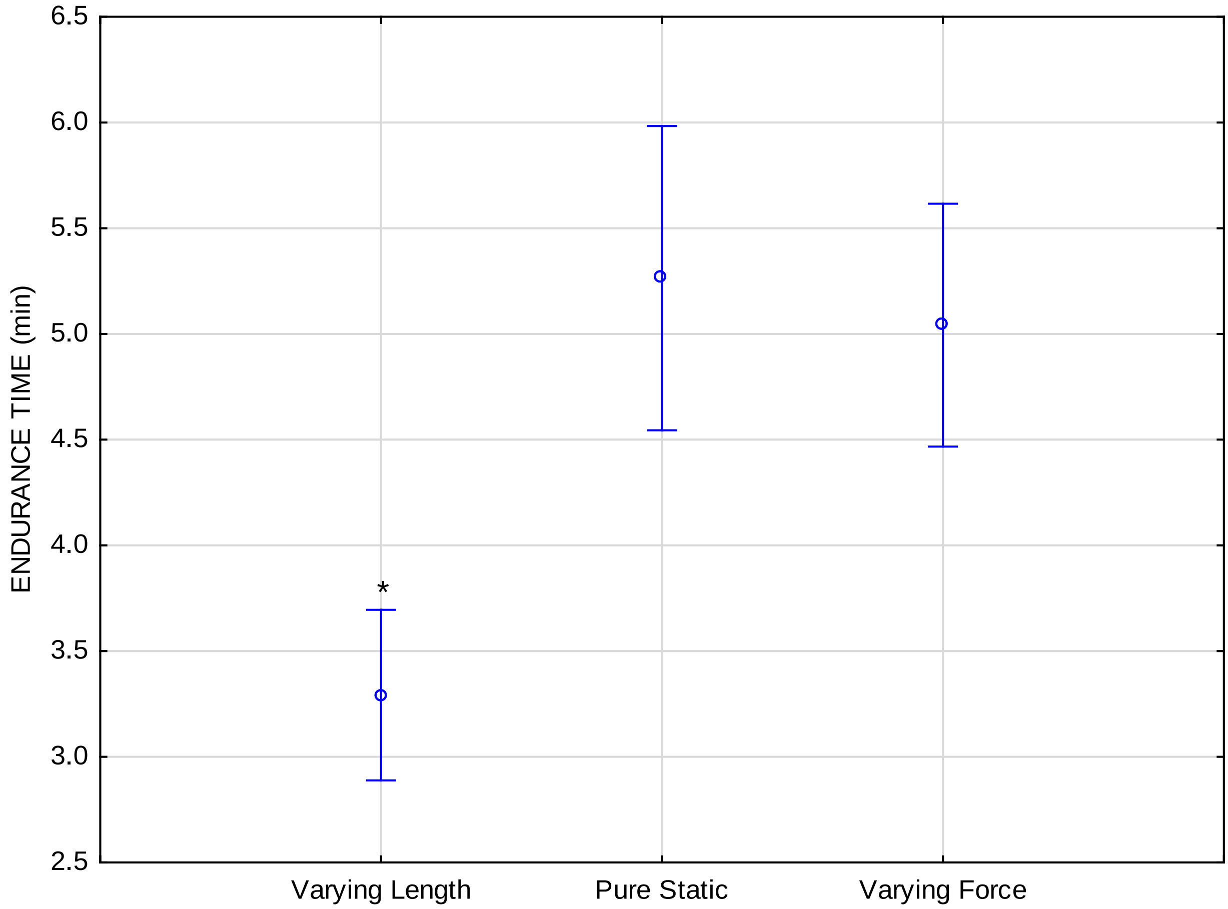
	All (n=36)	Females (n=18)	Males (n=18)	p-value
Age (yrs)	21.83 (±1.80); 8.23%	21.89 (±1.78); 8.13%	21.78 (±1.86); 8.56%	0.856
Stature (m)	1.70 (±0.09); 5.48%	1.64 (0.06); 3.92%	1.76 (±0.08); 4.61%	<0.0001 *
Mass (kg)	72.18 (±15.29); 21.19%	64.99 (±9.08); 13.98%	79.37 (±17.01); 21.44%	0.003 *
BMI ($kg.m^{-2}$)	24.86 (±4.03); 16.22%	24.18 (±3.49); 14.43%	25.55 (±4.51); 17.65%	0.318
Forearm length (m)	0.36 (±0.02); 5.81%	0.35 (±0.01); 4.15%	0.38 (0.02); 4.41%	<0.0001 *
Peak Strength (Isometric) (Nm)	61.17 (±22.86); 37.37%	41.89 (±6.53); 15.59%	80.44 (±15.69); 19.51%	<0.0001 *
Peak Strength (Concentric) (Nm)	53.10 (±21.86); 41.17%	34.74 (±7.55); 21.74%	71.46 (±14.60); 20.44%	<0.0001 *
Peak Strength (Eccentric) (Nm)	58.93 (±24.10); 40.90%	38.62 (±8.32); 21.56%	79.24 (±15.90); 20.07%	<0.0001 *

** Grey-shaded cells indicate a significant difference at p<0.05 between males and females.*

No significant difference was found for age between the sexes, with participants in the sample being, on average, 21.83 years of age. Although the Body Mass Index (BMI) showed no significant sex-related difference, the males were generally larger than the females, given the significant differences in stature ($p<0.0001$), body mass ($p=0.003$) and forearm length ($p<0.0001$). Male participants were also significantly stronger than their female counterparts for the maximum voluntary static and dynamic exertions ($p<0.0001$) obtained prior to the fatigue protocol, both in terms of absolute peak torque but also when relativizing peak torque to body weight ($p< 0.0001$).

5.2 ENDURANCE TIME

Endurance time was defined as the time recorded from the first movement or exertion cycle of the submaximal fatigue protocol to the last cycle of the activity, which corresponded to the time when participants reached an RPE of 9. For all experimental conditions, endurance times were found to be inversely proportional to the load moved, with correlation coefficients ranging between -0.43 and -0.46 (Appendix B.2 – Table 28). Similar associations were found for endurance time and peak torque, with correlation coefficients of between -0.39 and -0.44. All correlations were significant at $p<0.05$ (Appendix B.2 – Table 29). Figure 12 compares the endurance times for the three experimental conditions.



Asterisk (*) indicates a significant difference (p<0.05) to all other experimental conditions.

Figure 12: Endurance times recorded for the three exertion types.

Figure 12 shows the mean endurance times via the open circles as well as the 95% confidence interval via the error bars. Although endurance time appeared longest for the “Pure Static” condition (5.26 ± 2.12min), followed by the “Varying Force” (5.04 ± 1.74min), and shortest for the “Varying Length” condition (3.29 ± 1.23min), significant differences were only found between the “Varying Length” and “Pure Static” exertions ($p=0.0001$), as well as between the “Varying Length” and “Varying Force” conditions ($p=0.0001$). Endurance times were similar for the conditions with a constant force component (i.e., “Varying Length” and “Pure Static”) (Appendix B.3(a) – Table 32). Sex did not significantly affect endurance time ($p=0.107$), although female participants tended towards slightly longer endurance times than the males for all three exertion types (Appendix B.1(f) – Table 25).

The same outcomes were found when, instead of overall endurance time, the number of movement/force cycles were compared to one another (Appendix B.1(g); Table 26 and Appendix B.3(b) - Table 33). Under the “Varying Length” condition, participants completed significantly fewer cycles ($p<0.0001$) with 73.56(±27.06) cycles compared to the “Pure Static” (113.96±49.14) and “Varying Force” exertions (108.00±40.15). As with endurance time, no significant difference was found between the sexes ($p=0.125$).

5.3 MUSCULAR STRENGTH

Differences in muscular strength between the experimental conditions, as well as changes over time, were analysed by considering peak torque, time to peak torque, the Fatigue Index, work, and power responses. Similar to the protocol utilized by Kruger *et al.* (2019), these measures were obtained prior to and immediately upon termination of the fatigue protocol, as well as in one-minute intervals thereafter until five minutes after the end of the exercise protocol. The effectiveness of the recovery process for Peak Torque was analysed by comparing the responses recorded prior to (Pre), immediately after the fatigue protocol (Post-Min 0) and after five minutes of recovery (Post-Min 5). The differences in strength were also compared between all three exertion types.

5.3.1 Peak Torque

Peak torque responses are presented in Table 4, with statistical tables in Appendix B.3(c) - Table 34 and Table 35. The data from the “Varying Length” condition have been separated into its concentric and eccentric components due to the different characteristics of these two exertion types. Given the significant difference found between the sample’s male and female morphologies and peak strength characteristics, the peak torque data are presented relative to body weight (Table 4) and as the percentage change from the pre-protocol values (Figure 13).

Table 4: Summary data of Peak Torque relative to body mass ($Nm.kg^{-1}$) pre-protocol, immediately upon termination of the protocol (Post-Min 0), as well as after five minutes (Post-Min 5) (mean ±standard deviation; coefficient of variation).

	Peak Torque ($Nm.kg^{-1}$)			p-value		
	Pre	**Post-Min 0**	**Post-Min 5**	**Exertion**	**Time**	**Exertion x Time**
Varying Length (concentric)	0.73(±0.26); 34.84%	0.49(±0.21); 42.43%	0.56(±0.18); 32.48%	<0.0001	<0.0001	<0.0001
Varying Length (eccentric)	0.81(±0.28); 34.38%	0.63 (0.22); 34.66%	0.68(±0.23); 33.46%			
Pure Static	0.83(±0.24); 29.02%	0.69(±0.23); 32.56%	0.75(±0.23); 31.04%			
Varying Force	0.84(±0.25); 29.51%	0.71(±0.23); 32.32%	0.76(±0.24); 31.37%			

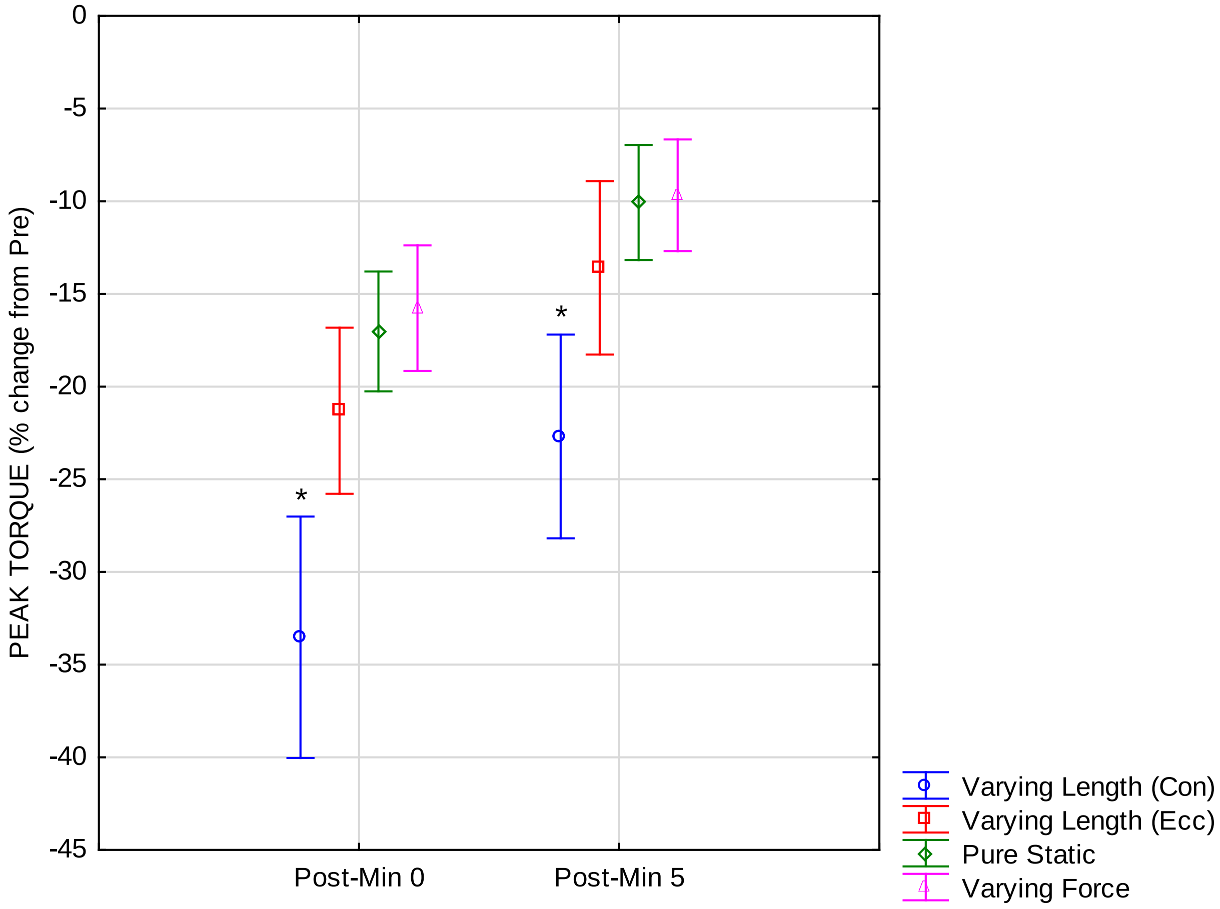
** Grey-shaded cells indicate a significant effect at p<0.05*

Peak torque data, when relativized to bodyweight, showed a significant time effect ($p<0.0001$), condition effect ($p<0.0001$), and interaction effect ($p<0.0001$). Peak torque recorded prior to the exercise protocol ranged between 0.73(±0.26)$Nm.kg^{-1}$ and 0.84(±0.25)$Nm.kg^{-1}$, and exhibited significant decrements in strength ($p=0.0001$) for all exertion types by the first post-protocol measurements (Post-Min 0), dropping to between 0.49(±0.21)$Nm.kg^{-1}$ and 0.71(±0.23)$Nm.kg^{-1}$.

Over the recovery period, peak torque increased significantly from the end of the fatiguing exercise to the fifth minute after termination ($p=0.0001$); however, the values obtained at minute 5 of the recovery period were still significantly lower compared to those obtained prior to the fatigue protocol ($p=0.0001$). Inter-individual variance remained relatively consistent across all time intervals and conditions, with the coefficient of variance never changing by more than 3.54%, except under the "Varying Length-Concentric" condition, whereas the variance range was around 10% (Table 4).

Regarding the condition effect, the "Varying Length–Concentric" condition consistently showed the lowest peak torque responses compared to all other exertion types across all three time intervals ($p=0.0001$). Similarly, the "Varying Length–Eccentric" condition also showed significantly lower peak torque values than the "Pure Static" ($p=0.014$)

and the “Varying Force” (p=0.004) conditions. The latter two exertion types yielded similar responses.



Asterisk (*) indicates a significant difference (p<0.05) to all other exertion types.

Figure 13: Percentage decrease in Peak Torque from pre-fatigue values to post-fatigue-minute 0 and minute 5 values for all exertions.

When considering the percentage change in peak torque relative to the pre-fatiguing values, significant effects were found for time ($p<0.0001$) and exertion type ($p<0.0001$). Immediately upon termination of the exercise protocol, the “Pure Static” and “Varying Force” conditions had undergone the smallest force decrements with values of 17.11±9.29% and 15.55±9.81%, respectively, while the “Varying Length–Concentric” condition yielded the greatest decrease in peak torque (33.59 ±18.70)%. Post hoc analyses revealed the “Varying Length-Concentric” exertion resulted in a significantly greater decrement in Peak Torque compared to the other exertions ($p<0.0001$), while no significant differences were found between the remaining three exertions.

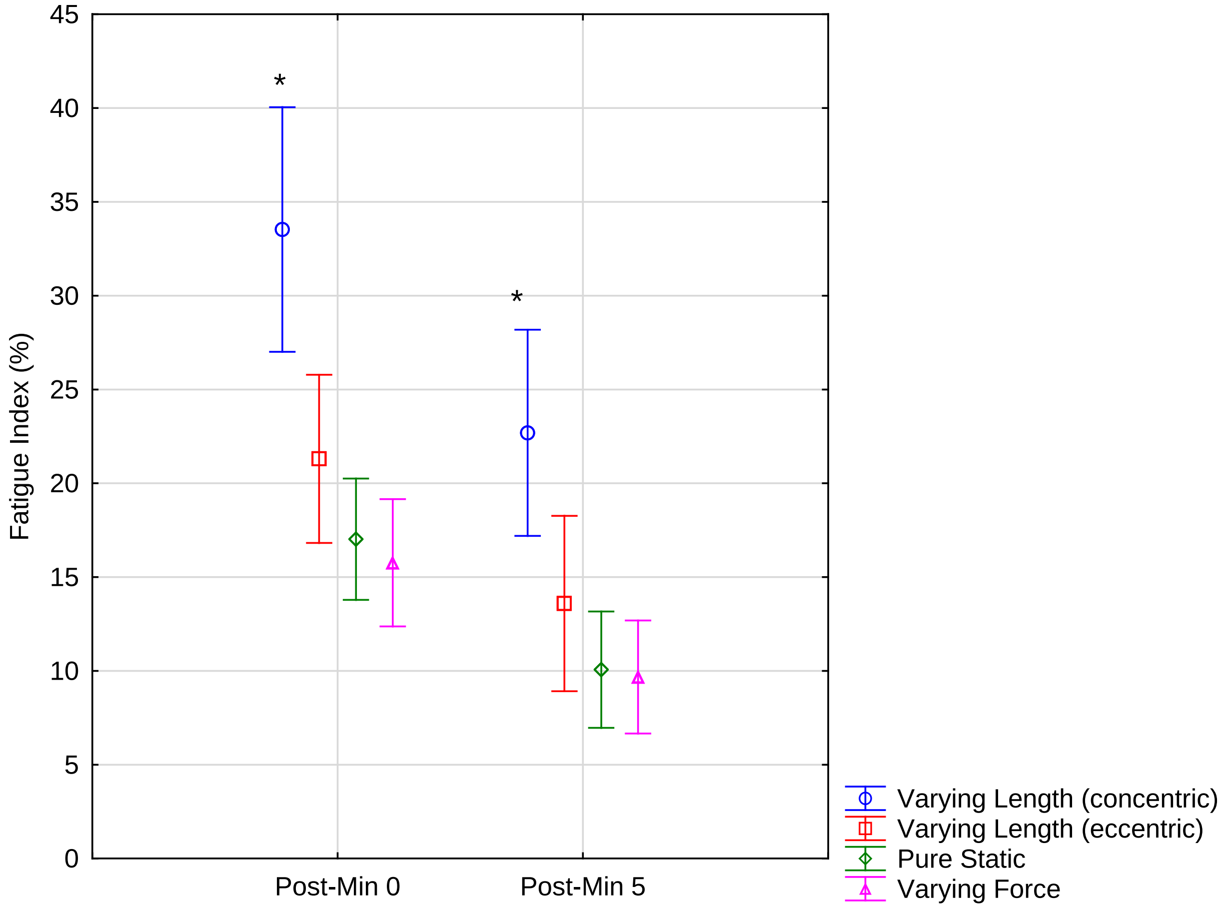
Peak torque responses during the recovery period increased significantly for all exertion types ($p<0.0001$). The "Varying Length-Eccentric", "Pure Static", and "Varying Force" conditions returned to within 14.32(±14.10)%, 10.13(±8.91)%, and 9.67(±8.77)% of pre-fatigue values, respectively, and were not significantly different to one another. However, the decrement in Peak Torque of the "Varying Length-Concentric" muscle action remained significantly larger after five minutes of recovery compared to all other exertions ($p=0.0001$).

5.3.2 Fatigue Index

The Fatigue Index was calculated as follows, utilizing the formula by Ditor (1999) and Morris *et al.* (2008):

$$\textit{Fatigue Index (\%)} = [(Torque_{initial} - Torque_{final}) / Torque_{initial}] * 100$$

A Fatigue Index of close to zero would infer no/minimal fatigue due to the small reduction in torque values, if any. However, the greater the decrement in torque from the initial to the final torque reading, the greater the Fatigue Index, implying greater fatigue. A positive value indicates a lower value for post-fatigue peak torque, compared to pre-fatigue levels. The Fatigue Index values for the three experimental conditions upon termination of the protocol (Post-Min 0) and at the end of the recovery period (Post-Min 5) are presented in Figure 14 and Appendix B.3(d) – Table 36.



Asterisk (*) indicates a significant difference (p<0.05) to all other exertion types.

Figure 14: Fatigue Index results for all exertion types.

Analysis of the Fatigue Index indicated significant effects for time ($p<0.0001$) and exertion type ($p<0.0001$), but no interaction effect ($p<0.434$). The larger the Fatigue Index, the greater the decrement in peak torque from beginning to end of a selected time period. For the current data, this means that both at time intervals, namely Post-Min 0 and Post-Min 5, muscle fatigue was present, with the Post-Min 0 interval reflecting significantly greater fatigue than the Post-Min 5 interval. Fatigue Index values improved between 5.88% and 11.39% over the 5-minute recovery period.

Considering the effect of exertion type on the Fatigue Index ($p<0.0001$), the “Varying Length-Concentric” exertion (33.59±18.70)% showed a significantly higher value, and therefore greater fatigue, compared to all other exertion types, which varied between 15.55(±9.81)% and 21.43(±12.88)%. The “Varying Length-Concentric” condition also showed considerably greater variability in its Fatigue Index. No differences were found in the Fatigue Index between the sexes, although males generally had a greater Fatigue Index compared to females (Appendix B.1(h) – Table 27).

5.3.3 Time to Peak Torque

Time to peak torque was analysed by comparing the pre-fatigue values with those obtained immediately upon termination of the fatigue protocol (Post-Min 0) and after five minutes of recovery (Post-Min 5) (Table 5; Appendix B.3(e) – Table 37). The percentage change in the time to peak torque is also presented in Figure 15 (with statistical tables in Appendix B.3(e) - Table 38).

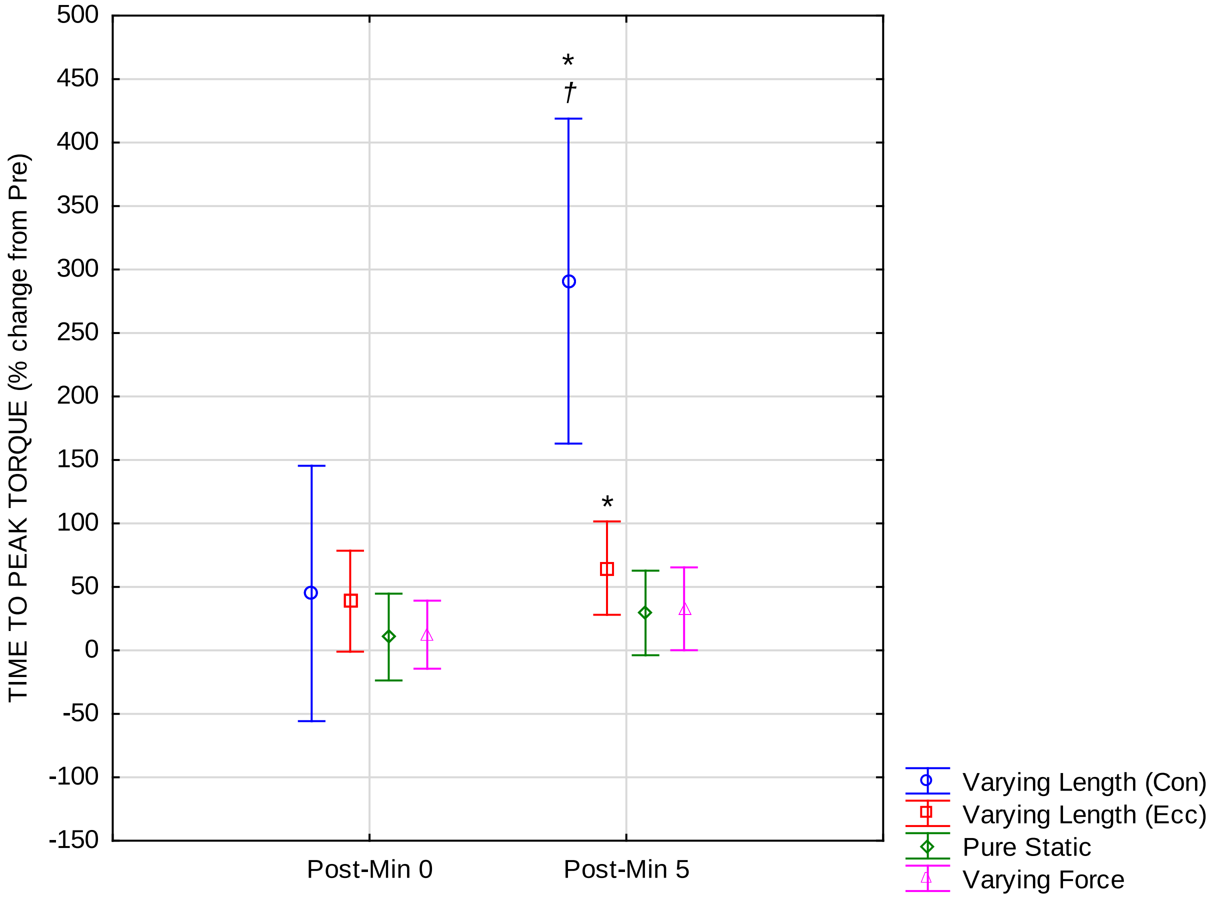
Table 5: Time to Peak Torque (sec) across time (mean ± standard deviation; coefficient of variance).

	Time to Peak Torque (sec)			p-value		
	Pre	**Post-Min 0**	**Post-Min 5**	**Exertion**	**Time**	**Exertion x Time**
Varying Length (concentric)	0.29(±0.36); 127.90%	0.20(±0.28); 139.09%	0.48(±0.31); 65.17%	<0.0001	0.008	0.587
Varying Length (eccentric)	0.71(±0.34); 48.08%	0.79(±0.39); 48.83%	0.88(±0.38); 42.77%			
Pure Static	2.11 (±1.28); 60.49%	1.75(±1.09); 62.19%	2.05(±1.15); 55.90%			
Varying Force	2.15 (±1.06); 49.21%	2.05(±1.23); 59.76%	2.39(±1.23); 51.37%			

** Grey-shaded cells indicate a significant effect at p<0.05*

The time taken to reach peak torque was the shortest for the “Varying Length-Concentric” exertion, followed by the “Varying Length-Eccentric”, both with less than a second (Table 5). The “Pure Static” and “Varying Force” conditions took longer to reach peak torque, with the recorded times being around two seconds. The inter-individual variances under the “Varying Length-Concentric” exertion for “Pre” and “Post-Min 0” were considerably larger than for the remaining conditions and intervals, while the “Varying Length-Eccentric” exertion had the lowest variance.

Statistical analyses revealed that the time to reach peak torque (when analysed in seconds) was significantly affected by the duration of the fatigue protocol and the subsequent recovery (p=0.008). However, post hoc analyses only revealed a significant time effect between the Post-Min 0 and Post-Min 5 intervals, but none between the Pre and Post-Min 0 intervals. Comparisons of the different exertion types revealed a significant condition effect (p<0.0001), with all exertions being significantly different to one another, except for the “Pure Static” and the “Varying Force” conditions.



** indicates a significant difference (p<0.05) to all other conditions in that time interval*

† indicates a significant difference (p<0.05) to Post-Min 0.

Figure 15: Percentage change in the time to peak torque for different exertions.

Analyses of the percentage change in the time to peak torque at the end of the protocol relative to the pre-fatigue time interval (Figure 15) revealed significant time (p=0.001) and exertion (p<0.0001) effects, but no interaction effect (p<0.59). At the end of the fatigue protocol, the time to peak torque had increased by between 35.78% and 36.34% for the “Varying Length – Concentric” and “Varying Length – Eccentric” exertions, respectively, but only between 8.99% and 8.50% for the “Pure Static” and “Varying Force” conditions, respectively. Only the normalized time to peak torque of the “Varying Length – Concentric” condition at Post-Min 5 was significantly longer compared to Post-Min 0. Furthermore, the time to peak torque for “Varying Length – concentric” exertion at Post-Min 5 was significantly greater compared to all other conditions within the same time interval. The “Varying Length–Concentric” exertion also had a considerably greater variance compared to the other exertion types (p<0.0001).

5.3.4 Work and Power

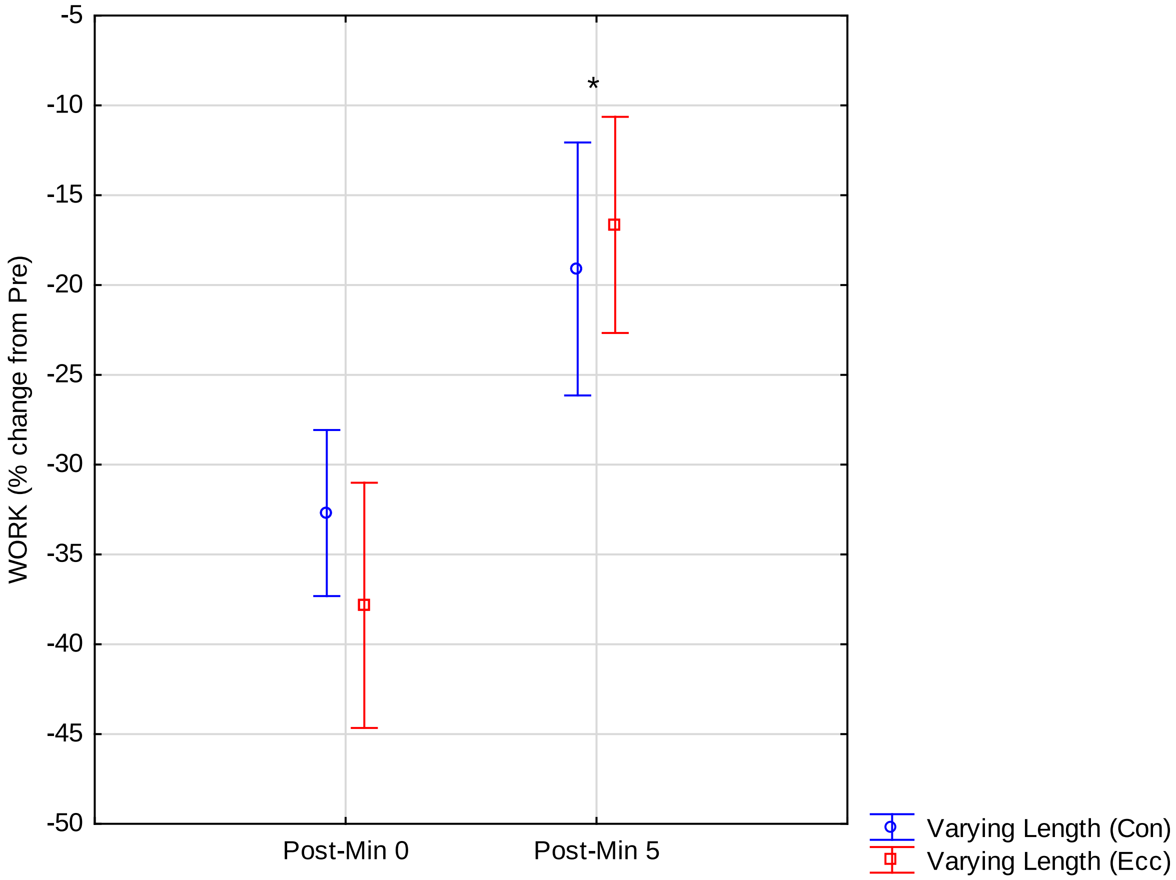
Analyses of the amount of work done and average power were limited to the concentric and eccentric movement phases of the “Varying Length” exertion since the lack of movement (and thus lack of displacement and speed) during the “Pure Static” and the “Varying Force” conditions rendered no outcomes for work or power. Absolute values for work and power are presented in Table 6, while the changes in these parameters over the recovery period are presented in Figure 16 and Figure 17. Appendix B.3(f) (Table 39 and Table 40) and Appendix B.3(g) (Table 41 and Table 42) contain the statistical tables for the work and power analyses, respectively.

Table 6: Work and power for the concentric and eccentric movement phases of the “Varying Length” condition over time (mean ± standard deviation; coefficient of variance).

		Pre	Post (Min 0)	Post (Min 5)	p-value		
					Exertion	**Time**	**Exertion x Time**
Work done (J)	**Varying Length (Con)**	54.99(±21.91) 39.84%	36.11(±14.15); 39.19%	43.29(±17.67); 40.83%	<0.0001	<0.0001	0.025
	Varying Length (Ecc)	68.09(±27.77) 40.78%	43.32(±19.20); 44.33%	55.86(±24.66); 44.14%			
Power (W)	**Varying Length (Con)**	32.62(±14.29) 43.82%	22.04(±8.90); 40.40%	26.08(±11.28); 43.26%	<0.0001	<0.0001	<0.0001
	Varying Length (Ecc)	46.38(±18.94) 40.83%	26.67(±12.79); 47.97%	34.03(±15.08); 44.31%			

** Grey-shaded cells indicate a significant effect at p<0.05*

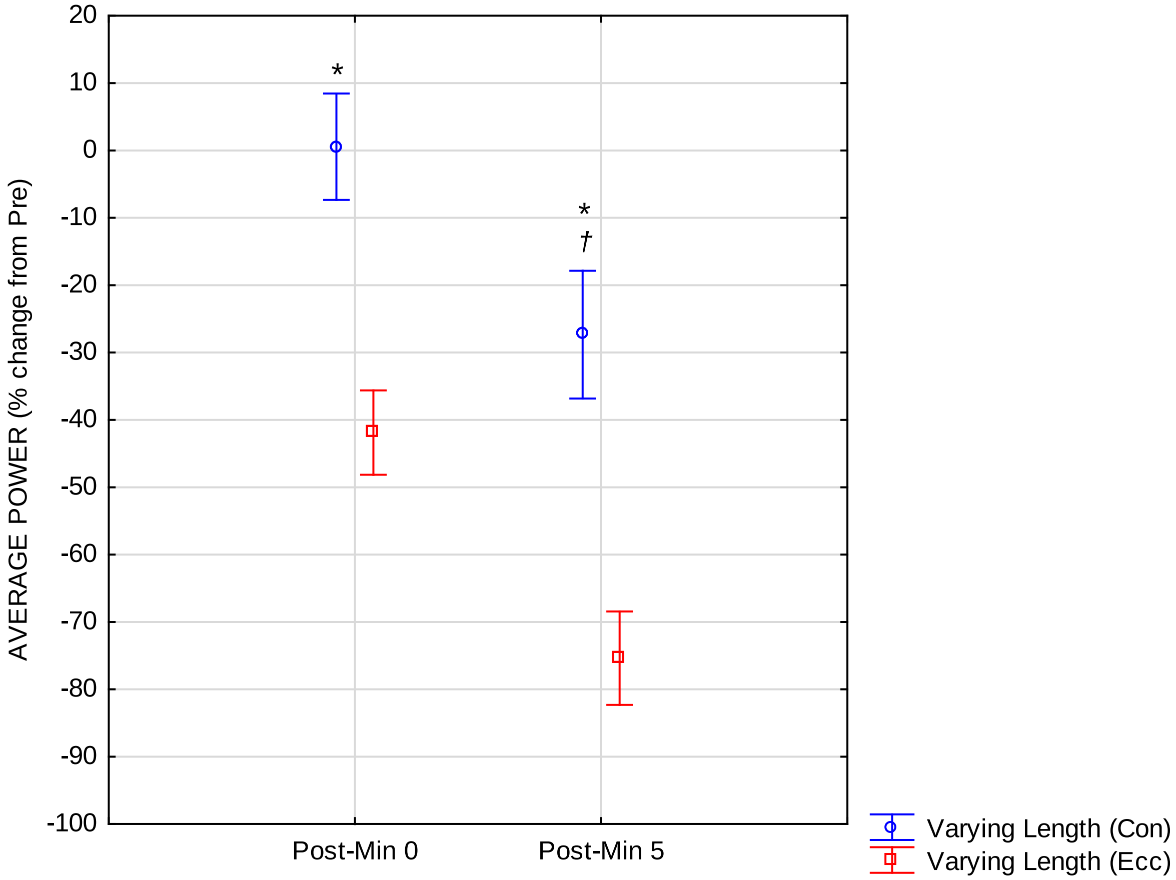
The values for both work done and average power were highest prior to the fatigue protocol (Pre), while the values obtained immediately post-protocol (Post-Min 0) were the lowest, but had increased again by Post-Min 5, albeit below pre-protocol levels. The inferential statistics indicate that the work done and average power were significantly influenced by time (p<0.0001) and exertion type (p<0.0001), as well as the interaction between these two factors (p=0.025 for work and p<0.0001 for power). Both variables were significantly different between all time intervals, and between the concentric and eccentric phases, with the concentric phase consistently resulting in lower work and power output than the eccentric phase.



** indicates a significant difference (p<0.05) to Post-Min 0.*

Figure 16: Work done as a percentage change from pre-fatigue responses.

When considering the change in work done relative to the pre-fatigue measurements (Figure 16), the "Varying Length-Concentric" condition revealed a mean decrement of 32.69(±13.65)% from the pre-protocol values upon termination of the protocol, but only 19.10(±20.81)% after five minutes of recovery. Under the "Varying Length-Eccentric" condition, work done decreased by 36.06(±17.39)% over the duration of the fatiguing exercise but then recovered to 16.65(±17.79)% of its pre-protocol values. While the effect of time was significant ($p<0.0001$), there was no statistical difference found between the exertion types ($p=0.909$), nor an interaction between time and exertion ($p=0.178$).



** indicates a significant difference (p<0.05) between conditions*

† indicates a significant difference (p<0.05) to Post-Min 0.

Figure 17: Average power as a percentage change from pre-fatigue responses.

Data obtained for average power (Figure 17) revealed a slight increase of 0.30(±22.71)% of pre-fatigue values for the concentric exertion phase but a decrement of 41.88(±18.23)% for the eccentric exertion immediately after the termination of the exercise protocol. These values decreased further to between 27.96(±27.46)% and 75.13(±19.94)% after the 5-minute recovery period, respectively. This time effect was found to be significant ($p<0.0001$), as were differences between the concentric and eccentric muscle exertions ($p<0.0001$). There was, however, no interaction effect between the time intervals and exertion type.

Correlation analyses revealed significant and strong associations between peak torque and work done for both concentric (r=0.81-0.87) and eccentric (r=0.75-0.84) exertions (Appendix B.2 - Table 30). Similarly, correlation coefficients for peak torque and power ranged between 0.81-0.84 and 0.67-0.76 for the concentric and eccentric exertions, respectively (Appendix B.2 - Table 31).

5.4 ELECTROMYOGRAPHY

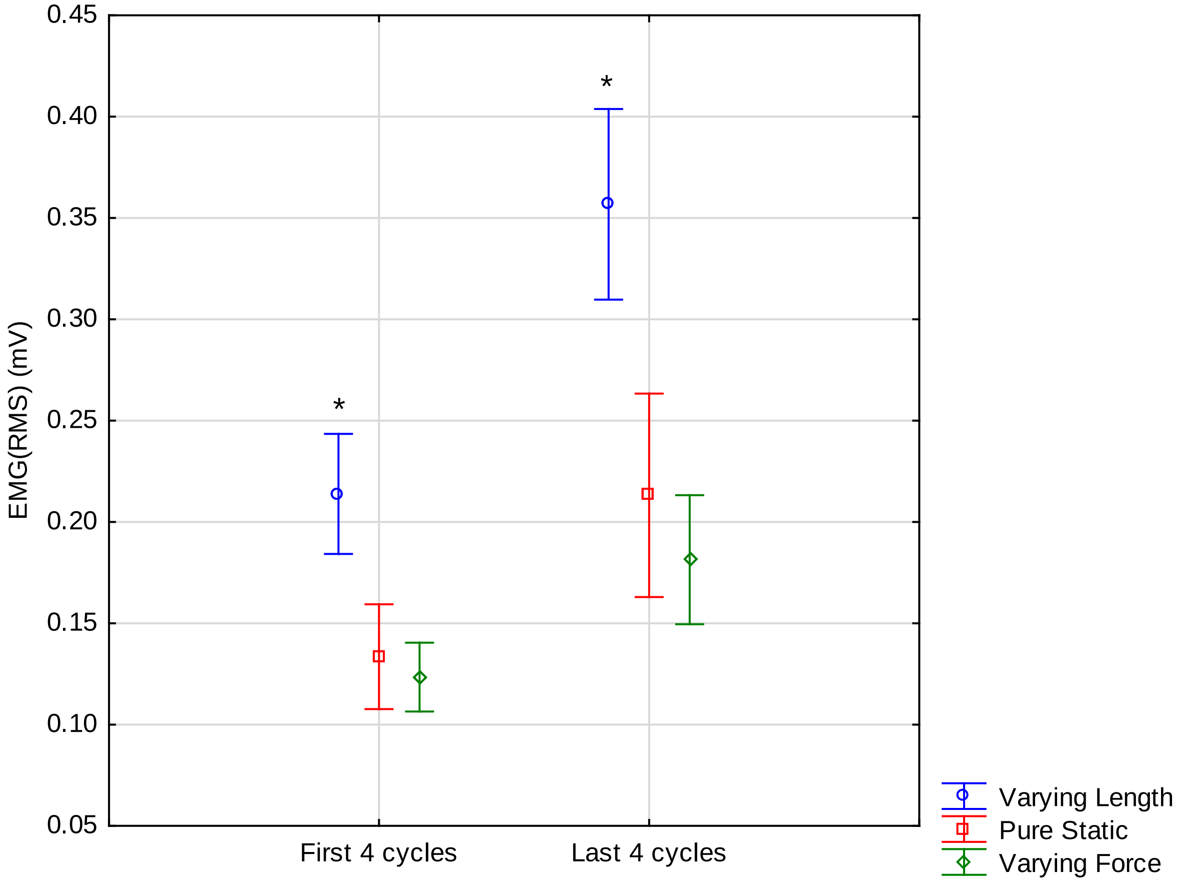
5.4.1 Amplitude (EMG-RMS)

The root-mean-square (RMS) in electromyography is an indicator of the amplitude of the EMG signal and, thus, the level of muscle activation. Muscle fatigue was inferred by comparing the EMG amplitude over time (Table 7; Figure 18). Additionally, the effects of exertion type on fatigue were also evaluated by considering the absolute differences (in mV), as well as calculating the percentage change from the first four to the final four cycles/intervals (Table 7; Figure 19). For EMG, the "Varying Length" conditions were not separated into their concentric and eccentric phases. Statistical tables for the EMG amplitude are in Appendix B.3(g) - Table 43 and Table 44.

Table 7: Electromyography RMS data (presented in mV) (mean ± standard deviation; coefficient of variance) and GLM results.

	First 4 cycles	**Last 4 cycles**	**p-value**		
			Exertion	**Time**	**Exertion x Time**
Varying Length	0.21(±0.09); 40.92%	0.36(±0.14); 38.96%	<0.0001	<0.0001	<0.0001
Pure Static	0.13(±0.08); 57.23%	0.21(±0.15); 69.65%			
Varying Force	0.12(±0.05); 40.71%	0.18(±0.09); 51.89%			

** Grey-shaded cells indicate a significant effect at p<0.05*



** indicates a significant difference (p<0.05) to all other conditions within that interval.*

Figure 18: EMG(RMS) of the first four cycles vs the last four cycles of the fatigue protocol.

As depicted in Figure 18, significant differences in the EMG(RMS) were found for time (i.e., first four vs last four cycles; $p<0.0001$), exertion type ($p<0.0001$), as well as their interaction ($p<0.0001$). The level of muscle activation was significantly greater at the end of the fatigue protocol, compared to that at the start, with the “Varying Length” condition recording significantly higher EMG amplitudes compared to the other two exertion types. Also noteworthy is that the EMG(RMS) variability for the “Varying Force” condition was considerably smaller than for the other two conditions, both at the beginning and at the end of the fatigue protocol, and that variability increased with time for all exertion types.

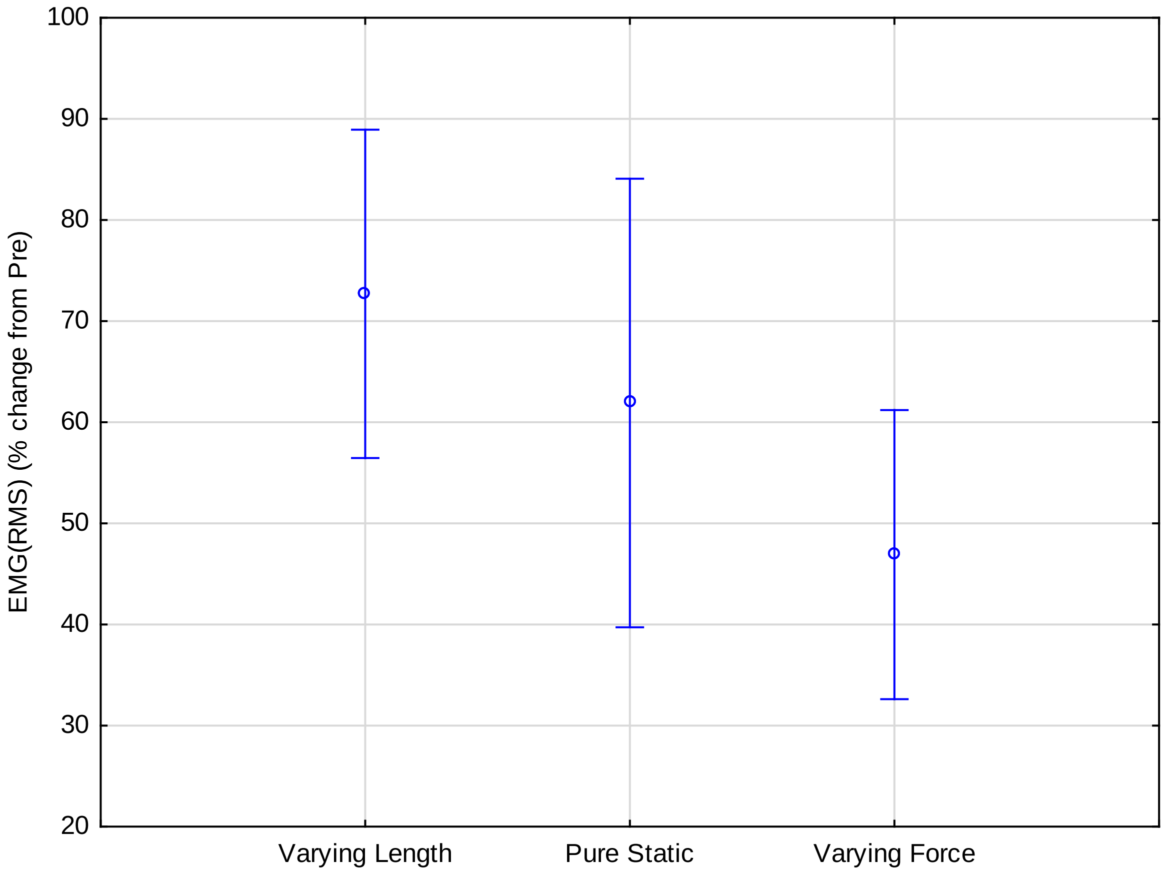


Figure 19: Effects of exertion type on the percentage change in EMG(RMS) from the first four cycles to the last four cycles of the fatigue protocol.

However, when comparing the percentage change in the EMG(RMS) from the start to the end of the protocol (Figure 19), no statistically significant effects were found between exertion types (p=0.067), even though the ”Varying Length” exertion still showed the tendency towards the greatest proportional increase in EMG amplitudes with 72.69(±47.99)%, and the ”Varying Force” condition had the lowest increase with 46.90(±42.25)%. Variability was, however, very large, with the coefficients of variation for the different exertion types ranging between 66.02% and 105.91%.

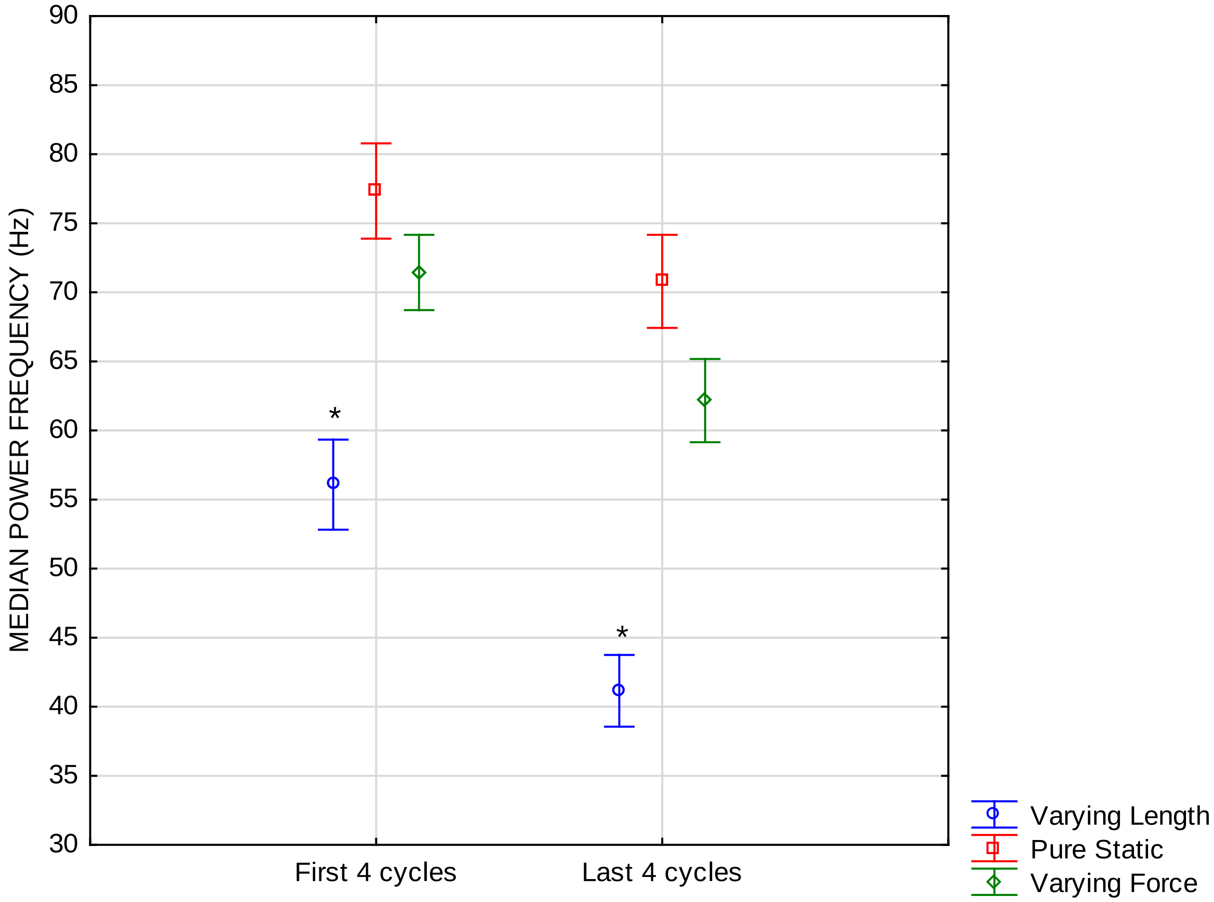
5.4.2 Median Frequency

Analysis of the so-called power spectrum is another technique to determine whether fatigue developed and can involve either mean or median values (Iridiastadi & Nussbaum, 2006; Zwarts *et al.* 2008). Only the median frequency values of the power spectrum are presented in Table 8, Figure 20 and Figure 21, as statistical analyses using the mean values yielded similar outcomes. As with EMG(RMS), the first four cycles were compared with the last four, and the percentage change was calculated to determine the effects of exertion type on fatigue. The statistical tables for the median frequency are presented in Appendix B.3(i) (Table 45 and Table 46).

Table 8: Electromyography Median Frequency data (mean ± standard deviation; coefficient of variance) with GLM results.

	First 4 cycles	Last 4 cycles	p-value		
			Exertion	**Time**	**Exertion x Time**
Varying Length	56.08(±9.64); 17.19%	41.16(±7.68); 18.67%	<0.0001	<0.0001	<0.001
Pure Static	77.34(±10.20); 13.18%	70.79(±9.96); 14.07%			
Varying Force	71.44(±8.07); 11.29%	62.17(±8.91); 14.34%			

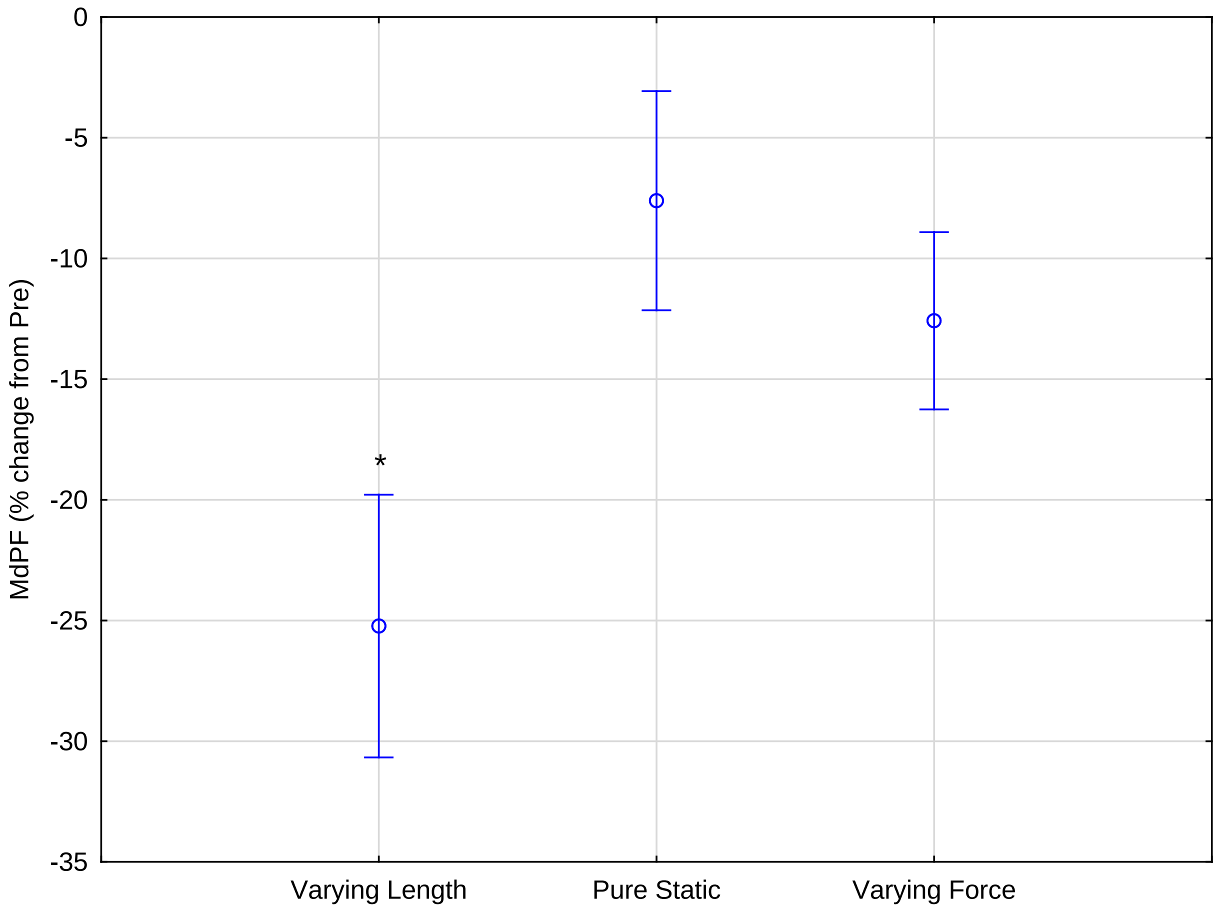
** Grey-shaded cells indicate a significant effect at p<0.05*



** indicates significant difference (p<0.05) to all other conditions.*

Figure 20: Effects of Time and Exertion Type on the median frequency.

The median frequency of the EMG spectrum, depicted in Figure 20, decreased significantly over time ($p<0.0001$), indicating a shift in the power spectrum to the left and therefore fatigue development. Furthermore, median values were significantly lower for the “Varying Length” condition compared to the other two exertion types ($p<0.0001$). The interaction effect between the two factors was also statistically significant ($p<0.001$). The coefficients of variation were consistently low for all conditions, ranging between 9.03 to 14.74%.



** indicates a significant difference (p<0.05) from all other conditions.*

Figure 21: Percentage change in the median frequency for all exertion types.

Comparisons of the percentage change in the median frequency for the three exertion types (Figure 21) revealed a significantly greater reduction in the median frequency for the “Varying Length” condition (25.23±16.08%) compared to the “Pure Static” (7.61±13.42%) and “Varying Force” (12.58±10.85%) exertions ($p<0.001$), but no difference between the latter two.

5.5 PERFORMANCE ACCURACY AND VARIABILITY

The inclusion of analyses relating to the accuracy and variability of motor performance was based on the assumption that increasing fatigue would lead to interferences with motor control, which in turn would influence the accuracy of the performance objectives of the experimental conditions, and, over numerous repetitions, their cycle-to-cycle variability. The performance accuracy outcomes considered in this study refer to the ability to achieve the intended objectives of the various exertions, i.e., the requirements of each exertion cycle. This meant adhering to the prescribed cycle times for the "Varying Length" and "Varying Force" conditions, reaching the required endpoints of the joint range of motion for the "Varying Length" condition, and achieving the set force requirements for the "Pure Static" and "Varying Force" conditions. Analyses of accelerations during the "Varying Length" condition were also considered since they provided insights into the mechanical performance itself, rather than its outcome.

The mean data for cycle times, forces, joint angles, and accelerations were compared for performance accuracy over time (first four vs last four cycles) and between exertions, where appropriate, while the range, standard deviation, and coefficient of variance of these variables indicated the cycle-to-cycle variability. Analyses of accelerations for the "Varying Length" condition were included as these could provide insights into the execution of this dynamic condition. The analyses of performance accuracy and cycle-to-cycle variability aimed to provide insights into any fatigue-related changes between the different exertion types during the sub-maximal protocol, rather than the maximal exertions prior to and after the protocol.

5.5.1 Cycle Time

Varying Length & Varying Force

Cycle times were identified under the "Varying Length" and "Varying Force" conditions using the minimum and maximum values obtained from the goniometry and force gauge data, respectively. In addition to the absolute cycle time, the deviation from the 2.6-second target cycle time was calculated (Table 9).

Table 9: Summary data and GLM results for mean cycle times (sec) and deviation from target cycle time (sec) for the "Varying Length" and "Varying Force" conditions (means ± standard deviation; coefficient of variation in percent).

		First 4 cycles	Last 4 cycles	p-values		
				Exertion	**Time**	**Exertion x Time**
Cycle time (sec)	**Varying Length**	2.64 (±0.14); 5.47%	2.68 (±0.20); 7.40%	0.002	0.922	0.210
	Varying Force	2.62 (±0.10); 3.85%	2.58 (±0.08); 3.17%			
Deviation from target time (sec)	**Varying Length**	0.13 (±0.12); 89.25%	0.20 (±0.166); 79.68%	0.418	0.069	0.253
	Varying Force	0.18 (±0.10); 56.39%	0.18 (±0.09); 47.91%			

* *Grey-shaded cell indicates a significant effect at p<0.05. Also note that the "Pure Static" condition did not undergo any cyclical variations in force or joint angles and was thus not included in this analysis*

The GLM analyses for accuracy entailed comparing the mean cycle times of the "Varying Length" and "Varying Force" conditions over time (Appendix B.3(j); Table 47). Table 9 indicates that exertion type had a significant effect on the overall cycle time (p=0.002), with the "Varying Length" exertion resulting in longer mean cycle times than the "Varying Force" condition. However, when considering the mean deviation from the target cycle time of 2.6 seconds, the differences between exertion types were no longer significant. Similarly, no significant time or interaction effects were identified for the mean cycle time, nor the deviation from the target.

Table 10: Summary data and GLM results for cycle time variability (sec) for the "Varying Length" and "Varying Force" conditions (means ± standard deviation; coefficient of variation in percent).

		First 4 cycles	Last 4 cycles	p-values		
				Exertion	Time	Exertion x Time
Range	**Varying Length**	0.311 (±0.280); 89.89%	0.471 (±0.458); 97.30%	0.049	0.158	0.167
	Varying Force	0.507 (±0.321); 63.38%	0.507 (±0.282); 55.68%			
Standard Deviation	**Varying Length**	0.143 (±0.130); 90.75%	0.209 (±0.211); 101.13%	0.063	0.200	0.213
	Varying Force	0.225 (±0.139); 61.68%	0.225 (±0.121); 53.95%			
Coefficient of Variation	**Varying Length**	5.291 (±4.156); 78.56%	7.585 (±6.777); 89.35%	0.018	0.179	0.218
	Varying Force	8.652 (±5.459); 63.10%	8.720 (±4.698); 53.88%			

** Grey-shaded cells indicate a significant effect at p<0.05*

Analyses of the cycle-to-cycle variability data in Table 10 (with statistical tables in Appendix B.3(n); Table 55 to Table 57) showed no significant changes in cycle time from the start to the end of the protocol, nor an interaction effect. Significant differences between exertion types were, however, identified for the range (p=0.049) and the coefficient of variation (p=0.018), but not the standard deviation. Irrespective of the variability measure, the "Varying Force" condition consistently tended to indicate greater variability between the cycles.

5.5.2 Force

The purpose of assessing variations in force recordings was to determine whether the accuracy and consistency of force application would differ over time and between experimental conditions. This was determined by comparing the mean deviation of forces from the target force at the start and the end of the protocol, as well as the variations in timing from cycle to cycle (i.e., range, standard deviation, and coefficient of variance).

Results for forces exerted under the “Varying Length” and “Pure Static” conditions are presented together in Table 11 and Table 12 since both conditions aimed at maintaining an external load of 25% of maximum voluntary exertion (MVE), with the only difference between these two conditions being that the elbow flexion angle of the “Pure Static” condition remained at 90^o, while for the “Varying Length” condition, the forearm would move through a set range of motion. Given the commonality of a constant force, these two conditions were not only assessed for changes in mean force and cycle-to-cycle variability over time (i.e., first four cycles vs last four cycles) but also between conditions. Force data under the “Varying Force” condition (Table 13), on the other hand, were only assessed for changes over time since the force applied to the pulley handle had to vary between the set minimum and maximum values. It is important to mention that the force readings in the following analyses are presented in kilograms (kg), rather than Newtons, since this was the measurement unit recorded by the Futek load cell. Furthermore, since the target force differed between participants, force data were analysed as the absolute deviation from the target force values, but also the deviation relative to the target load.

Varying Length & Pure Static

Table 11: Summary data for mean deviation from target force under the “Varying Length” and “Pure Static” conditions (means ± standard deviation; coefficient of variation in percent).

		First 4 cycles	**Last 4 cycles**	**p-values**		
				Exertion	**Time**	**Exertion x Time**
Deviation from Target Force	**Varying Length (kg)**	0.81(±0.43); 53.25%	0.88(±0.46); 51.98%	<0.0001	0.005	0.086
	Pure Static (kg)	0.48(0.16); 32.98%	0.51(±0.19); 36.83%			
	Varying Length (%)	21.16(±6.08); 28.74%	22.95(±6.29); 27.41%	<0.0001	0.034	0.055
	Pure Static (%)	12.01(±4.41); 36.75%	12.25(±3.26); 26.61%			

** Grey-shaded cells indicate a significant effect at p<0.05*

Accuracy of achieving the set target force of 25% of MVE showed significant time and exertion effects, irrespective of whether the deviation values were analysed in absolute force (kg) or as a percentage of each participant's load held/moved (Appendix B.3(k); Table 48 and Table 49). Force deviations increased significantly from the start to the end of the fatigue protocol ($p<0.0001$), while the force deviations were consistently higher under the "Varying Length" condition compared to the "Pure Static" exertion ($p=0.005$ and $p=0.034$). No interaction effects were identified between exertion type and time for both accuracy measures.

Table 12: Summary data and GLM results for cycle time variability (% deviation from target) for the "Varying Length" and "Pure Static" conditions (means ± standard deviation; coefficient of variation in percent).

		First 4 cycles	Last 4 cycles	p-values		
				Exertion	**Time**	**Exertion x Time**
Range	**Varying Length**	2.63(±1.25); 47.56%	2.93(±2.46); 83.93%	<0.0001	0.150	0.070
	Pure Static	5.61(±4.56); 81.25%	4.16(±1.83); 44.08%			
Standard Deviation	**Varying Length**	1.19(±0.56); 46.99%	1.31(±1.10); 84.03%	<0.0001	0.138	0.075
	Pure Static	2.50(±2.08); 83.20%	1.84(±0.81); 44.08%			
Coefficient of Variation	**Varying Length**	0.06(±0.03); 56.61%	0.06(±0.07); 106.97%	<0.0001	0.403	0.329
	Pure Static	0.19(±0.10); 54.61%	0.17(±0.12); 69.86%			

* *Grey-shaded cells indicate a statistically significant effect at p<0.05.*

When considering the cycle-to-cycle variability for force, when relativized to the load moved or held, a significant difference was only found between the two conditions ($p<0.0001$) with the "Pure Static" condition revealing a significantly greater range, standard deviation and coefficient of variance, compared to the "Varying Length" condition (Table 12). No significant effects were identified for time or the interaction between exertion type and time, although the interaction effects for the range and standard deviations approached significance ($p=0.07$ and $p=0.075$, respectively). Detailed statistical tables can be found in Appendix B.3(o); Table 58 to Table 60.

Varying Force

During the “Varying Force” condition, participants attempted to exert forces against the handle of the pulley system that would vary between 15% and 35% of the maximum voluntary forces recorded for each individual. An analogue scale provided visual feedback to participants. The analyses conducted aimed to determine any significant changes over time for both the mean minimum and maximum values achieved, as well as the cycle-to-cycle variability (Table 13) of minimum and maximum forces achieved under the “Varying Force” condition (refer to Appendix B.3(k); Table 50).

Table 13: Summary data (means ± standard deviation; coefficient of variation) and GLM results for accuracy and variability (presented as a percentage deviation from target force) of minimum and maximum forces achieved under the “Varying Force” condition.

		First 4 cycles	Last 4 cycles	p-values		
				Time	Force (Max/Min)	Time x Force
% Deviation from Target Force	**Maximum Force**	8.31(±3.86); 46.47%	8.41(±3.95); 46.96	0.15	0.0003	0.1687
	Minimum Force	12.10(±6.80); 56.18%	13.92(±8.29); 59.53%			
Range	**Maximum Force**	5.88(±3.06); 52.02%	6.56(±3.63); 55.23%	0.795	<0.0001	0.336
	Minimum Force	10.10(±8.05); 79.68%	8.91(±5.53); 62.11%			
Standard Deviation	**Maximum Force**	2.58(±1.31); 50.82%	2.91(±1.65); 56.60%	0.735	<0.0001	0.273
	Minimum Force	4.59(±3.53); 76.84%	3.98(±2.42); 60.95%			
Coefficient of Variation	**Maximum Force**	36.11(±20.65); 57.18%	38.60(±19.23); 49.81%	0.543	<0.0001	0.521
	Minimum Force	0.45(±0.29); 63.25%	0.39(±0.28); 73.58%			

* *Grey-shaded cells indicate a significant effect at p<0.05.*

The analyses show that mean deviations from the set force levels were significantly different between the two force levels ($p=0.003$), with the mean deviation being greater for the minimum force levels than the maximum levels. However, no time, nor interaction effects were identified.

Cycle-to-cycle variability for range, standard deviation and coefficient of variation all yielded similar outcomes, with no time or interaction effects, but with the expected significant differences ($p<0.0001$) between the maximum and minimum deviation for the target force. For range and standard deviation, the minimum force had a greater cycle-to-cycle variability, whereas the maximum force showed a considerably greater coefficient of variation (Appendix B.3(o); Table 61 to Table 63).

5.5.3 Goniometry

Variations in muscle length were assessed using the minimum and maximum joint angle values recorded by the goniometer while participants performed the fatigue protocol under the three experimental conditions. As with force, the purpose of assessing variations in joint angles was to determine the performance accuracy and outcome variability over four cycles at the start and four cycles at the end of the fatigue protocol. Due to the different movement characteristics of the experimental conditions, changes in joint angles for the “Varying Length” condition were assessed separately from the “Pure Static” and “Varying Force” conditions.

Varying Length

For the “Varying Length” condition, the mean range of motion (ROM) and deviations from the target ROM were only analyzed across time. However, no comparisons were made to the other two experimental conditions. This is because this condition entailed active movement by reaching distinct minimum and maximum endpoints, totalling a full range of motion of 80 degrees, while the other two conditions did not include movement. Table 14 shows the summary data for the full ROM achieved, as well as the deviation from the target ROM. Furthermore, Table 15 details the maximum joint angles achieved during flexion and extension, both in absolute terms as well as relative to the target angles of 40^{o} for flexion and 40^{o} for extension. The data were additionally analysed for time and movement effects. The detailed statistical tables are presented in Appendix B.3(l); Table 51 and Table 52 (range of motion) and Table 53 (joint angles at ROM endpoints).

Table 14: Summary data (means ± standard deviation; coefficient of variation in percent) and GLM results for overall movement accuracy for range of motion (ROM) and degrees of deviation from the target range of motion under the “Varying Length” condition.

	First 4 cycles	Last 4 cycles	p-value
Mean Range of Motion (º)	63.38(±9.72); 14.87%	60.23(±9.51); 15.79%	<0.0001
Deviation from Target Range of Motion (º)	14.80(±9.49); 64.13%	20.01(±9.05); 45.24%	<0.0001

** Grey-shaded cells indicate a significant effect at p<0.05*

Table 15: Summary (means ± standard deviation; coefficient of variation in percent) and GLM results for maximum joint angle and deviation from the target angle (in degrees) for the flexion and extension movement phases under the “Varying Length” condition.

		First 4 cycles	Last 4 cycles	p-value		
				Time	**Move-ment**	**Time x Move-ment**
Maximum Joint Angle (º)	**Flexion**	35.60(±8.38); 23.55%	37.60(±9.93); 26.41%	<0.0001	<0.0001	<0.0001
	Extension	29.78(±9.59); 32.19%	22.63(±9.14); 40.39%			
Deviation from Target Angle (º)	**Flexion**	7.68(±5.63); 73.32%	8.23(±6.09); 74.05%	<0.0001	0.0002	<0.0001
	Extension	11.75(±7.59); 64.56%	17.70(±8.46); 47.82			

** Grey-shaded cells indicate a significant effect at p<0.05*

When dividing the overall ROM into its flexion and extension endpoints, comparisons of the joint angle endpoints revealed significant time, movement, and interaction effects, irrespective of the measurement unit. Flexion endpoints increased by 2º over time, while extension endpoints decreased by over 7º. When considering the mean angular deviation from the target angles, accuracy became worse over time as indicated by increasing values. Comparisons of the movement endpoints revealed a smaller deviation from the target angle for flexion compared to extension.

The results presented in Table 16 (and with details of the statistical analyses in Appendix B.3(p); Table 64 to Table 66), show that the cycle-to-cycle range, standard deviation, and coefficient of variance measures for mean ROM.

Table 16: Summary (means ± standard deviation; coefficient of variation in percent) and GLM results for variability data, comparing times and movement phases under the “Varying Length” condition.

			First 4 cycles	Last 4 cycles	p-values: Time	p-values: Movement	p-values: Time x Movement
Mean ROM (°)	Range	Flexion	6.62(±3.60); 54.29%	6.33(±3.38); 53.39%	0.848	<0.0001	0.397
		Extension	3.03(±2.03); 67.04%	3.49(±2.00); 57.41%			
	Standard deviation	Flexion	3.01(±1.62); 53.63%	2.84(±1.52); 53.63%	0.966	<0.0001	0.345
		Extension	1.37(±0.95); 69.44%	1.57(±0.88); 56.08%			
	Coefficient of Variation	Flexion	8.87(±5.12); 57.73%	7.72(±3.91); 50.61%	0.070	0.454	0.003
		Extension	5.13(±4.25); 82.77%	9.36(±10.59); 113.10			
Mean Deviation from Target Angle (°)	Range	Flexion	5.71(±3.37); 59.08%	5.22(±3.22); 61.74%	0.968	<0.0001	0.271
		Extension	2.91(±1.89); 64.99%	3.43(±2.02); 58.85%			
	Standard deviation	Flexion	2.60(±1.50); 57.60%	2.31(±1.45); 62.41%	0.887	<0.0001	0.220
		Extension	1.31(±0.89); 67.52%	1.55(±0.89); 57.47			
	Coefficient of Variation	Flexion	49.05(±35.27); 71.89%	38.76(±24.59); 63.45%	0.011	<0.0001	0.862
		Extension	21.08(±27.64); 131.10%	12.32(±11.54); 93.72			

** Grey-shaded cells indicate a significant effect at p<0.05*

The cycle-to-cycle variability of the mean ROM did not change significantly from the start to the end of the fatigue protocol. However, a movement effect was identified for the ROM's range and standard deviation ($p<0.0001$), with flexion endpoints being consistently larger than those for extension. The coefficient of variance yielded no significant movement effect, but a significant interaction effect ($p=0.003$).

Considering the deviation of the minimum and maximum joint angles from 40^o, time only yielded a significant effect on the coefficient of variation ($p=0.011$), which decreased over time. All three variability measures showed significant differences between the flexion and extension movements. More specifically, the variability of the flexion endpoint decreased, while the endpoints during extension increased, but only for the range and the standard deviation.

Pure Static & Varying Force

Contrary to the "Varying Length" condition, the "Pure Static" and "Varying Force" conditions did not undergo any limb movement and, therefore, did not have a range of motion per se. Instead, participants had been instructed to maintain their elbow joint at a set angle of 90 degrees. During the "Pure Static" condition, this angle had to be maintained while also exerting a constant force, while under the "Varying Force" condition, varying levels of force were applied against a stopper attached to the pulley system, but without changing the elbow angle. The timing of a 'cycle' for the "Varying Force" condition was defined by marking the cyclical changes in the force exerted, while for the "Pure Static" condition, a 'cycle' was identified at a duration of 2.6 seconds since this was the set cycle time for all experimental conditions. Table 17 depicts the mean deviations from the 90^o elbow flexion angle over time (with statistical tables in Appendix B.3(l) - Table 51). The measures of cycle-to-cycle variability at the start of the protocol were also compared to those at the end of the protocol, as well as between exertion types (Appendix B.3(p) - Table 67 to Table 69).

Table 17: Summary data and GLM results for mean deviations from the target joint angle and movement variability during the "Pure Static" and "Varying Force" conditions for the first and the last four movement cycles (means ±standard deviation; coefficient of variation in percent).

		First 4 cycles	Last 4 cycles	p-values		
				Exertion	Time	Exertion x Time
Mean deviation (°)	Pure Static	0.69(±0.35); 51.13%	0.63(±0.33); 52.03%	<0.0001	0.018	0.084
	Varying Force	3.37(±0.93); 27.70%	3.04(±1.05); 34.53%			
Range (°)	Pure Static	1.01(±1.00); 99.13%	0.70(±0.70); 99.33%	0.874	0.112	0.400
	Varying Force	0.88(±0.61); 69.19%	0.79(±0.43); 54.07%			
Standard Deviation (°)	Pure Static	0.47(±0.48); 101.62%	0.32(±0.32); 100.38%	0.643	0.099	0.342
	Varying Force	0.39(±0.27); 68.72%	0.35(±0.18); 51.35%			
Coefficient of Variation (%)	Pure Static	59.29(±28.72); 48.43%	45.85(±24.28); 52.97%	<0.0001	0.114	0.070
	Varying Force	12.75(±9.94); 77.97%	14.20(±13.87); 97.67%			

** Grey shaded cells indicate a significant effect at p<0.05*

Comparisons of the average deviations from the set 90º elbow flexion angle, revealed significant condition (p<0.0001) and time effects (p=0.018), but no interaction between exertion and time, although the p-value trended towards significance (p=0.084). The deviation from the target joint angle of 90º decreased with time, while the "Varying Force" condition consistently yielded a significantly larger deviation from the target compared to the "Pure Static" condition.

Cycle-to-cycle variability yielded no significant condition, time, or interaction effects, with the exception of the coefficient of variation, which indicated a significant difference between the "Pure Static" and "Varying Force" exertions (p<0.0001) and an interaction effect that fell just short of the significance threshold (p=0.07). The "Pure Static" condition had a significantly greater average coefficient of variation, between 46% to

59% than the “Varying Force” condition, which had a coefficient of variation ranging only between 13% to 14%.

5.5.4 Accelerometry

The inclusion of accelerometry data was to gauge the performance variability of the “Varying Length” condition since this was the only condition that exhibited purposeful movement (and therefore velocities and accelerations). Acceleration data for the “Varying Length” condition were analysed for any time effects by comparing the average maximum accelerations obtained for the first four feasible cycles to those of the final four cycles of the fatigue protocol. Peak accelerations during the elbow flexion phase have been separated from those of the extension phase (Table 18, Appendix B.3(m) - Table 54).

Table 18: Summary data and GLM results for peak accelerations ($m.s^{-2}$) for flexion and extension at the start and the end of the fatigue protocol (means ±standard deviation; coefficient of variation in percent).

		First 4 cycles	Last 4 cycles	p-values		
				Time	Movement (flex/ext)	Time x Movement
Peak Acceleration ($m.s^{-2}$)	Flexion	12.76(±3.69); 28.87%	14.15(±4.31); 30.42%	0.011	<0.001	0.092
	Extension	10.65(±2.39); 22.46%	10.81(±2.47); 22.89%			
Range ($m.s^{-2}$)	Flexion	4.73(±2.95); 62.37%	4.33(±2.48); 57.22%	0.557	<0.0001	0.706
	Extension	2.85(±1.86); 65.50%	2.75(±1.91); 69.49%			
Standard Deviation ($m.s^{-2}$)	Flexion	2.11(±1.28); 60.39%	1.92(±1.06); 55.04%	0.527	<0.0001	0.668
	Extension	1.27(±0.84); 66.06%	1.23(±0.87); 70.66%			
Coefficient of Variation (%)	Flexion	17.68(±11.59); 65.58%	13.92(±6.47); 46.47%	0.237	0.058	0.204
	Extension	12.07(±8.00); 66.29%	12.24(±11.03); 90.13%			

** Grey shaded cells indicate a significant effect at p<0.05*

The mean peak accelerations during the "Varying Length" condition revealed a significant time effect as peak accelerations recorded for the first four movement cycles increased significantly to those of the last four movement cycles of the fatigue protocol ($p<0.001$). A further significant difference was found between the two movement phases ($p<0.012$), with flexion yielding greater peak accelerations compared to extension. No interaction effect for time and movement was identified.

The GLM analyses for the cycle-to-cycle variability indicators, namely, range and standard deviation, only yielded statistical differences for the movement factor, with the variability in flexion accelerations being significantly greater than for extension (Appendix B.3(q) – Table 70 to Table 72). No time or interaction effects were identified for the cycle-to-cycle ranges and standard deviations. No time, movement or interaction effects were, however, found for the coefficient of variation, despite trending towards a significant movement effect.

5.6 RESPONSES TO HYPOTHESES

With the above analyses in mind, the following responses have been formulated to the hypotheses proposed in Section 4.2 Statistical Hypotheses.

5.6.1 Hypothesis 1: Effects of Time

This hypothesis tested whether the exercise protocol would affect (a) muscle fatigue development and (b) recovery from fatigue, as evidenced by changes in the fatigue parameters over time. Assessing the effects of time on fatigue indicators was considered to be a necessary manipulation check to confirm the presence of fatigue in order to compare fatigue responses between the experimental conditions relating to exertion type.

a) The null hypothesis focusing on time-related changes over the duration of the exercise protocol is tentatively rejected, and the alternative hypothesis is accepted since significant time-related deteriorations were found for peak torque, time to peak torque, the Fatigue Index, work, power, EMG amplitude (RMS), the EMG median frequency, and movement accelerations, thereby confirming the presence of muscle fatigue. Movement consistency measures revealed no significant time-related differences. Although the findings for performance accuracy yielded inconsistent outcomes, these were not in conflict with the alternative hypothesis.

b) The null hypothesis focusing on time-related changes after the fatigue protocol is also tentatively rejected, and the alternative hypothesis is accepted since significant improvements in peak torque, the Fatigue Index, time to peak torque, work and power confirmed the recovery from muscle fatigue.

5.6.2 Hypothesis 2: Effects of Exertion Type

This hypothesis proposed that fatigue and recovery could be affected by (a) variations in muscle length or (b) variations in muscle force when compared to an exertion exhibiting neither length nor force changes.

a) The null hypothesis for the effects of muscle length on the magnitude of muscle fatigue and recovery is tentatively rejected, and therefore the alternative hypothesis tentatively accepted for endurance time, number of cycles, peak torque, time to peak torque, the Fatigue Index, power, EMG median frequency, as well as performance accuracy and cycle-to-cycle variability relating to force. However, varying muscle length was found not to significantly affect work and EMG amplitude. Thus, the null hypothesis is tentatively accepted for these parameters.

b) No significant differences in fatigue responses between the “Pure Static” and “Varying Force” led to the tentative acceptance of the null hypothesis and, thus, the rejection of the alternative hypothesis. Exertions with varying force levels did not change the cycle time, number of cycles, peak torque, time to peak torque, the Fatigue Index, EMG amplitude and median frequency, cycle-to-cycle variability and most performance accuracy measures, except the deviation from the target joint angle.

CHAPTER 6: DISCUSSION

The aim of this dissertation was to investigate the gaps identified in the literature pertaining to dynamic exertions under submaximal conditions. More specifically, the study focussed on two task-dependent factors believed to influence the fatigue process, namely muscle length and the force generated by the muscle. This discussion is divided into three main parts: first, a consideration of the participant sample, including individual factors that may have influenced fatigue responses. This is followed by a discussion around the changes in fatigue responses over time to verify the fatiguing nature of the exercise protocol. Finally, the fatigue responses to the different exertion types ("Varying Length" and "Varying Force") are discussed relative to the baseline responses of the "Pure Static" exertion.

6.1 STUDY SAMPLE

Individuals who participated in the current study were a relatively homogenous sample of young adults who, on average, were 21.83(±1.80) years of age. They were considered healthy since none had reported any health issues, particularly relating to musculoskeletal injuries or disorders, and were classified as "normal" according to their Body Mass Index (BMI) values. It is, however, acknowledged that BMI is only a crude indicator of health since it only uses height and weight in its formula, with no differentiation of fat and fat-free mass, and has been shown to vary between population groups (Weir & Jan, 2023). Furthermore, participants were only included in the study if they regularly engaged in physical activity, although the amount of physical activity was not recorded or controlled for.

The sample consisted of an equal number of male and female participants, and this presented as a major source of variability in the responses recorded since it is well known that, on average, males are significantly taller and stronger than females, mainly due to their greater testosterone secretion (Bishop *et al.*, 1987; Miller *et al.*, 1993). In the current study, the male participants were 7% taller and 22% heavier than their female counterparts. On average, the males also generated twice the amount of force for all three exertion types during the MVEs compared to the females (91.9%, 105.7% and 105.2% greater for isometric, concentric and eccentric exertions, respectively). These findings align with numerous studies (e.g., Maughan *et al.*, 1983; Bishop *et al.*, 1987; Miller *et al.*, 1993), which found muscle strength between 58% to 172% greater in males than in females. Given the significant difference in the size and force-producing capacity between the sexes, the loads manipulated during the fatigue protocol were relativised to each individual's maximum torque, and peak torque was normalized to body weight. While this may not have completely eliminated any sex-related differences, the researcher was confident that it was sufficient to not significantly influence the outcomes of the study, as statistical analyses did not show any sex-related effects on endurance time, number of cycles completed and the Fatigue Index (Appendix B.1(f-h); Table 25 to Table 27).

6.2 EFFECTS OF TIME ON LOCALISED MUSCLE FATIGUE

Although the purpose of the current study was to determine differences in muscle fatigue development during different muscle exertions, it first had to be established that the protocol itself was indeed effective in inducing muscle fatigue. The mechanisms underlying fatigue are, however, multifaceted and complex due to a variety of physiological and psychological contributors (De Luca, 1984). It is for this reason that obtaining an accurate impression of localised muscle fatigue requires the use of several measurement methods (Seghers & Spaepen, 2004; Madeleine, 2010). The current study, therefore, made use of various variables relating to endurance time, maximum strength, muscle activity, performance accuracy and variability, as well as perceived exertion at various points throughout the experiment. Although it is common practice in fatigue research to interrupt a fatiguing activity with brief maximal exertions to determine the decline in the maximal force capacity at specific points in time during the exercise (Enoka & Duchateau, 2008), Carroll *et al.* (2017) highlighted that despite these maximal measures providing information on neuromuscular function that is valid and easy to interpret, they are not sensitive to some physiological changes that occur during the fatigue process. Furthermore, interrupting a submaximal exercise protocol with maximal exertions may influence fatigue development, which, according to the principle of task-dependency, has been speculated to be different between maximal and submaximal exertions (Enoka, 1995; St Clair Gibson *et al.*, 2001). Therefore, maximal exertions were only performed prior to and upon termination of the submaximal exercise protocol, as well as during the recovery period. The fatigue indicators obtained during these maximum voluntary exertions included peak torque, time to peak torque, work, and power. Further measurements were obtained during the fatigue protocol while performing submaximal exertions and included EMG and measures of performance accuracy and cycle-to-cycle variability. Additionally, endurance time and number of movement/force cycles were recorded, which, together with the aforementioned variables, would provide a comprehensive picture of localised muscle fatigue development.

Considering the responses of various fatigue indicators, it can confidently be argued that the exercise protocol for all three experimental conditions resulted in localised muscle fatigue of the forearm flexors. This conclusion was derived initially since all participants terminated the experimental conditions when they rated the muscular effort of the involved elbow flexor muscles as "very, very hard" (i.e., RPE of 9 on the Borg CR10 scale). Protocol durations during fatigue studies are either determined by a set time or by the point of volitional exhaustion (Yassierli & Nussbaum, 2009; Dickerson *et al.*, 2015). Given the anticipated sizeable inter-individual variability due to various physical capacities, as well as between different experimental conditions, setting a fixed protocol duration for the current study was not considered appropriate. Instead, the protocol duration was set to the point at which each participant reached a local RPE of 9, as it assured fatigue and yet avoided complete task failure, which accompanies exhaustion. A study by Dickerson *et al.* (2015), for example, terminated their experiment when participants indicated they no longer had the physical capacity to continue or when they reached a rating of 7 on the CR10 scale. To determine the point of muscular near-exhaustion, participants in the current study were asked to focus on the amount of effort invested by the working muscles and to rate this localised effort at 30-second intervals. Active verbal encouragement of participants aimed at reducing the influence that lack of motivation may have had on perceived exertion, although it is acknowledged that the influence of motivation on (near) maximal exercise performance is multifaceted, complex, and has had variable findings in the literature (e.g., Hulleman *et al.*, 2007; Dias Neto, 2015; Behrens *et al.*, 2023).

A more objective indicator of localised muscle fatigue popular in the literature is a reduction in maximum force-producing capacity (Bigland-Ritchie & Woods, 1984; Gandevia, 2001; Paillard, 2012; Dickerson *et al.*, 2015; Behrens *et al.*, 2023). When comparing the peak torque responses prior to and after the fatigue protocol (i.e., “Pre” vs “Post-Min 0” recordings) in this study, significant decreases in peak torque were found for all exertion types, irrespective of the measurement unit. Peak torque values obtained post-protocol decreased between 15.85% and 16.88% for the isometric conditions (i.e., “Varying Force” and “Pure Static”, respectively) and between 22.84% and 33.30% for the eccentric and concentric phases of the “Varying Length” condition, respectively. A range of values is cited in the literature regarding the most reliable indication of fatigue with regard to force reduction. For example, Schwendner *et al.* (1995) indicated a peak quadriceps force of less than 50% of MVE to be a reliable indicator of muscle fatigue, while Maquet *et al.* (2004) highlighted a coefficient of variation of more than 11% of the elbow flexors, but also emphasized that comparisons of strength performance are near impossible since strength is influenced by a range of factors, including the limb’s arrangement, the participant’s position, the range of motion and test velocity. In addition, natural variability in strength expression occurs throughout the day due to the circadian rhythm, resulting in significant variability in maximal performance throughout the course of a day (Wyse *et al.*, 1994; Knaier *et al.*, 2019). It is for this reason that across testing days, participants were tested within a 2-hour window period to minimise the influence that time-of-day may have on the strength performance within individuals. Rainoldi *et al.* (2001) indicated that a normal day-by-day variation in maximum isometric strength was around 10%, while Knaier *et al.* (2019) found day-to-day variations for elbow flexion of 3.14%, 3.64%, and 5.03% for isometric, concentric and eccentric exertions, respectively. The same authors reported diurnal variations ranging between 12.6% and 13.2%. The changes in peak torque responses obtained in the current study exceeded these thresholds, thus indicating that the participants did, in all likelihood, experience localised muscle fatigue under all conditions as a result of the exercise protocol.

Along with time-related reductions in peak torque, it was expected that the time to peak torque would increase from the start to the end of the fatigue protocol, since increases in muscle fatigue are said to slow down a muscle’s contractile ability, thereby increasing the time it takes the muscle to reach its maximum torque (Hunter *et al.*, 2004; D’Emanuele *et al.*, 2021). However, no significant change was found in the time it took for a muscle exertion to reach peak torque in this study. Motor unit synchronization, i.e., the concurrent discharge of several motor units at the same time, is known to occur with fatigue development (Naeije & Zorn, 1982). The adjustment in motor unit recruitment could have enabled the muscle fibres to still generate fast contractions and thus reach their peak torque without any time delays. Another explanation for this finding could be that a compensatory mechanism prioritized contractile speeds over peak torque, thus achieving similar times to peak torque, but at a lower force output.

Given the close relationship that work and power have with peak torque (ACSM, 2006), it was not surprising that these measures, too, decreased with increasing muscular fatigue (Enoka & Duchateau, 2008; Senefeld *et al.*, 2013). These variables revealed significant time effects ($p<0.0001$), with both work and power showing significant decrements from the “Pre” to the “Post-Min 0” measurements of the exercise protocol. These parameters could, however, only be obtained for the “Varying Length” condition for the concentric and eccentric movement phases since mechanical work is obtained by multiplying the force generated by the displacement caused by the force ($W = F \times d$) (McLester & St Pierre, 2008), and can thus only be calculated for exertions that display movement. In the current study, displacement only occurred during the isokinetic (i.e., Varying Length), but not the isometric (i.e., Pure Static and Varying Force) conditions due to the absence of movement under those conditions. Similarly, power could not be calculated for the conditions with an isometric component, since power is defined as work done within a particular unit of time (i.e., $P = W / \Delta t$) (McLester & St Pierre, 2008). In this study, the data for work obtained during the concentric and eccentric phases of the “Varying Length” condition resulted in reductions over time of between 34.66% and 36.38%, respectively, while power decreased between 32.43% and 42.50%, respectively. These decreases in work and power can be attributed to a reduction in the ability to maintain full activation during voluntary exertions (James *et al.*, 1995). The work done is influenced by the amount of force produced over a given time, while power is dependent on force and velocity, both of which can change with muscle fatigue (Krüger *et al.*, 2019). However, since the movement speed of the isokinetic dynamometer was set to a constant $60^{o}.sec^{-1}$, and the time to peak torque did not change significantly from the “Pre” to the “Post-Min 0” intervals, the reductions in the amount of work done and power generated can only be explained by the reductions in peak torque. This is corroborated by the significant and strong correlations (ranging between $r=0.84$ and $r= 0.93$) between the peak torque and the work and power values. These decreases in work and power can be attributed to a reduction in the ability to maintain full activation during voluntary exertions (James *et al.*, 1995).

Similar to work and power, the Fatigue Index of the maximum voluntary exertions, calculated from the peak torques obtained prior to and immediately after the fatiguing exercise, indicated that all exertion types experienced fatigue since their values were positive, and ranged between 15.6% and 33.6%. A larger value for the Fatigue Index is indicative of a greater decline in the force an individual is able to generate over the duration of an exercise protocol, or, as Morris *et al.* (2008) pointed out, a larger Fatigue Index shows lower fatigue resistance. While none of Ditor (1999), Morris *et al.* (2008), or Carpenter *et al.* (1998) specified a threshold for the Fatigue Index, their Fatigue Index values ranged between 37.8(±14.1)% and 44.7(±10.5)% for young adults, and even averaged as high as 46% for shoulder rotation in the study by Carpenter *et al.* (1998). The highest Fatigue Index for the current study was 33.59(±18.70)% for the "Varying Length–Concentric" condition, while the Fatigue Index values for the other conditions were 21.43(±12.88)% at most. One reason for the differences in the Fatigue Index between the studies may be the use of different body segments or muscles (e.g., the shoulder in the Carpenter *et al.* (1998) study, the adductor pollicis by Ditor (1999) and the quadriceps muscle group by Morris *et al.* (2008)). The investigation by Seghers and Spaepen (2004) conducted a fatigue protocol on the biceps brachii muscle; however, the Fatigue Index values are not directly comparable due to a slight difference in the formula used for calculating the fatigue index. Despite this, Seghers and Spaepen (2004) indicated that a Fatigue Index of 15.2% and 15.3% was enough to indicate fatigue in the biceps brachii muscle.

While variables such as torque, work and power represent the outcomes of muscle fatigue, they reveal little about the neurophysiological processes that occur during the fatiguing exercise. The electromyographical signal has long been used in the understanding of muscular functioning (Kumar, 2006; Al-Mulla *et al.*, 2011), even though it comes with limitations in the determination of muscle fatigue (Dimitrova & Dimitrov, 2003) and should therefore be used alongside other fatigue indicators (De Luca, 1984). Distinct changes in both the EMG signal amplitude (RMS) and frequency spectrum from the start to the end of an exercise protocol are the two most common approaches to determining the myoelectrical signs of fatigue (Enoka & Duchateau, 2008), as they reflect the increased effort via adjustments to the temporal and spatial recruitment of motor units to maintain the task requirements. In the current study, comparisons of the EMG(RMS) of the first four cycles with the last four cycles of the exercise protocol revealed significant increases over time for all exertion types ($p<0.0001$), with the amplitude increasing between 46.90% and 72.69% for the different exertions, thus confirming fatigue (Nur *et al.*, 2015). At the peripheral level, motor unit recruitment consists of a complex interplay between spatial and temporal recruitment, both of which adapt as the muscle adjusts to numerous mechanical and chemical changes (Sjøgaard *et al.*, 1988). As a fatiguing submaximal exercise progresses, various peripheral metabolic factors inhibit the muscle fibres' ability to generate force (Boyas & Guével, 2011). To maintain the required motor output, the motor units' excitation rates increase, and additional motor units are recruited to maintain the required force output. It is this adjustment in the motor unit recruitment and rate coding that shows up as an increase in the EMG amplitude (Chaffin, 1973; Bigland-Ritchie & Woods, 1984; Seghers & Spaepen, 2004; Kamen & Gabriel, 2010; Vigotsky *et al.*, 2018). The recruitment of additional motor units, or the rotation of fatigued motor units with unfatigued ones, is only possible during submaximal activities since not all motor units are recruited during the exercise, and there is a pool of unfatigued motor units available to replace the fatigued ones (Enoka & Duchateau, 2008; Al-Mulla *et al.*, 2011). Kapitaniak (2001b) found that the electrical activity associated with this recruitment strategy increases linearly with greater force exertion up to a force level of 50% of maximum voluntary exertion, after which the EMG signal increases to a greater extent than the increases in force production. However, the forces exerted during the current study remained considerably below this threshold, with the maximum forces exerted reaching 35% of MVE.

A further indicator of muscle fatigue in the current study was the significant reduction of the median frequency values of the power spectrum between 7.6% and 25.2%. It is commonly accepted that reductions in mean and median power frequency during exercise are indicative of muscle fatigue (Hagberg, 1981; Naeije & Zorn, 1982; Sun *et al.*, 2022), and considering the frequency domain in conjunction with the time domain is said to improve the reliability of the outcomes of a muscle fatigue assessment (Yousif *et al.*, 2019). Only median frequencies were presented in the current study since the analyses showed similar results to those of the mean frequencies.

Furthermore, Sun *et al.* (2022) pointed out that although both mean and median frequency can be used to identify the reductions in conduction velocity, it is the median frequency that is more sensitive to indicating changes in a muscle's functional state. The analyses of the current study's EMG data indicated that MdPF decreased between 56.08Hz and 77.34Hz for the different exertion types at the start of the fatiguing exercise to between 41.16Hz and 70.79Hz, resulting in decrements of between 5.6% and 20.09%. Öberg *et al.* (1990) found that the normal variation in the spectral frequency of an unfatigued muscle was around 8%. They concluded this from tests on the trapezius muscle as the arm moved through a functional range of motion along two movement planes with and without a load. An experiment by Naeije and Zorn (1982) revealed a 45.22% decrease in isometric exertions of the biceps brachii. However, their load level was set at 50% of maximum voluntary exertion, whereas the current study's exertion levels only averaged 25% of MVE, with a range of 15-35% MVE. A study by Nagata *et al.* (1990) used a more comparable exertion level, namely 20% of maximum voluntary isometric exertion (MVIE) and also found significant differences in pre- vs post-fatigue MdPF. Furthermore, Nagata *et al.* (1990) reported that higher exertion levels increased the magnitude of the shift to lower frequencies. Their study revealed a shift to the left on the power spectrum for biceps brachii of 8.6% at a level of 20% MVIE and 20.05% at a level of 80% MVIE. The decrease in the EMG power spectrum with muscle fatigue is said to be the result of several mechanisms, including increased synchronization in motor unit firing (De Luca *et al.*, 1986; Farina *et al.*, 2002), recruitment of larger (low frequency) motor units to replace the smaller fatigued unit (Fallentin *et al.*, 1993; Bawa *et al.*, 2006), and the reduction in the nerve impulse conduction velocity (Zwarts, 1987; Vollestadt, 1997; Farina *et al.*, 2002).

Since motor unit recruitment changes with the onset of fatigue, it is reasonable to assume that the accuracy and consistency of the task performance would also change as fatigue progresses due to decrements in motor control (Cortes *et al.*, 2014; Gates & Dingwell, 2011). Kumar *et al.* (2017) describe "movement accuracy" as the closeness of the movement to an intended motion, whereas "movement precision" is the consistency or repeatability of several cycles of movement, irrespective of whether the movement is accurate. In the current study, various kinematic indicators were used to determine any time-related changes in the accuracy of the motor performance, i.e., whether participants maintained the required performance output prior to and at the end of the fatigue protocol, as well as the consistency of the motor performance. However, given the lack of overt movement under the “Varying Force” and “Pure Static” conditions, the terms “performance accuracy” and “cycle-to-cycle variability” are considered more appropriate.

For the “Varying Length” condition, the performance objectives communicated to participants were: 1) to maintain the takt time of 2.6 seconds for one repetition of moving the forearm through a given range of motion, and 2) to reach the set endpoints of 40° elbow flexion and 40° elbow extension on either side of the 90° elbow flexion angle, thereby moving through a total range of motion of 80°. For the “Varying Force” condition, on the other hand, the objectives were to cycle between the specified minimum and maximum force measures at the set takt time, while maintaining an elbow flexion angle at 90°. The “Pure Static” condition merely required participants to exert a set force with the elbow flexion angle at 90°.

Performance accuracy for the three conditions of the current study yielded varying outcomes. Under the “Varying Length” condition, the timing of the movement cycles did not show any significant changes in the mean cycle time from the start to the end of the protocol, which did not align with expectations since the onset of fatigue has been reported to cause a decline in a muscle’s force-generating capacity (Iridiastadi & Nussbaum, 2006) and slowing down of contractile speeds (Bigland-Ritchie & Woods, 1984). This finding may be explained by the lack of change in the time to peak torque. However, the mean cycle time remained unchanged under the “Varying Length” condition despite significant changes in other movement characteristics, such as reductions in the overall range of motion, as well as increases in the deviations from the target force and increases in peak accelerations. Similarly, under the “Varying Force” condition, no significant time effects were found for the mean cycle time, but also not for the deviations from the target force. However, the mean deviation in elbow flexion angle from 90° did reveal significant reductions over the duration of the fatigue protocol, thereby deviating from the task objectives. Under the “Pure Static” condition, overall accuracy also decreased from the start to the end of the fatigue protocol, but with a significant reduction in the deviation in the mean elbow joint angle from the target angle and a significant increase in the deviation from the target force.

Muscle fatigue is not only characterised by a decreased force-generating capacity but also by the inability to maintain task performance (Sharpe & Miles, 1993; Jaric *et al.*, 1999). The deviations from the set task objectives under the three experimental conditions can, therefore, be considered further evidence that the exercise protocol induced localised muscle fatigue in the elbow flexors. The unexpected findings that some variables indicated time-related changes in their performance outcomes under different exertions, while others did not, can be explained by a trade-off between different parameters in an attempt to still achieve at least some of the performance objectives. An experiment by Gates and Dingwell (2008) showed trade-offs in movement characteristics, yet still maintained the required performance output, albeit in different variables. They reported no significant changes in average movement distance, movement speed, and timing errors during a fatiguing sawing task, but found significant reductions in the cycle-to-cycle consistency of movement speed and timing errors, which did suggest faster corrections of performance errors when fatigued. Cheng & Rice (2005) observed significant reductions in the range of motion, which they attributed to the reduced force-generating capacity. A study by Côté *et al.* (2002) also found no changes in the cycle time or trajectory amplitude with fatigue development of their sawing task, but significant decreases in elbow motion amplitude. These authors, too, concluded that adjustments in certain movement parameters (in their case, elbow trajectory amplitude and force output) allowed other spatial and temporal motion characteristics, such as endpoint trajectory and cycle time, to be maintained. Similarly, Selen *et al.* (2007) found no significant deviations in the performance of a tracking task but did find significant changes in other parameters, which led these authors to conclude that adjustments in the motor control strategy had occurred.

Gates and Dingwell (2008) concluded that there were adjustments of the "biomechanical movement patterns in response to fatigue only in such a way that they specifically preserved the goal-relevant features of their motor performance" (p.10). For example, Hunter *et al.* (2002) found differing responses when participants were asked to focus on maintaining a target force at 15% of their MVE versus maintaining a set joint angle but at the same force. Under the "Varying Length" condition, for example, the task objective of maintaining a set movement cycle time (as had been emphasised by the researcher to the study participants) appears to have been prioritized above reaching the set endpoints in the range of motion, thus resulting in a significant reduction in the overall range of motion and mean forces exerted, and which would support the assumption of increasing neuromuscular fatigue. Contrarily, the significant increases in the peak accelerations over time appear counterintuitive initially; however, it can be speculated that these were attempts to periodically increase movement velocities in an attempt to maintain the set cycle time. A similar argument can explain findings for the “Pure Static” condition, namely that the objective of maintaining a consistent joint angle was prioritized as neuromuscular fatigue set in, as deduced from the decreased deviations from the 90^{o} elbow flexion angle, but at the expense of maintaining the required force. In their paper, Gates & Dingwell (2008) further stated that it was not clear which variables specifically were being controlled. Based on the results obtained from the current study, it could be speculated that movement control is either individual- or task-dependent, and even that adjustments in movement strategies vary as fatigue progresses. These speculations are, however, outside the scope of this dissertation.

Even though similar trade-offs could have occurred under the “Varying Force” condition, the findings from the variables recorded are less clear-cut. Under the “Varying Force” condition, no significant time-related changes were found for cycle time, nor the deviations in the force produced, yet performance accuracy improved with fatigue as indicated by the reductions in the mean deviations from the target elbow flexion angle. Selen *et al.* (2007) suggested that the predictability of a task may be the reason why target performance is maintained. Alternatively, studies by Vuillerme *et al.* (2001) and Melnikov *et al.* (2016) speak to the importance of visual input in regulating movement performance, arguing that any fatigue-related reductions in movement control could be compensated by “guidance of visual feedback” (Huysmans *et al.*, 2008). The experimental set-up of the current study entailed providing visual feedback to assist participants in achieving the required joint angles (for the “Varying Length” condition) and forces (for the “Pure Static” and “Varying Force” conditions). Similarly, auditory feedback from the metronome with reinforcement from the researcher could have achieved the same effect for maintaining cycle time. Bove *et al.* (2007) did point out that spatial and timing accuracy during rhythmic and repetitive movements are dependent on the external signals (e.g., auditory or visual) as well as the movement rate. These authors found reductions in the number of tapping errors with externally paced movement, which they attributed to a reduction in the time allocated for sensory information processing and adjustments in the movement strategy.

The absence of a consistent pattern in the outcome parameters under the three experimental conditions may speak to underlying shifts in the neuromuscular strategies as muscle fatigue progresses. This is, however, contested in the literature where, for example, Jaric *et al.* (1999) found distinct performance decrements with muscle fatigue, while other studies (e.g., Côté *et al.*, 2002; Selen *et al.*, 2007) revealed no significant changes in performance. It is therefore assumed that neural control and coordination strategies not only change between subjects (Walsh *et al.*, 2004) but even within subjects as fatigue progresses (Gates & Dingwell, 2008). Bove *et al.* (2007) found that different neural strategies are used for sensorimotor processing when the movement rate and sequence complexity are increased due to the reduced time available for information processing. Changes in performance outcomes of repetitive tasks with muscle fatigue must constantly be evaluated against the set criteria and adjusted as fatigue progresses, but may not always be successful, therefore resulting in further performance decrements. However, Selen *et al.* (2007) also suggested that a feedforward strategy is employed by the neural system to remain close to the target centre during a tracking task.

Quantifying the consistency of repetitive movements could be a means of gaining insights into these neuromuscular strategies. In the current study, cycle-to-cycle variability for the "Varying Length" condition was determined by comparing the variability measured during the first four repetitions of the fatigue protocol with that obtained from the last four cycles. These measures included range, standard deviation, and coefficient of variance for the mean cycle times, forces applied, ranges of motion and joint angle endpoints, as well as accelerations. However, given the differences in movement characteristics between the experimental conditions, cycle-to-cycle variability for the "Varying Force" condition was only analysed for cycle time, force fluctuations and deviation from the 90º elbow flexion angle, while for the "Pure Static" condition, the analyses of variability were limited to force and joint angles.

Motor variability, or sensory-motor noise, is an inevitable phenomenon of voluntary muscle actions (Hamilton *et al.*, 2004). Regarding the effects of muscle fatigue on cycle-to-cycle variability, numerous studies have found increases in the variability (i.e., decreases in consistency) of kinetic and kinematic parameters during repetitive movements, such as force (Yao *et al.*, 2000; Contessa *et al.*, 2009; Missenard *et al.*, 2009), endpoint positions / joint angles (Gates & Dingwell, 2011; Vafadar *et al.*, 2012), and accuracy (Selen *et al.*, 2007). Harris & Wolpert (1998) pointed out that the strength and quality of the sensory signal directly influence motor planning. Poor-quality signals, in other words, those with a large amount of 'noise', are magnified as motor command signals increase, such as during larger or faster movements. Muscle fatigue is another reason why sensory-motor noise may be amplified and can thus increase the variability of the motor output (Contessa *et al.*, 2009). This increased noise can be attributed to reduced proprioception (i.e., poor quality signal) (Voight *et al.*, 1996) as well as changes in the motor unit recruitment patterns, as more motor units are recruited (Potvin & Bent, 1997), or fatigued motor units are exchanged for unfatigued ones (Sjøgaard & Jensen, 2006). It has also been proposed that motor unit synchronization, i.e., the simultaneous discharge of numerous motor units in an attempt to increase force production and which increases with fatigue, could be a reason for reduced force steadiness (Yao *et al.*, 2000; Semmler, 2002; Gates & Dingwell, 2011).

However, in the current study, movement variability data for cycle time, force, and goniometry data remained mostly unchanged over the duration of the fatigue protocol, with the only exception being the coefficient of variation for the movement endpoints in the range of motion under the "Varying Length" condition, which showed a significant time-related reduction. The absence of any time-related changes contradicted the researcher's expectation that movements would become more erratic and uncontrolled as fatigue progressed and would thus result in reduced motion consistency. A trade-off between performance accuracy and motor consistency could again be an explanation for this outcome. In the current study, it was emphasized to participants that they should maintain the set takt time, as well as reach the required joint angles (for the "Varying Length" condition) or forces (for the "Pure Static" and "Varying Force" conditions). External visual and auditory cues may have aided participants in maintaining regular motions from one cycle to the next and across the fatigue protocol (Bove *et al.*, 2007), but at the expense of performance accuracy, hence the increased deviations in force and joint angles from their respective targets observed during the current experiment. However, Cortes *et al.* (2014) suggested that fatigue may have a differential effect on the motor variability of kinetic and kinematic parameters, as their study yielded increases in performance variability for sample entropy (indicating increased motor variability) during a side-stepping task, but decreased variability of the knee flexion moments and ground reaction forces. Adjustments in the neuromuscular coordination strategies and muscle activation patterns under fatiguing conditions could therefore be an attempt to minimize changes in overall kinematic variability to maintain task performance (Cortes *et al.* 2014; Coté *et al.*, 2002; Gorelick *et al.* 2003; Fuller *et al.*, 2011). Altered coordination strategies may also entail differential recruitment of the synergistic muscles. In a study by Gates & Dingwell (2008), participants performed repetitive pushing and pulling actions, simulating a sawing task, until exhaustion. Even though movement distance, speed and deviations remained relatively unchanged at the end of the protocol, the variability of the timing errors decreased. The authors concluded that participants may have employed various compensatory mechanisms to meet the task requirements. Using different muscles involved in the overall movement may have enabled them to maintain the required task performance, but without the overt increase in movement variability that was expected. In the current study, co-contraction of the three elbow flexor muscles, namely Biceps Brachii, Brachialis and Brachioradialis (Tortora & Derrickson, 2009), contributed to the force generation required to maintain the stipulated elbow posture or movement. With the onset of fatigue, the relative contributions of each of these muscles may, however, have changed in an attempt to maintain task performance, which may have masked the effects of fatigue on movement variability. Solomonow (2003) also claimed that with fatigue progression, there is an increased need for coactivation to maintain joint stability, thus reducing motor variability. Furthermore, Taylor *et al.* (2016) pointed out that there are multiple neural alterations that are associated with exercise-induced localised muscle fatigue and that each presents with its own movement and performance characteristics.

In addition to task-related influences, individual factors can also impact motor physiological processes that manifest in variability (Srinivasan & Matthiassen, 2012; Gaudez *et al.*, 2016), including age and gender (Svendsen & Madeleine, 2010), experience (Srinivasan & Matthiassen, 2012), and pain (Madeleine, 2010), amongst others. Even though the current study's sample was considered to be a relatively homogeneous group of individuals in terms of age and health, factors such as sex, the type of physical activity accustomed to, as well as their responses to the muscular discomfort experienced towards the end of the fatigue protocol, could have resulted in different movement strategies, and which could have masked the effects of fatigue. The use of changes in performance accuracy and cycle-to-cycle variability may, therefore, not be a reliable indicator of localised muscle fatigue.

Time-related Changes During Recovery

While a reduction in a muscle's force-generating ability over time is a valid indicator for exercise-induced muscle fatigue, so is the return to its original force-producing capacity (Schwendner *et al.*, 1995). In the current study, peak torque, work and power showed significant improvements over the five-minute recovery period after termination of the exercise protocol, with final peak torque values (i.e., at Post-Min 5) returning to between 77.80% and 89.87% of their pre-fatigue values. The Fatigue Index, too, revealed a significant drop from the end of the fatigue protocol to the 5-minute recovery mark. Schwendner *et al.* (1995) referred to an 80% recovery in neuromuscular force-generating potential as full recovery; however, the analyses of the current study's data still showed a significant difference in the peak forces at the end of the recovery period compared to the pre-fatigue values for all exertions. Similarly, time to peak torque, work, and power at Post-Min 5 also showed significant improvements compared to Post-Min 0 values, but while the time to peak torque had returned beyond its pre-fatigue levels, work and power measures remained significantly lower than those obtained prior to the exercise protocol. Recovery follows a logarithmic curve, with rapid improvements occurring at the start of the recovery period, which decreases over time (Lind, 1959; Yates *et al.*, 1987). While Lind (1959) defined two phases of the recovery process, Yates *et al.* (1987) proposed that recovery has three components, each of which can be attributed to the recovery of different central or peripheral processes: a first phase during which recovery occurs very fast, a slightly slower second phase, and the third, but slowest, phase. This logarithmic curve is evident from their data, which showed a 35% recovery in the biceps brachii in the first 30 seconds of rest, 50% recovery after 2:15 minutes, 75% after 7 minutes and just less than 90% after 20 minutes of recovery. The rate of recovery in the current study also appeared to vary between parameters, with peak torque, work and power recovering by between 78% and 90% after five minutes of rest, while the time to peak torque even exceeded its pre-fatigue values. This appears to vary somewhat between studies though; for instance, Cheng & Rice (2005) found that maximum voluntary force had not recovered after ten minutes of rest, but power had recovered fully after five minutes. The different phases of recovery have been attributed to the reversal of various central and peripheral mechanisms that have led to muscle fatigue, including muscle temperature, perfusion, restoration of metabolites and removal of by-products, but also central activation influencing motor neuron recruitment, amongst others (Yates *et al.*, 1987; Cheng & Rice, 2005), and all of which have different time courses (Carroll *et al.*, 2017). In fact, Carroll *et al.* (2017) stated that the initial restoration of voluntary force after a sustained submaximal exertion could generally be attributed to central factors, such as the recovery of the neural drive, as well as cortical and spinal neuronal excitability, while peripheral factors, i.e., metabolic factors and those influencing the excitation-contraction coupling mechanism, were responsible for the slower and more delayed phase of recovery. Additionally, Allen *et al.* (2008) pointed out that recovery was dependent on the muscle involved and the way fatigue was induced, but the same general curve has been found in other studies (e.g., Lind, 1959; Cheng & Rice, 2005; Senefeld *et al.*, 2013).

In summary of this first section of the discussion, analyses of RPE and endurance time, as well as time-related changes in peak torque, the Fatigue Index, power, work, and EMG before, during and immediately after the exercise protocol, as well as significant changes in EMG amplitude and power frequency, clearly confirmed the presence of muscle fatigue. Although the results for performance accuracy and movement variability were unexpected, findings similar to those of the current study have been reported in the literature. Since the current understanding of neuromuscular control and movement strategies is still contested in the literature, the absence of or even decreases in time-related changes in the cycle-to-cycle variability parameters recorded in the current study can be explained by the fatigue-related changes found in other kinematic parameters. Further supporting the conclusion that the exercise protocol did indeed induce muscle fatigue is evidence presented by the data obtained from the recovery period, with peak torque, work and power increasing significantly from the termination of the exercise protocol to the fifth minute of rest. Similarly, the Fatigue Index indicated recovery from muscle fatigue, but via significantly reduced values.

6.3 EFFECTS OF EXERTION TYPE ON LOCALISED MUSCLE FATIGUE

This part of the discussion focuses on the question of whether combinations of task characteristics, specifically the changes in muscle length in the absence of varying force generation ("Varying Length" condition) or changes in muscle force in the absence of changing length ("Varying Force" condition), influence the development of localised muscle fatigue. Since the "Pure Static" condition displayed neither variations in length nor force, it was used as a baseline condition against which to compare the other two exertion types. The discussion on the effect of exertion type on muscle fatigue has been thus divided into two parts: 1) the effects of muscle length, and 2) the effects of muscle force on muscle fatigue and its recovery.

6.3.1 Effects of Muscle Length

To assess the effects of changes in muscle length on localised fatigue, relevant fatigue indicators obtained for the "Varying Length" condition were compared to those of the "Pure Static" condition during the maximal and sub-maximal protocols. The "Pure Static" condition had an isometric component, meaning there was no change in the muscle length, i.e., the elbow remained at a 90^{o} angle, and participants exerted a constant external load under this condition. Under the "Varying Length" condition, the external load manipulated also remained constant, but the muscle length changed by moving the forearm through a total range of motion of 80^{o}. Furthermore, since motor unit recruitment and metabolic processes are known to differ between dynamic and isometric exertions (Ortega *et al.*, 2015; Orantes-Gonzalez *et al.*, 2023), responses obtained at the end of the fatiguing exercise protocol were normalized to their pre-fatigue values. Due to the dynamic nature of the "Varying Length" condition and the lack of movement under the "Pure Static" condition, direct comparisons between these conditions could only occur for endurance time, peak torque, Fatigue Index, time to peak torque, EMG signal amplitude and median frequency, deviation from their force targets as well as variability from cycle to cycle to determine the impact of changing muscle length on fatigue responses.

All recorded variables indicate that fatigue was induced by the fatigue protocol for both the "Varying Length" and the "Pure Static" exertions. However, the "Varying Length" condition generally produced greater muscle fatigue in response to the exercise protocol compared to the "Pure Static" condition. The data collected under the "Varying

Length” condition revealed a significantly shorter mean endurance time, fewer number of cycles, significantly greater decrements in peak torque relative to the pre-fatigue level, a significantly greater Fatigue Index, relatively longer times to peak torque, significantly greater proportional decreases in the median frequency, as well as significantly larger proportional deviations from the target force. This is despite no differences in the changes in EMG amplitude and significantly lower cycle-to-cycle variability compared to the “Pure Static” condition.

These greater fatigue responses under the dynamic condition (i.e., “Varying Length”) were an unexpected finding since it was anticipated that dynamic exertions, i.e., those exhibiting changes in muscle length, would be more fatigue-resistant than static exertions, i.e., those without changes in length (e.g., Masuda *et al.*, 1999; Baxi *et al.*, 2017). Dynamic exertions have consistently been reported to result in a later onset of localised muscle fatigue and thus have longer endurance times compared to static efforts (Rohmert, 1960; Masuda *et al.*, 1999; Kay *et al.*, 2000). Differences in fatigue parameters between exertions are generally attributed to metabolic factors and the resultant mechanical interferences at the level of the neuromuscular junction (Sjøgaard *et al.*, 1988; Sjøgaard & Jensen, 2006). One dominant argument presented in the literature for differences between static and dynamic fatigue responses is that inhibition of blood flow can severely reduce the provision of oxygen and substrates necessary for the contractile process and prevent the removal of metabolites (Sjøgaard *et al.*, 1988; Noakes, 2000; Kapitaniak, 2001b; Behrens *et al.*, 2023), thereby resulting in lower fatigue resistance (Barcroft & Millen, 1939; Humphreys & Lind, 1963; Bigland-Ritchie & Woods, 1984; Petrofsky & Hendershot, 1984; Masuda *et al.*, 1999; Sjøgaard & Jensen, 2006; Wan *et al.*, 2017; Oranchuk *et al.*, 2020). The reduced removal of metabolic waste products increases acidosis and calcium ion concentration, which in turn are thought to slow down muscle fibre conduction velocity (Gonzalez-Izal *et al.*, 2014), resulting in a decrease in the total electromyographical (EMG) frequency spectrum of the muscle, which is a valid indicator of muscle fatigue (Zwarts *et al.*, 1987; Arendt-Nielsen & Mills, 1988). The following reasons are speculated to have led to the unexpected greater fatigue under the “Varying Length” condition compared to the “Pure Static" condition.

Differential loading between conditions

One important factor for restrictions in blood flow is the magnitude of the external load, which requires greater forces to be generated by the muscle to move it. Reduced endurance times and peak torque decrements are known to be inversely proportional to loading or the intensity of the muscle exertion (Rohmert, 1960; Hunter *et al.,* 2005; Sjøgaard & Jensen, 2006). More muscle fibres are activated under high-level exertions, increasing intramuscular pressure and resulting in greater circulatory occlusion (Hietanen, 1984; Petrofsky & Hendershot, 1984). In the current study, the external loads manipulated during each of the exertion types differed from one another due to varying force-producing capacities and could therefore be speculated to have led to greater changes in fatigue parameters under the "Varying Length" condition. However, the external loads moved under the "Varying Length" condition were, on average, 3.88kg, while the "Pure Static" condition had to resist a mean load of 4.21kg. Only moderate negative correlation coefficients between -0.43 and -0.46 were found. Therefore, differences in external loads between conditions cannot be used as a satisfactory explanation for the greater fatigue response during the dynamic exertion.

Insufficient loading to occlude blood flow

Another possibility for the greater fatigue responses under the "Varying Length" condition could be that the level of the external load did not significantly affect perfusion. For example, McNeil *et al.* (2015) found that complete occlusion occurred between 50-60%MVE, while Humphreys and Lind (1963) cited force levels of 20%MVE as restricting blood flow enough to interfere with metabolic removal, but this was also dependent on the type of muscle action. Contrarily, Kahn *et al.* (1998) found no changes in muscle oxygenation below 25% of isometric MVE, and Vedsted *et al.* (2006) found similar muscle tissue oxygenation between static and dynamic exertions at 20% MVE. Hagberg (1981) also pointed out that continuous dynamic exertions without any intervening rest breaks would result in endurance times similar to those of sustained static muscle actions. While at high levels the occlusion would have been a reasonable expectation and would have justified the inclusion of measures to quantify blood flow, this could not necessarily be assumed for low-level exertions. The selection of low force levels in the experiment was a conscious decision, not only because equipment to record blood flow was unavailable to the researcher, but also because they mimicked a variety of occupational tasks (Sjøgaard & Jensen, 2006).

Therefore, even if the force generation, which was assumed to be continuous and with equal contributions from all three synergists, had resulted in similar levels of occlusion for both the “Varying Length” and “Pure Static” conditions, it would still not explain the greater fatigue under the “Varying Length” condition.

Muscle length affects tortuosity

If the magnitude of the load cannot explain the greater fatigue under the “Varying Length” condition, then the change in muscle length may, as blood flow to the working muscles can be influenced by muscle length. Sjøgaard *et al.* (1988) pointed out that blood vessels are pliable and can, therefore, undergo changes in length, i.e., they can become stretched or ‘collapse’. Particularly at the extremes of the range of motion, capillaries become elongated, reducing the luminal diameter, which in turn increases the resistance to blood flow (Poole *et al.*, 1997). It is, therefore, plausible that, despite the low load manipulated by the muscle (i.e., 25%MVE), the perfusion rates during the dynamic exertions decreased to a greater extent compared to the static ones due to the reduced capillary diameter. As the demand for oxygen and substrates increased over the course of the exercise protocol, the “Varying Length” condition showed greater changes in the fatigue indicators.

Differential responses for concentric and eccentric muscle actions

The unexpected findings of the greater fatigue responses under the “Varying Length” condition necessitated further analyses by separating its concentric and eccentric movement phases, as there may have been a differential effect of the one or other isotonic muscle action on the recorded parameters and thus contribution to muscle fatigue under the “Varying Length” condition. Peak torque revealed noticeable differences between the concentric and eccentric phases of the dynamic movement, suggesting that the direction of the muscle length change may influence fatigue. Peak torque during the concentric movement phase decreased by 34% relative to its unfatigued values, but only by 17% and 21% for the “Varying Length-Eccentric” and “Pure Static” exertions, respectively; the latter two being significantly less than the “Varying Length–Concentric” condition, but no different between one another. The significantly greater Fatigue Index obtained during concentric exertions also implies greater force decrements, which was expected given its close association with peak torque (Ditor, 1999; Gentil *et al.*, 2017; Morris *et al.*, 2008). This finding concurs with those of various authors (e.g., Pasquet *et al.*, 2000; Gonzalez-Izal *et al.*, 2014; Nuzzo *et al.*, 2023), who also found greater strength decrements after concentric compared to eccentric actions, although Linnamo *et al.* (2000) found the contrary.

The greater strength decrements during the concentric movement phase of the "Varying Length" condition can be explained by the larger energy requirements of a shortening muscle action. The contractile process during a concentric exertion requires significant amounts of energy to form the cross-bridges and generate the powerstroke through which it creates tension (Tortora & Derrickson, 2009; Peñailillo *et al.*, 2017). Conversely, eccentric exertions are the most energy efficient. The protein titin allows for more stretch energy to be stored in eccentric exertions, thus requiring a larger force to break cross-bridges before lengthening (Freivalds, 2004; Hoppeler & Herzog, 2014; Hessel *et al.*, 2017), but at a lower metabolic cost (Peñailillo *et al.*, 2017). The differing metabolic costs of the two exertion types may, therefore, explain the greater fatigue under the "Varying Length–Concentric" condition.

Differences in relative workload

Another reason for the lower fatigue under the eccentric compared to the concentric exertion phase could be the relative workload that the elbow flexor muscles were exposed to. The workloads for the "Varying Length" and "Pure Static" conditions were assumed to be the same, i.e., 25% of their respective maximum voluntary exertions. However, this assumption relied on the peak torques obtained prior to the exercise protocol to be accurate reflections of their voluntary maximum.

It is well accepted in the literature that maximum strength is greatest under eccentric exertions, followed by isometric muscle actions, and with concentric exertions yielding the lowest force (Doss & Karpovich, 1965; Linnamo *et al.*, 2000; 2003; Pasquet *et al.*, 2000; Freivalds, 2004; Ducrocq *et al.*, 2023; Nuzzo *et al.*, 2023; Ruas *et al.*, 2024). For example, Doss & Karpovich (1965) found eccentric forces to be 13.5% greater than isometric forces, and 23% greater than concentric forces in their cohort of male participants. Similarly, results by Griffin (1987) showed the same order of maximum strength-generating ability of different exertions in females, but with concentric exertions being between 4% and 24% less than those obtained during isometric exertions, while maximum strength during eccentric muscle actions was only 0.3-10% greater than isometric strength, depending on movement velocity. The protein titin, which is found within the muscles' sarcomeres, acts like a spring that increases in stiffness with eccentric exertions and allows for more stretch energy to be stored during eccentric exertions. A larger force is required to break cross-bridges before lengthening (Freivalds, 2004; Hoppeler & Herzog, 2014; Herzog *et al.*, 2016; Hessel *et al.*, 2017), therefore resulting in the greatest maximal torque, despite lower EMG readings.

In the current study, however, peak torques recorded on the isokinetic dynamometer prior to the start of the exercise protocol did not correspond with findings from the literature. Although the concentric exertion resulted in the lowest peak torque, this was followed by the eccentric exertion with the next highest peak torque, while the isometric exertions resulted in the highest peak torques. However, only the concentric exertion was significantly different from the other two exertions. A search of the literature, however, found no studies that corresponded to the peak torque findings of the current research, except for Koekemoer (2022), whose peak torque was lowest for eccentric exertions, followed by isometric muscle actions and the highest for concentric exertions of the ankle dorsiflexors. Since the experimental set-ups of the current study and that of Koekemoer (2022) both made use of the Biodex isokinetic dynamometer, it is assumed that the reason for the discrepancies in the order of peak torque generation between the different exertion types lies with the unfamiliarity amongst participants of demonstrating maximum strength on the Biodex. Furthermore, a review by Duchateau and Baudry (2014) on the neural control of eccentric contractions revealed that it is difficult for untrained individuals to maximally activate muscles in eccentric exertions. Even though participants in the current study were habituated to the isokinetic dynamometer, and in particular the eccentric mode of this apparatus, involuntary inhibition of maximum eccentric force may still have occurred. Given the voluntary nature of the maximal strength protocol, it is suspected that the study participants did not produce their maximum force, possibly out of fear of injuring themselves or even damaging the equipment, despite active encouragement by the researcher. If this was indeed the case, then the load moved during the submaximal fatigue protocol that was less than the intended 25% MVE and could explain why the changes in the various fatigue parameters under the "Varying Length–Eccentric" exertion revealed similar responses to the "Pure Static" exertion and a lower fatigue than for the "Varying Length–Concentric" movement phase.

A methodological consideration explaining the significantly greater fatigue responses under the concentric action compared to the eccentric exertion is that the load each participant moved under the "Varying Length" condition was calculated as the mean of the concentric and eccentric peak torques. The assumption was made that the workload of both isotonic movement phases would average at 25% MVE. However, given the lower force-producing capacity of concentric compared to eccentric exertions, the forces that had to be generated by the elbow flexors during the exercise protocol would thus have exceeded 25% of their maximum force-producing capacity, while under the eccentric movement phase, the muscles would have exerted less. This could, therefore, have contributed to the significant differences in fatigue between the concentric and eccentric responses. More recent technological advancements have led to the development of a connected adaptive resistance exercise (CARE) machine that allows for active adjustment of resistance (Nuzzo *et al.*, 2023) without interrupting the exercise. While such a machine would have enabled the appropriate adjustment of the resistance to 25% MVE for the concentric and eccentric movement phases, this type of technology was unfortunately not available to the researcher and repetition of the current study using the CARE machine may be considered in future studies.

Finally, a constant external load does not translate to constant forces generated within the muscle, particularly during the "Varying Length" condition, where the length-tension relationship would have influenced the forces that had to be generated by the muscle during the concentric and eccentric movement phases to move the external load (Freivalds, 2004; Lieber & Ward, 2011). Therefore, the internal workload of the muscle, while assumed to have been 25% of MVE, may not have been the same across all three experimental conditions.

Differences in motor unit recruitment

EMG readings are a popular method to quantify muscle activity and fatigue progression. However, the EMG trace can be influenced by a variety of central and peripheral factors, including muscle temperature, conduction velocity, discharge rate and synchronization of motor units (Masuda *et al.*, 1999; Vigotsky *et al.*, 2018). Peripheral factors influencing fatigue development include metabolites, which slow down the muscle fibre conduction velocity and decrease the excitability of the muscle fibre membrane (Arendt-Nielsen & Mills, 1988; Zwarts *et al.*, 1987; Masuda *et al.*, 1999; Farina *et al.*, 2002), thereby decreasing the muscle's force-producing capacity. Contributions of central factors, however, decrease the motor unit firing rate but enhance motor unit synchronization (Bigland-Ritchie & Woods, 1984; Liu *et al*., 2021).

The results of the current study show that during both the "Varying Length" and "Pure Static" exertions, the EMG amplitude increased with the development of fatigue, which can be attributed to the additional recruitment of motor units (Enoka & Fuglevand, 2001; Contessa *et al.*, 2009). Masuda *et al.* (1999) found that the increments in EMG amplitude after a fatigue protocol were greater for dynamic muscle actions than for static ones. In the current study, the EMG amplitude of the "Varying Length" exertion rose to a greater extent than that of the "Pure Static" muscle action, but the difference between the two was not significant. This may, however, be explained by the high variance of the EMG(RMS) data (CV of "Varying Length" = 66.02% and CV of "Pure Static" = 105.91%), which could have masked any differences in the proportional changes in EMG amplitude between the two exertion types.

The median frequency in the current study, on the other hand, did decrease to a greater extent relative to its pre-fatigue values under the "Varying Length" condition when compared to the "Pure Static" condition. Masuda *et al.* (1999), however, found that by the end of the fatigue protocol, the decrements in the median frequency were greater during the static exercise (22.5%), compared to the dynamic exercise (15.2%).

Motor unit recruitment strategies are complex and vary between different exertion types (Komi & Tesch, 1979; Grabiner & Owings, 2002; Cheng & Rice, 2005; Morris *et al.*, 2008). James *et al.* (1995) attributed the differences between fatigue after static and dynamic exercise to limitations in maintaining the motor drive. To generate the appropriate amount of force, the temporal and spatial recruitment of motor units must be carefully adjusted under all muscle activations, but particularly under dynamic exertions, due to the muscle length changes and the transitions from lengthening to shortening, and vice versa (Chaffin *et al.*, 2006; Duchateau & Baudry, 2014; Enoka & Duchateau, 2017). Furthermore, complex adjustments occur in the motor unit recruitment patterns in response to muscle fatigue and under submaximal exertions; more motor units are first recruited to maintain the task performance, followed by adjustments in their firing rates (De Luca *et al.*, 1982; Moritani *et al.*, 1981; Enoka & Fuglevand, 2001; Contessa *et al.*, 2009; Boyas & Guével, 2011). It is acknowledged that several mechanisms falling under both the central and peripheral categories contribute to the overall fatigue outcome and possibly even in changing proportions as fatigue progresses and in an attempt to prevent task failure (Enoka & Duchateau, 2008; Fitts, 2008; Gandevia, 2001; Potvin & Bent, 1997; Vøllestad, 1997). The recruitment patterns may therefore adjust, depending on the task characteristics (Farina, 2006), which is probably the reason why the performance accuracy data showed such inconsistent results.

This complexity is the reason why Christensen *et al.* (1995) and Krüger *et al.* (2019) cautioned against comparing static and dynamic responses with the same indicators since the different types of exertions do not share the same physiological mechanisms and therefore "present divergent responses". It may also be the reason why limited studies have investigated localised muscle fatigue specifically under continuous submaximal static and continuous submaximal dynamic exertions.

Effect of muscle length variation on recovery

While all parameters showed time-related improvements over the course of the recovery period, not all parameters or exertion types returned to their pre-fatigue levels after five minutes of recovery. For example, final recovery peak torque values (i.e., at Post-Min 5) returned to between 85.68% and 89.87% of their pre-fatigue values for the "Varying Length-Eccentric" and "Pure Static" conditions, but with the "Varying Length-Concentric" condition only recovering to 77.80% of its pre-fatigue values. The final peak torque levels at the end of the recovery period remained significantly different between the three exertion types. Recovery of the work data also showed a shortfall of between 17% and 19%. However, other variables, such as time to peak torque or power for the concentric exertion, had returned to, or even exceeded, their pre-fatigue levels.

While the absolute difference from the starting values is one way of judging recovery from fatigue, the rate of recovery (i.e., Post-Min 5 compared to the Post-Min 0 responses) is arguably more important. In the current study, the fastest recovery rates for peak torque occurred under the concentric condition (14.29% relative to Post-Min 0 values), followed by the eccentric exertion (8.59%) and, finally, the "Pure Static" condition (8.07%). Studies by Linnamo *et al.* (2000) and Yates *et al.* (1987) reported similar findings in terms of the order of recovery rates. Furthermore, Yates *et* *al.* (1987) explained that faster recovery from dynamic exercise compared to static exercise was the result of less occluded blood vessels. Since blood vessels after a dynamic exertion were “more open”, this would have allowed faster reperfusion compared to those following a static activity, particularly during the early phase of recovery. Cheng & Rice (2005) and Carroll *et al.* (2017) speculated that central processes, such as the restoration of the cortical drive as well as spinal excitability, recover fastest and are, therefore, the dominant factor in the rapid first phase of recovery.

Contrary to peak torque, recovery rates for work and power in this study only revealed an 18-20% improvement for the concentric data, but between 28-30% for the eccentric data. A study by Cheng & Rice (2005), however, found that power had recovered by five minutes, but not for MVEs. The contradictory findings in terms of the recovery data are not surprising, as Yates *et al.* (1987) reported that rates of recovery for maximum strength depend on the type of exercise being performed and the parameter being assessed.

6.3.2 Effects of Force Level

To investigate the impact of changing force levels on localised fatigue, the responses obtained during the “Varying Force” condition were compared to those of the baseline “Pure Static” condition. While both exertion types could be described as isometric since the elbow flexion angle was maintained at 90 degrees, the force generated under the “Pure Static” condition remained a steady 25% of maximum voluntary isometric exertion (MVIE), while under the “Varying Force” condition, the forces exerted against an immovable object varied between 15% and 35%, thereby averaging 25%MVIE. As with the comparisons between the “Varying Length” and “Pure Static” exertion, the fatigue indicators recorded towards the end of the fatigue protocol, or just after its termination, were relativized to their pre-fatigue values.

Comparisons of the “Varying Force” to the “Pure Static” responses indicated no significant effect of force variations on cycle time, number of cycles, peak torque, time to peak torque, the Fatigue Index, EMG amplitude and median frequency, as well as cycle-to-cycle variability. The only significant difference found between these two exertion types was that the “Varying Force” condition displayed significantly greater deviations from the target angle than the “Pure Static” exertion, which was attributed to minor wrist movements that occurred as participants pulled on the pulley handle to vary between minimum and maximum forces, rather than muscle fatigue itself. The insignificant results between the two conditions were unexpected, as the impact of varying force levels was anticipated to have one of two outcomes: either greater fatigue development or greater fatigue resistance compared to the "Pure Static" condition and are discussed in more detail below.

Higher force levels induce greater fatigue

The maximum forces generated at 35%MVIE could have contributed to greater fatigue despite the mean muscle load for the "Varying Force" condition equalling that of the "Pure Static" exertions (i.e., 25%MVIE). The concept of the arterio-venous pressure gradient proposes that increases in intramuscular pressure reduce the amount of blood flowing to the working muscles (Humphreys & Lind, 1963; Hietanen, 1984; Sjøgaard *et al.*, 1988), thereby limiting the supply of nutrients and oxygen as well as waste product removal. Although the force level at which blood flow becomes occluded is still debated in the literature (e.g., Petrofsky & Hendershot, 1984; Fallentin & Jørgensen, 1992; Kapitaniak, 2001b; Yassierli & Nussbaum, 2009), it is plausible that the maximum forces of 35%MVIE exerted under the "Varying Force" condition could have had a negative effect on perfusion which could have led to the cascading event of decrements associated with reduced blood flow (as described in section 2.5.1 of the literature review). However, since the "Varying Force" condition did not result in greater fatigue responses, it is possible that no significant occlusion occurred during the maximum force exertions or that sufficient reperfusion occurred during the phases of minimum force application of 15%, thereby reducing the fatigue effects. Furthermore, Sjøgaard & Jensen (2006) proposed that it may be the duration of the increased intramuscular pressure rather than the absolute level that may be deleterious for the muscle rather than the force level itself. The takt time for each force cycle was a relatively short 2.6 seconds, which means that the duration of higher force applications (around 35%MVIE) may not have been long enough to have negative haemodynamic consequences.

Varying force levels enhance blood flow

The second possible outcome in this part of the experiment was that varying force levels promote blood flow and therefore retard fatigue development. Despite the evidence that greater exertion intensities result in greater restrictions in blood flow (e.g., Rohmert, 1960), an increase in the force production of a muscle may benefit blood flow due to the *muscle pump*. An opinion piece by Sheriff (2005) highlighted that the muscle pump is the result of several local and central circulatory mechanisms. While details of the muscle pump effect have been contested (Clifford *et al.*, 2005; Sheriff, 2005), there is general consensus that variations in mechanical compressions, such as the application and relaxation of pressure during muscular exertions, contribute to muscle perfusion in a similar manner to how rhythmical muscle actions can promote blood flow during dynamic exertions (Humphreys & Lind, 1963; Hamann *et al.*, 2003; Tschakovsky *et al.*, 1996). The force produced by the activated muscle(s) compresses the blood vessels weaving between and through muscles, emptying the venous circulation, which in turn creates a pressure gradient during relaxation, allowing for arterial inflow (Hamann *et al.*, 2003; Osada *et al.*, 2015). Clamann (1993), too, speculated that these rhythmical exertions might assist the venous return, thus excluding the influence of ischemia as a causal factor of muscle fatigue.

If varying muscle forces had a peristaltic effect on blood flow, fatigue responses under the “Varying Force” exertion should have been more favourable, while the “Pure Static” exertion should have shown greater fatigue. Given the uncertainty around the force levels at which blood flow starts having negative effects on haemodynamics, it is possible that the intramuscular pressure generated during the minimum and maximum force exertions under the “Varying Force” condition was insufficient to occlude blood flow. The force generated at the end of each cycle did not return to nil but had to maintain the low underlying force component of 15%MVIE, which may have resulted in similar blood flow restrictions as the “Pure Static” exertion and, therefore, insignificant differences in terms of fatigue.

The similar EMG outcomes further suggest that motor unit recruitment also did not change, which is supported by the cycle times and cycle-to-cycle variability results for the range of motion endpoints. According to Jørgensen *et al.* (1988), this does, however, vary from muscle to muscle. It can, therefore, be concluded that force variations cycling between 15% and 35% of MVIE do not have a significantly greater fatigue impact compared to a steady force exertion at 25%MVIE.

Effect of force variation on recovery

Over the course of the recovery period, peak torque returned to within 89.87% and 90.32% of the pre-fatigue values for the “Pure Static” and “Varying Force” exertions, respectively, a minor difference that was not statistically significant. Similarly, the rates of recovery from Post-Min 0 to Post-Min 5 also only showed minor differences between one another, with proportional improvements in peak torque of 8.70% and 6.76% for the “Pure Static” and “Varying Force" conditions, respectively, 40.79% vs 37.81%, respectively for the Fatigue Index, and 16.95% vs 16.49%, respectively, for the time to peak torque. Yates *et al.* (1987) attributed the slower recovery under isometric exertions to the (partial) occlusion of the blood vessels compared to fatigue incurred by dynamic exertions. The reasons for the lack of a differential recovery between the “Varying Force” and “Pure Static” exertions may thus be the same as the lack of greater fatigue under the two conditions.

6.3.3 Conclusion

When considering the results as a whole, the condition with the “Varying Length” exertions stands out as the one resulting in the greatest fatigue over the course of the fatigue protocol. However, this is mainly attributed to the contributions of the concentric movement phase, reasons for which may include a variety of factors relating to the occlusion of blood flow as well as motor unit recruitment. Conversely, varying forces exerted in a continuous isometric exercise protocol do not appear to have had any effect compared to a static exertion without the force variations. It is plausible that any time-related changes in some fatigue indicators are sensitive to the type of muscle action performed, since significant interaction effects were found for peak torque (when normalized to body weight), work, power and EMG(RMS) and median frequency, whereas none were identified for the Fatigue Index, the time to peak torque and performance accuracy and variability. These findings support the opening statements of this dissertation that localised muscle fatigue is complex, dynamic, and multifactorial in nature (e.g., Enoka, 1995; Hunter *et al.*, 2004; Seghers & Spaepen, 2004; Rashedi & Nussbaum, 2015; Tornero-Aguilera *et al.*, 2022; Behrens *et al.*, 2023). It is also this complexity that also prevents any further inferences from being made about the fourth type of muscle exertions that was proposed in the conceptual matrix in Chapter 3 (Figure 1 – field 4) even though it could not be included in the experiment itself.

6.4 LIMITATIONS

As with any emerging concept attempting to provide a novel contribution to a field, in this case, localized muscle fatigue under continuous low-level exertions, it is important to acknowledge several limitations inherent in the research concept, study sample, as well as the methods and procedures used to test the hypotheses, and which should be considered in future studies in this area of research.

6.4.1 Study Sample

Firstly, the relatively small sample size of 36 participants limits the inferences that can be made about the current study's findings to the greater population. Similarly, the lack of diversity of the participants, i.e., they were all young, relatively healthy and physically active students, would also impact the external validity of the study. Having said that, selecting a relatively homogeneous group of individuals was considered important since fatigue responses can be influenced by a variety of individual factors, including age, sex and training status (Sjøgaard & Jensen, 2006). A concerted effort was made to prevent any demographic and anthropometric differences from influencing fatigue responses. This was done by personalizing equipment setup, relativizing the external loads to each participant's maximum voluntary strength, and, during the data analysis phase of the study, normalizing individual responses to their unfatigued states. Nonetheless, sex may still have had a differential influence on fatigue. For example, Nuzzo *et al.* (2023) found significant differences in fatigue responses after a maximal fatigue protocol between males and females, while other studies found greater fatigue resistance in females (refer to a review by Hicks *et al.*, 2001 and Senefeld *et al.*, 2013). It can, therefore, not be excluded that male-female differences could have influenced strength expression, muscle recruitment and perceived exertion. Comparisons of the sexes were, however, beyond the scope of this study.

Similarly, training status is another factor that may have influenced fatigue responses. Participants in the current study were required to engage in regular physical activity; however, the rationale for this was primarily for ethical reasons, to prevent overexertion and injury, particularly during maximal exertions. Therefore, the type of physical activity participants engaged in was not controlled for. This could have led to, for example, well-trained (and therefore strong) participants having to move larger loads, even though they may not have had the stamina to maintain such forces. Komi & Tesch (1979) also reported different proportions of fibre types in the muscles of trained and untrained individuals, which could have influenced fatigue resilience.

Another limitation was the point of exercise termination, which relied on subjective perceptions (i.e., termination of the exercise protocol when participants rated their perceived exertion at 9). This means that from a metabolic and neurological perspective, participants may have achieved different levels of fatigue between one another, but also from one experimental session to the next. However, by having engagement in regular physical activity as a pre-requisite, it was hoped that participants would be familiar with their own physical capabilities as well as the sensations associated with muscle fatigue and achieve similar levels of fatigue across conditions. Physical exercise comes with a variety of sensations, including force, effort, pain, and discomfort (Pageaux & Lepers, 2016), which, amongst others, may have influenced the perceived exertion and thus influenced the point of termination of the fatigue exercise protocol. These factors, alongside others such as the level of motivation or environmental temperature, are rapidly variable and could not be controlled in the current study.

6.4.2 Methodological & Protocol-related factors

Equipment Limitations

Arguably the greatest limitation in the current study was the selection of the forces that participants had to manipulate during the experimental conditions. The external loads to be held or moved were calculated from the maximum voluntary static or dynamic exertions produced during maximal exertions on the isokinetic dynamometer. Obtaining the ‘true’ maximum forces was, however, challenging since the ability to generate a brief maximum voluntary force varies from individual to individual, from day to day, and between trials (Allen *et al.,* 1995). Furthermore, there may have been the influence of the unfamiliarity with the equipment, which may have influenced the maximum force readings. Along a similar line of thought, since the loads moved under the “Varying Length” condition were calculated as the average of the concentric and eccentric peak torques, which were significantly different from one another, the external muscle loading for these two exertion phases would not have been 25% of

MVE. Unfortunately, the researcher did not have access to a CARE machine, which could have overcome this challenge.

Another equipment-related limitation of the study was the absence of a measure that could quantify blood flow. While occlusions in blood flow were suspected to play a role in the development of muscle fatigue, no equipment that could measure blood flow (see review by Jayanthy *et al.*, 2011) was available to the researcher. Any inferences about the impact of blood flow on muscle fatigue during the different exertions remain, therefore, speculative.

The use of surface EMG also comes with various limitations since muscle and skin properties, as well as muscle geometry and the nature of the muscle exertion, can influence the electrical reading picked up by the surface electrodes (Schmitz *et al.,* 2002). The most pertinent factor to the current study was the movement of the muscle belly relative to the EMG electrodes during dynamic exertions, which may invalidate comparisons, particularly between the "Varying Length" and "Pure Static" conditions. Along the same lines, despite great care, the placement of the electrodes from one testing session to another may not have been exactly in the same position, and this could have influenced the electrical signal picked up by the electrodes (Rainoldi *et al.*, 2001). Additionally, the muscular force generated during the elbow movement is the combined effort of three synergistic muscles, namely, biceps brachii, brachialis and brachioradialis, yet only the EMG of the biceps brachii was recorded due to its suitability for surface EMG measurements. Contributions of each muscle towards overall force generation vary between different exertion types (Orantes-Gonzalez *et al.*, 2023), and also change as fatigue progresses (Stutzig & Siebert, 2015). The inability to separate the individual contributions of the three elbow flexors would have limited the interpretations that could be made about the effects of muscle length and muscle force on fatigue. Furthermore, although the input of wrist flexor muscles on the pulling force was assumed to be negligible and participants were instructed to keep the wrist stable, these muscles may still have influenced the force produced.

Procedural Limitations

One potential source of measurement error could have occurred when assessing the maximum torque exerted during the maximum voluntary isometric exertions prior to the fatigue protocol. Ten seconds of rest were allocated between each 5-second maximum voluntary isometric exertion, rather than the generally recommended 1-2 minutes rest break, which would ensure substrate replenishment and neural readiness. Similarly, assessment of peak torque during the maximal dynamic exertions involved three repetitions of alternating concentric and eccentric exertions on the isokinetic dynamometer. By only affording participants a limited recovery period between maximal exertions, both static and dynamic, could have resulted in lower peak torque values, which, in turn, would have influenced the determination of the sub-maximal loads to be manipulated during the fatigue protocol. Differences in the mean muscle loading between conditions would influence the comparability between conditions and could have contributed to the lack of differences found between the experimental conditions.

Testing occurred throughout the day from 7:30 am to 4:30 pm. Given that inter-individual performance varies at different times during the day (Wyse *et al.*, 1994; Drust *et al.*, 2005; Douglas, 2021), individuals should have been tested during the same time slot for every experimental condition, and ideally even at times that corresponded with their circadian peak. While this was practically not feasible, it was ensured that testing sessions for each individual always occurred within a two-hour window period and, therefore, at the same phase of their circadian rhythm. Furthermore, the time elapsed between testing sessions one and two was, on average, 16 days, while between sessions two and three, it was 13 days. However, this was highly variable, with the maximum number of days between testing sessions for one participant being 73 days (due to equipment breakdown and replacement delays). Changes in strength capacity due to illness, training or other factors may have had an impact on the fatigue responses under the experimental conditions that were far apart.

The study aimed at understanding fatigue responses during continuous force generation under different muscle exertion types. To ensure continuous muscle activation and to avoid the influence of any micro rest breaks, the fatigue protocol purposefully was not interrupted. However, there was a brief transition period in the experimental protocol where, upon termination of the exercise, participants had to be moved from the pulley system to the Biodex isokinetic dynamometer, resulting in an approximately 10-second delay before performing the first of the post-protocol maximum exertions. During this period, some reperfusion is likely to have occurred, and possibly at different rates for the different conditions, which may thus have influenced the Post-Min 0 and subsequent recovery readings. Although this transition period would have been similar for all experimental conditions, it may still have influenced the recovery data.

Practical Relevance

It may be claimed that this study has limited practical relevance in terms of affecting working conditions and fatigue during low level physical work. However, the contribution of this thesis lies in the systematic approach in which muscle contractions are classified and studied, which in future will allow for a more systematic way to approach activities that may result in muscular fatigue.

From a methodological perspective, performing an activity that lasted, on average, just under five minutes, may not reflect real-world working conditions, which is why some investigations used protocol durations lasting until exhaustion or task failure (e.g. Yassierli & Nussbaum, 2009; Dickerson *et al.*, 2015), while other studies have restricted the protocol duration to between 20 and 60 minutes and obtained significant outcomes (Hermans & Spaepen, 1997; Seghers & Spaepen, 2004; Iridiastadi & Nussbaum, 2006). However, to test the concept put forward in this study, longer durations would not have been necessary, nor could they have been possible since participants were close to task failure anyway.

Finally, from a theoretical perspective, it could be argued that an understanding of the subtle variations in muscle exertions is probably not pragmatic, nor do the findings of the current study solve the problem of muscle fatigue, particularly since muscle actions in occupational and sports settings consist of a combination of varying muscle lengths and forces, in other words, the exertions depicted in Figure 1- field 4 of the conceptual matrix in Chapter 3. However, rather than trying to provide solutions, this study set out to bridge some of the existing knowledge gaps regarding localised muscle fatigue of submaximal exertions.

CHAPTER 7: CONCLUSION AND RECOMMENDATIONS

7.1 SUMMARY OF KEY FINDINGS

Research into muscle fatigue is challenging due to its many influencing factors, including individual, task-related, and environmental factors. The current study was concerned with two task-related factors known to influence fatigue development. More specifically, muscle length and force exertion level were selected as variables of interest in this study, as these are governed by the requirements of occupational, sports or daily living activities. This study posed the question whether static and dynamic muscle exertions could be classified according to length and force characteristics, and if combinations of these, i.e., varying length combined with static force or static length combined with a varying force, would influence fatigue responses and recovery.

All parameters recorded indicated that the exercise protocol participants engaged in had indeed fatigued the elbow flexors. Maximum voluntary exertions obtained prior to the fatigue protocol and immediately after revealed significant decrements in peak torque ($p<0.0001$), as well as a more than 20-fold increase in the Fatigue Index. Work and power obtained only during the “Varying Length” condition also revealed significant decrements during maximal exertions ($p<0.0001$). The 5-minute recovery period demonstrated significant improvements in these variables, but the rate of recovery for the different parameters varied, with some variables, such as peak torque, Fatigue Index, work, and power, only partly recovering, while others, e.g., time to peak torque, increased beyond pre-fatigue levels. Fatigue indicators obtained at the start and the end of the submaximal exercise protocol revealed significant increases in the EMG amplitude ($p<0.0001$) and reductions in the median frequency of the power spectrum ($p<0.0001$). These findings concur with the literature, confirming that localised muscle fatigue had occurred. Performance accuracy and variability data, used as indicators of motor control, revealed inconsistent findings in the deviation from the target cycle times, forces, and joint angles. These inconsistencies are explained by varying compensatory mechanisms employed under the different exertions in an attempt to maintain task performance.

Comparisons of exertions with and without varying muscle lengths, but with the same external load, indicated some unexpected fatigue responses. The “Varying Length” condition experienced greater muscle fatigue and longer recovery compared to the "Pure Static" condition, even though the opposite was expected. The greater fatigue was evidenced by shorter endurance times ($p<0.0001$), fewer movement cycles completed ($p<0.0001$), greater percentage change (relative to pre-fatigue values) in peak torque ($p<0.0001$), a greater Fatigue Index, greater proportional increases in the time to peak torque ($p<0.001$), relative decreases in the median frequency of the EMG ($p<0.0001$) and a greater proportional increase in the deviation from the target force. The greater fatigue found for the "Varying Length" condition was attributed to the concentric movement phase of the "Varying Length" condition. While the greater fatigue under the "Varying Length" condition contradicts other studies, it may be the outcome of greater loading during the "Varying Length-Concentric" exertion relative to its force-producing capacity. EMG amplitude during the dynamic exertion experienced only marginally greater increases than the static one, while cycle-to-cycle variability for the "Varying Force" condition exhibited significantly smaller changes than the "Pure Static" muscle actions.

Contrary to the significantly greater fatigue responses under the "Varying Length" compared to the "Pure Static" exertions, comparisons of the "Varying Force" to the "Pure Static" responses yielded no differences in fatigue magnitude or recovery. The statistical analyses showed no significant differences ($p>0.05$) in cycle time, number of movement cycles, peak torque, time to peak torque, the Fatigue Index, EMG amplitude and median frequency, as well as cycle-to-cycle variability between the two conditions. Only the performance accuracy of the "Varying Force" condition was significantly poorer, but this can be explained by the kinematics of the force applications. The lack of differences in fatigue between the "Varying Force" and "Pure Static" exertions may be explained by the level of force generation and takt time. The mean load level of 25% maximum voluntary isometric exertion may be too low to significantly occlude blood flow, particularly when the takt time from one force cycle to the next is only 2.6 seconds.

7.2 RECOMMENDATIONS FOR FUTURE RESEARCH

This study merely provided a snapshot insight into the effects of task-related factors hypothesised to influence muscle fatigue in an attempt to identify the more subtle differences in fatigue development during static and dynamic muscle actions. Although two task-related characteristics, namely muscle length and muscle force, were investigated, the classification of muscle exertions used in the current study excluded a third factor highlighted in the literature to be an important contributor to fatigue, namely movement velocity. In terms of characterising force generation within a muscle, movement speed should be an additional consideration, and the proposed classification of muscle exertions should be expanded in future studies to include velocity.

The markedly different fatigue responses between the concentric and eccentric movement phases of the “Varying Length” condition, and the lack of difference in fatigue responses between the eccentric and isometric exertions, warrant further research efforts. Since one limitation discussed was the inability to actively adjust the external load during concentric and eccentric movement phases of an exercise, a consideration in future studies of a similar kind would be to incorporate different technology that would allow the active adjustment of the resistance applied, such as by means of a CARE machine. One further step to consider in truly controlling the force factor in future studies would be to quantify and control the internal force generated by the muscle, rather than the external load, which was considered a significant methodological limitation of the current study.

Further suggestions for future research of the current concept include expanding on the number of muscles tested, as different muscle fibre types are known to determine fatigue resilience. Firstly, considerations of all synergistic muscles involved in force production during a movement are important since the relative contributions of muscles change with exertion type as well as with fatigue. Furthermore, it is recommended to expand on the range of motion during dynamic exertions, the forces applied, and movement speeds. The range of motion in the current experiment was set to avoid the extremes in joint angles; however, it is at those angles that the muscle may be maximally stretched or “coiled up” and where the reduced perfusion due to greater stretch may be detected. Minimum and maximum forces generated under the different conditions in the current study were set at relatively low levels of maximal capacity to mimic muscle loading common in a variety of occupational settings. Further studies should expand on these force levels (both the average muscle load, as well as minimum and maximum levels), as this may provide further insights into whether the suspicion of reductions in blood flow is worth investigating in more detail. Additionally, differences in motor unit recruitment to different movement velocities during dynamic exertions should be investigated with different cycle durations and/or movement speeds.

The current study investigated fatigue in a relatively uniform group of volunteers and within a very narrow range of task characteristics. The decisions regarding these were mostly made for pragmatic and ethical reasons and have been detailed in Chapter 4. Study participants were young, healthy, and generally active, yet other characteristics, such as training status, nutrition status, and psychological states (e.g., mood or motivation), just to name a few characteristics known to influence strength-producing capacity, were not controlled for and could have contributed to the variability in the fatigue and recovery responses. Conversely, a wider range in age, as well as physical capabilities, would allow better extrapolation of the findings to the general population.

The data collected in the current study could benefit from some mathematical modelling. While this analysis technique was considered beyond the scope of the current study, it may be useful to make inferences about the combined effects of varying muscle length and varying muscle force on fatigue, i.e., addressing field 4 in the experimental framework (Section 3.1; Figure 1).

7.3 CONCLUDING REMARK

The current study made use of a novel approach of redefining muscle exertions in an attempt to better understand the intricate and multifaceted nature of localised muscle fatigue development and recovery during submaximal static and dynamic exertions, since task-dependent characteristics are known to influence the metabolic and neuromuscular processes during force generation. The time-related decrements in the kinetic and outcome parameters recorded over the duration of the fatigue protocol confirmed that the fatigue parameters selected were indeed suitable to use for the identification of localized muscle fatigue. The unexpected findings of significantly greater fatigue during muscle exertions with a varying muscle length component, and the similarities in the fatigue responses between the exertions with an isometric component, suggest that comparative studies of static and dynamic exercises must be approached carefully, as the assessment of muscle fatigue during dynamic force exertions may be influenced by numerous task-related factors that can result in divergent physiological responses and may, therefore, not be directly comparable.

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APPENDICES

APPENDIX A: SUPPORTING DOCUMENTATION

Appendix A.1: Letter of Information to Participants

RHODES UNIVERSITY

DEPARTMENT OF HUMAN KINETICS AND ERGONOMICS

LETTER OF INFORMATION TO PARTICIPANTS

Dear ______________________________,

Thank you for your interest in my study provisionally titled "*An investigation into the effects of muscle force and muscle length on the development of localized muscle fatigue*".

BACKGROUND AND AIM OF THE STUDY

My name is Miriam Mattison and I am an academic staff member of the Department of Human Kinetics and Ergonomics. I am currently pursuing a PhD degree under the supervision of Dr Swantje Zschernack. My research interests lie in the area of ergonomics in general, and for this project specifically, in muscle fatigue development under sub-maximal working conditions.

The purpose of this particular study is to improve our understanding of muscle fatigue responses for different types of muscle exertions (muscle contractions). Muscle fatigue plays a significant role in discomfort, accidents and injuries, low performance, poor perceptions of work, reduced quality of life, and is even thought to contribute to the development of work-related musculoskeletal disorders. It is hence in the interest of both employers and employees that muscular fatigue is better understood, so that work practices and routines can be improved.

Muscle exertions can be described in terms of their length (e.g. whether the muscle lengthens, shortens or remains unchanged during a contraction), and the force produced (e.g. whether the force created remain consistent, or varies during an exertion). However, the effects these two factors have on muscle fatigue in isolation, and in conjunction with one another, is not well known, and this is what my PhD is investigating.

PROCEDURES

This study and its procedures have been reviewed and approved by the Rhodes University Ethical Standards Committee. To participate in this study, you will be required to present yourself for four sessions at the Human Kinetics and Ergonomics Department.

Session 1 is an information and familiarization session, which will be dedicated to informing you verbally of the details of this experiment, as well as familiarizing you with the equipment and procedures of the study. Once you are satisfied with the requirements and procedures of the experiment, you will sign an informed consent form. Some basic demographic and anthropometric data will be recorded and you will be given the opportunity to familiarize yourself with the experimental conditions and the equipment. In addition, you will participate in an endurance test during which you will need to hold a weight (25% of your maximum strength) in your hand with your elbow flexed at 90° and for as long you can maintain. This entire session should last approximately 1.5 hours.

Sessions 2-4: During these sessions, which will last about 45 minutes to 1 hour each, you will perform the three experimental conditions (one condition per session). The tasks you will be performing during these conditions have been designed to fatigue your biceps brachii muscle using different types of muscle exertions – either by holding a set load still in your hand, or by continuously moving your elbow through a range of motion, or by pulling against an immovable object with different levels of force. During these sessions, we will measure the force, muscle activity (EMG), movement parameters (e.g. joint angles, velocities and accelerations) of your biceps brachii muscle, as well as your perceived effort of the biceps brachii muscle during a given task.

During the experimental sessions, we will first need to prepare your skin for the EMG equipment, which will measure muscle activity. This entails cleaning a small area of skin over your biceps brachii muscle of your upper arm with an alcohol swab, and shaving this area of your upper arm where the surface EMG electrodes will sit to prevent any hair from interfering with the measurement quality. These EMG electrodes (stick-on pads to which cables are connected) will be secured using medical tape and connected to the Datalogger system. Similarly, you will have two other sensors attached to your arm with medical tape – one sensor is an electrogoniometer, which measures your joint movement angles, while the second sensor, known as an accelerometer, measures the velocities and accelerations of your movements. You will then be requested to perform 20-30 continuous elbow flexion and extensions movements with a light weight (0.5kg) and at a prescribed speed, to warm up your muscle, preparing it for the testing protocol. Thereafter, for baseline measurements, you will perform three maximal static and dynamic exertions against the handle of the Biodex Isokinetic Dynamometer (a sophisticated piece of equipment that measures maximum force produced for different types of exertion - you would have practised this during the familiarization session). Each exertion type is interspersed by a sufficient rest period to prevent fatigue accumulation. The force recorded during the maximal exertions will be used to calculate 25% of your maximal force. A pulley system (similar to the types of machines you encounter in a gym) allows for a weight to be attached that weighs the equivalent of 25% of your maximum force produced on the isokinetic dynamometer. This is the load you have to move during the subsequent three experimental conditions.

1) Condition 1: you will produce a constant force equal to 25% of your static maximum elbow flexion force against the pulley system with your elbow at a 90° angle for a duration slightly shorter than the one determined during the endurance test in the information session.
2) Condition 2: you will move a resistance equalling 25% of your dynamic maximum elbow flexion force through a range of motion of 80° and at a speed of $60°.s^{-1}$. This means you will have 1.6 seconds to flex your elbow, followed by another 1.6 seconds to extend your elbow; a metronome will help you maintain the correct pace.
3) Condition 3: during this condition, you will have to vary your force production between 15% and 35% of your static maximum torque, but without the pulley system allowing any movement to occur. By pulling against an immovable handle that is attached to an analogue force gauge (scale), you can visually control these force exertions. The timing between force exertions will be 1.6 seconds for each force, (i.e. 1.6 seconds for the 15%MVE force and 1.6 seconds for the 35%MVE force). Recovery of muscle function will be recorded during the recovery period after the fatigue protocol of each condition. You will be provided a 1-minute rest break, after which you will need to provide another 3 maximum exertions against the Biodex handle. After another minute of rest, you need to produce another 3 maximum exertions. This is repeated 5 times during which the force produced and your muscle activity will be recorded.

Throughout each condition force, muscle activity, and movement parameters (joint angles, velocities and accelerations) will be recoded continuously, while once every minute you will be asked to rate the effort of your biceps brachii muscle on a Likert scale (Ratings of Perceived Exertion CR-10 scale). Once your perceived effort reaches a rating of '8', the protocol is terminated and you will have to produce another three maximum isometric exertions against the Biodex isokinetic dynamometer each minute for the next 5 minutes.

PARTICIPANT CHARACTERISTICS

To participate in this study, you should be between the ages of 18 – 25 years, be generally in good health, and more specifically, have no history of musculoskeletal injury or disorder in the past 12 months. Please do not smoke, drink alcohol or caffeine-rich drinks 12 hours before testing, as this may affect your endurance and force-producing capability. Also, please avoid any strenuous exercise two days before testing as this too may affect your muscles' ability to produce force.

RISKS AND BENEFITS

The above activities should pose little risk for physical and psychological harm, if you have no previous or present musculoskeletal injuries of the dominant arm, upper back or neck regions, and if you correctly follow the protocol as per the researcher's instructions. The risks include:

- Minor cuts: Since the skin area where the EMG electrodes will be situated has to be shaved, there is the remote chance of minor skin abrasions or cuts. Since a new razor is used for each participant, and the area is thoroughly cleaned prior to shaving, the chance of infection is minimized. Should a minor cut or excessive skin abrasion occur, the injured area will again be disinfected and treated with antiseptic cream and a plaster.
- Embarrassment / psychological discomfort: The preparation and set-up of the EMG equipment will entail physical contact, as I have to locate the appropriate position on your upper arm to place the electrodes. This may possibly make you feel uncomfortable. However, the procedures of EMG electrode placement are standardized and are minimally invasive. Wearing a short-sleeved shirt will help me to easily access the muscle under investigation. Additionally, since you will have to perform maximal exertions, you may feel your performance is not adequate. Rest assured, however, that maximum muscle strength is individually specific and that your best effort is all that is required. Testing will also occur in a private area in an HKE laboratory, so there are no interruptions by other people.
- Muscle strain: You are requested to perform maximal voluntary muscle exertions. This carries the risk of muscle strain, i.e. small tears in the muscle that are painful, but heal with time. The chances of this happening have however been minimized by having you warm up the biceps brachii muscle and by standardizing the maximum exertion procedure. Additionally, a muscle itself will not overexert itself during a voluntary exertion, so the risk of muscle strain is minimal.
- Delayed Onset of Muscle Soreness (DOMS): Due to the possibly unaccustomed nature of the muscle exertions, particularly after eccentric (lengthening) muscle actions, you may develop some DOMS (colloquially referred to as a "stiff muscle") in your biceps brachii, which is associated with some discomfort/pain and occurs within 2 days after the exercise. However, this is only temporarily and usually disappears within 7-10 days. Should you still suffer from DOMS by the next scheduled testing session, please let me know so we can postpone the testing until it has disappeared, since feelings of discomfort/pain are not only unpleasant to you, but will also influence the testing results.

It is your duty to inform the researcher in advance if you are not feeling well on the day of testing, or during the experiment. If you experience any negative effects as a result of the testing protocol, e.g. pain, dizziness, nausea etc., please let the researcher know immediately, so that these issues can be addressed.

The benefits of the study to you are mainly educational in nature, since you will get a first-hand insight into scientific testing, equipment used in our field of research and you may learn something about yourself and your capabilities you did not know before. The greater benefits of the research will be the contribution towards a better understanding of muscle fatigue and the possible reduction of work-related fatigue by improving work regimes.

OTHER RIGHTS AND RESPONSIBILITIES

Anonymity: your anonymity will be ensured by using a code, instead of your name when collecting the data. The data will remain in the possession of the HKE Department and will only be used to inform my main study and a publication. In the latter case, only summary data of all participants will be used, so no individual data will be presented.

Finally, be aware that you have the right to withdraw from the study at any point in time, without any negative consequences to you. Your withdrawal will not influence your relationship with any person in the HKE Department negatively.

Please feel free to take a day or two to decide whether, or not, you want to participate in my study. If you are unsure about any of the above information, requirements, or the reasoning behind them, please feel free to ask me at any time, or during the information session. You can also address any concerns about the testing protocol to my supervisor, Dr Zschernack (s.zschernack@ru.ac.za; 046-603 8472), or the Rhodes University Ethical Standards Committee's ethics coordinator, Mr Manqele (s.manqele@ru.ac.za; 046-603 7727).

Yours sincerely,

Miriam Mattison

____

Miriam Mattison
Tel: 046-603 8468
Cell: 082-319 4626
Email: m.mattison@ru.ac.za

Appendix A.2: Informed Consent Form

RHODES UNIVERSITY

Grahamstown • 6140 • South Africa

DEPARTMENT OF HUMAN KINETICS AND ERGONOMICS
INFORMED CONSENT AND INDEMNITY

I, ..., have been fully informed of the PhD research project provisionally titled: "*An investigation into the effects of muscle force and muscle length on the development of localized muscle fatigue*".

I have read the information sheet and understand the testing procedures that will take place. All procedures, associated risks and the benefits from partaking in this study have been verbally explained to me verbally, as well as in writing (refer to letter of information appended to this document). I have had ample opportunity to ask questions and clarify any concerns or misunderstandings. These have been answered to my satisfaction. I understand that all data collected for publication purposes will be kept anonymous and all information gained in this regard will be treated confidentially. Furthermore, I consent to photographs, knowing that these will be altered to ensure my anonymity. I understand that I am able to withdraw from the study at any point, irrespective of external influences placed on me by the researcher and that withdrawal will not negatively affect my relationship with individuals in the Department of Human Kinetics and Ergonomics.

In agreeing to participate in this research study, I waive any legal recourse against the researchers from the Department of Human Kinetics and Ergonomics (HKE), Rhodes University, from claims resulting from personal injuries sustained whilst participating in the above mentioned research due to my personal negligence and non-compliance with instructions. This waiver shall be binding upon my heirs and personal representatives.

I have read and understood the above information, as well as the information provided in the letter accompanying this form. I therefore consent to voluntarily participate in this research project.

PARTICIPANT PROVIDING CONSENT:

________________	________________	________________
(Print Name)	(Signed)	(Date)

WITNESS:

________________	________________	________________
(Print Name)	(Signed)	(Date)

RESEARCHER:

________________	________________	________________
(Print Name)	(Signed)	(Date)

www.ru.ac.za

Appendix A.3: Ethical Clearance Letter

RHODES UNIVERSITY
Where leaders learn

Human Ethics subcommittee
Rhodes University Ethical Standards Committee
PO Box 94, Grahamstown, 6140, South Africa
t: +27 (0) 46 603 8055
f: +27 (0) 46 603 8822
e: ethics-committee@ru.ac.za

www.ru.ac.za/research/research/ethics
NHREC Registration no. REC-241114-045

1 May 2019

Ms Miriam Mattison

Human Kinetics & Ergonomics

Email: M.Mattison@ru.ac.za

Dear Ms Miriam Mattison

Re: Effects of muscle force and length on fatigue , Effects of muscle force and length on fatigue (0150 , Mar , 2019)

Principal Investigator: Dr Swantje Zschernack

Collaborators: Mrs Miriam Mattison ,

This letter confirms that the above research proposal has been reviewed and **APPROVED** by the Rhodes University Ethical Standards Committee (RUESC) – Human Ethics (HE) sub-committee.

Approval has been granted for 1 year. An annual progress report will be required in order to renew approval for an additional period. You will receive an email notifying when the annual report is due.

Please ensure that the ethical standards committee is notified should any substantive change(s) be made, for whatever reason, during the research process. This includes changes in investigators. Please also ensure that a brief report is submitted to the ethics committee on completion of the research. The purpose of this report is to indicate whether the research was conducted successfully, if any aspects could not be completed, or if any problems arose that the ethical standards committee should be aware of. If a thesis or dissertation arising from this research is submitted to the library's electronic theses and dissertations (ETD) repository, please notify the committee of the date of submission and/or any reference or cataloguing number allocated.

Sincerely

JDames

Prof Joanna Dames

Chair: Human Ethics sub-committee, RUESC- HE

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Appendix A.4: Permission Letter from Gatekeeper

RHODES UNIVERSITY
Grahamstown • 6140 • South Africa

OFFICE OF THE REGISTRAR
P O Box 94, Grahamstown, 6140
E-mail: registrar@ru.ac.za
Tel: +27 (0)46 603 8101
Fax: +27 (0)46 603 8127

Ms Miriam Mattison
Department of Human Kinetics and Ergonomics

25 April 2019

Dear Ms Mattison

Name of research proposal: Effects of muscle force and length on fatigue.

This serves to confirm that you have been granted permission to conduct your proposed research at Rhodes University as requested.

The University is not obliged to make any arrangements in terms of this research. The onus is on the researcher.

Yours sincerely

AL Moody

Dr Adéle Moodly
REGISTRAR

www.ru.ac.za

APPENDIX B: STATISTICAL TABLES

Please note the following:

- Analyses highlighted in red represent statistically significant findings.
- Common acronyms used in the statistical tables:
  VL = “Varying Length” condition
  PS = “Pure Static” condition
  VF = “Varying Force” condition

Appendix B.1: Comparisons for Sex

(a) Age

Table 19: Comparison between the sexes for age

Effect	Univariate Tests of Significance for Age (yrs) (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 1.8222 SS	Degr. of Freedom	MS	F	p
Intercept	17161.00	1	17161.00	5168.569	0.000000
Sex	0.11	1	0.11	0.033	0.855937
Error	112.89	34	3.32		

(b) Stature

Table 20: Comparison between the sexes for stature

Effect	Univariate Tests of Significance for Stature (m) (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 0.0731 SS	Degr. of Freedom	MS	F	p
Intercept	103.9550	1	103.9550	19442.57	0.000000
Sex	0.1219	1	0.1219	22.80	0.000034
Error	0.1818	34	0.0053		

(c) Body Mass

Table 21: Comparison between the sexes for body mass

Effect	Univariate Tests of Significance for Mass (kg) (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 13.6382 SS	Degr. of Freedom	MS	F	p
Intercept	187546.7	1	187546.7	1008.307	0.000000
Sex	1860.5	1	1860.5	10.003	0.003283
Error	6324.1	34	186.0		

(d) Body Mass Index (BMI)

Table 22: Comparison between the sexes for BMI

Effect	Univariate Tests of Significance for BMI (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 4.0320 SS	Degr. of Freedom	MS	F	p
Intercept	22254.70	1	22254.70	1368.955	0.000000
Sex	16.73	1	16.73	1.029	0.317528
Error	552.73	34	16.26		

(e) Peak Torque

Table 23: Comparison between the sexes for absolute peak torque (Nm)

Effect	Repeated Measures Analysis of Variance (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 22.3384 SS	Degr. of Freedom	MS	F	p
Intercept	492558.3	1	492558.3	987.0819	0.000000
Sex	53041.9	1	53041.9	106.2955	0.000000
Error	16966.2	34	499.0		
EXERTION	1493.8	3	497.9	15.2047	0.000000
EXERTION*Sex	75.7	3	25.2	0.7704	0.513190
Error	3340.3	102	32.7		

Table 24: Comparison between the sexes for peak torque relative to Body Mass ($Nm.kg^{-1}$)

Effect	Repeated Measures Analysis of Variance (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 0.3057 SS	Degr. of Freedom	MS	F	p
Intercept	93.21616	1	93.21616	997.3294	0.000000
Sex	5.36484	1	5.36484	57.3990	0.000000
Error	3.17784	34	0.09347		
EXERTION	0.26743	3	0.08914	14.0304	0.000000
EXERTION*Sex	0.01505	3	0.00502	0.7896	0.502423
Error	0.64807	102	0.00635		

(f) Endurance Time

Table 25: Comparison between the sexes for endurance time

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 2.5277

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	2218.613	1	2218.613	347.2316	0.000000
Sex	17.521	1	17.521	2.7422	0.106935
Error	217.241	34	6.389		
EXERTION	84.019	2	42.009	34.7977	0.000000
EXERTION*Sex	0.389	2	0.194	0.1611	0.851561
Error	82.093	68	1.207		

(g) Number of Cycles

Table 26: Comparison between the sexes for number of completed cycles

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 58.1385

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	1146197	1	1146197	339.1026	0.000000
Sex	8356	1	8356	2.4720	0.125149
Error	114923	34	3380		
EXERTION	47298	2	23649	37.4486	0.000000
EXERTION*Sex	343	2	172	0.2720	0.762712
Error	42943	68	632		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 631.51, df = 68.000

EXERTION	Varying Length	Pure Static 120.17	**Varying Force**
Varying Length		0.000112	0.000112
Pure Static	0.000112		0.694349
Varying Force	0.000112	0.694349	

(h) Fatigue Index

Table 27: Comparison between the sexes for the Fatigue Index

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 18.6844

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	90398.98	1	90398.98	258.9424	0.000000
Sex	144.10	1	144.10	0.4128	0.525008
Error	11520.58	33	349.11		
TIME	4380.42	1	4380.42	34.6000	0.000001
TIME*Sex	22.71	1	22.71	0.1794	0.674632
Error	4177.86	33	126.60		
EXERTION	10564.22	3	3521.41	17.0752	0.000000
EXERTION*Sex	211.12	3	70.37	0.3412	0.795553
Error	20416.74	99	206.23		
TIME*EXERTION	223.08	3	74.36	0.8961	0.446102
TIME*EXERTION*Sex	15.88	3	5.29	0.0638	0.978855
Error	8215.82	99	82.99		

Appendix B.2: Correlation Analyses

Table 28: Correlation results for endurance time vs load manipulated

Correlations (Data for Statistica 2025)
Marked correlations are significant at p < .05000
N=36 (Casewise deletion of missing data)

Variable	**VL - Time to RPE 9 (min)**	PS - Time to RPE 9 (min)	VF - Time to RPE 9 (min)
VL - Weight Moved (kg)	-0.425121	-0.378344	-0.447480
PS - Weight Held (kg)	-0.346136	-0.456750	-0.451040
VF - Min Weight Moved (kg)	-0.331433	-0.397538	-0.438036
VF - Max Weight Moved (kg)	-0.331970	-0.396608	-0.438754

Table 29: Correlation results for endurance time vs peak torque

Correlations (Data for Statistica 2025)
Marked correlations are significant at p < .05000
N=36 (Casewise deletion of missing data)

Variable	**VL - Time to RPE 9 (min)**	PS - Time to RPE 9 (min)	VF - Time to RPE 9 (min)
VL - Max Con Torque (pre) (Nm)	-0.388708	-0.375088	-0.441106
VL - Max Ecc Torque (pre) (Nm)	-0.393132	-0.372560	-0.416419
PS - Max Isometric Torque (pre) (Nm)	-0.318748	-0.436638	-0.426141
VF - Max Isometric Torque (pre) (Nm)	-0.294564	-0.372388	-0.412877

Table 30: Correlation results for Peak Torque and Work

Correlations (Data for Statistica 2025)
Marked correlations are significant at p < .05000
N=35 (Casewise deletion of missing data)

Variable	Means	Std.Dev.	VL - con - Work done (Pre)	VL - con - Work done - (Post-Min 0)	VL - con - Work done - (Post-Min 5)	VL - ecc - Work done (Pre)	VL - ecc - Work done - (Post-Min 0)	**VL - ecc - Work done - (Post-Min 5)**
VL - con - Work done at PT (Pre)	56.09714	21.18616	1.000000	0.869520	0.810036	0.880999	0.688218	0.875932
VL - con - Work done - (Post-Min 0)	36.78571	13.75130	0.869520	1.000000	0.841204	0.808276	0.713533	0.797190
VL - con - Work done - (Post-Min 5)	44.18286	17.08091	0.810036	0.841204	1.000000	0.732355	0.725968	0.838350
VL - ecc - Work done at PT (Pre)	69.12857	27.46032	0.880999	0.808276	0.732355	1.000000	0.753540	0.844605
VL - ecc - Work done - (Post-Min 0)	43.32000	19.20196	0.688218	0.713533	0.725968	0.753540	1.000000	0.793462
VL - ecc - Work done - (Post-Min 5)	56.76571	24.40693	0.875932	0.797190	0.838350	0.844605	0.793462	1.000000

Table 31: Correlation results for Peak Torque and Power

Correlations (Data for Statistica 2025)
Marked correlations are significant at p < .05000
N=35 (Casewise deletion of missing data)

Variable	Means	Std.Dev.	VL - con - Ave Power (Pre)	VL - con - Ave Power - (Post-Min 0)	VL - con - Ave Power - (Post-Min 5)	VL - ecc - Ave Power (Pre)	VL - ecc - Ave Power - (Post-Min 0)	**VL - ecc - Ave Power - (Post-Min 5)**
VL - con - Ave Power at PT (Pre)	33.30857	13.88368	1.000000	0.813180	0.841577	0.844567	0.644818	0.762322
VL - con - Ave Power - (Post-Min 0)	22.46000	8.65595	0.813180	1.000000	0.844051	0.738089	0.737005	0.731006
VL - con - Ave Power - (Post-Min 5)	26.63714	10.93489	0.841577	0.844051	1.000000	0.767381	0.726007	0.804886
VL - ecc - Ave Power at PT (Pre)	47.11029	18.69498	0.844567	0.738089	0.767381	1.000000	0.669345	0.755708
VL - ecc - Ave Power - (Post-Min 0)	26.66571	12.79168	0.644818	0.737005	0.726007	0.669345	1.000000	0.713632
VL - ecc - Ave Power - (Post-Min 5)	34.60571	14.89192	0.762322	0.731006	0.804886	0.755708	0.713632	1.000000

Appendix B.3: GLM Analyses for Time and Condition Effects, including Tukey Post Hoc Analyses

(a) Analyses for Endurance Time

Table 32: Analyses for Endurance Time (min)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 2.5277

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	2218.613	1	2218.613	347.2316	0.000000
Sex	17.521	1	17.521	2.7422	0.106935
Error	217.241	34	6.389		
EXERTION	84.019	2	42.009	34.7977	0.000000
EXERTION*Sex	0.389	2	0.194	0.1611	0.851561
Error	82.093	68	1.207		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 1.2072, df = 68.000

EXERTION	VL 3.2917	PS 5.2639	VF 5.0417
Varying Length		0.000112	0.000112
Pure Static	0.000112		0.668454
Varying Force	0.000112	0.668454	

(b) Analyses for Number of Movement Cycles

Table 33: Analyses for Number of movement cycles

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 58.1385

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	1146197	1	1146197	339.1026	0.000000
Sex	8356	1	8356	2.4720	0.125149
Error	114923	34	3380		
EXERTION	47298	2	23649	37.4486	0.000000
EXERTION*Sex	343	2	172	0.2720	0.762712
Error	42943	68	632		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 631.51, df = 68.000

EXERTION	Varying Length	Pure Static 120.17	**Varying Force**
Varying Length		0.000112	0.000112
Pure Static	0.000112		0.694349
Varying Force	0.000112	0.694349	

(c) Analyses for Peak Torque

Table 34: Analyses for Peak Torque (Nm.kg^{-1})

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.7522

Effect	**SS**	Degr. of Freedom	MS	F	p
Intercept	209.2921	1	209.2921	369.9447	0.000000
Error	19.2351	34	0.5657		
EXERTION	2.1318	3	0.7106	45.5078	0.000000
Error	1.5927	102	0.0156		
TIME	2.2194	2	1.1097	99.4412	0.000000
Error	0.7589	68	0.0112		
EXERTION*TIME	0.1729	6	0.0288	5.3305	0.000039
Error	1.1029	204	0.0054		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .01561, df = 102.00

EXERTION	Varying Length (Con)	Varying Length (Ecc)	Pure Static .75972	**Varying Force .76770**
Varying Length (Con)		0.000139	0.000139	0.000139
Varying Length (Ecc)	0.000139		0.014767	0.003596
Pure Static	0.000139	0.014767		0.966982
Varying Force	0.000139	0.003596	0.966982	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .01116, df = 68.000

TIME	Pre .80310	Post-Min 0 .62829	**Post-Min 5 .68635**
Pre		0.000112	0.000112
Post-Min 0	0.000112		0.000162
Post-Min 5	0.000112	0.000162	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .00541, df = 204.00

EXERTION	TIME	VL(Con) - Pre	VL(Con) - Min 0	VL(Con) - Min 5	VL(Ecc) - Pre	VL(Ecc) - Post Min 0	VL(Ecc) - Post Min 5	PS - Pre .83409	PS - Post Min 0	PS - Post Min 5	VF - Pre .83900	VF - Post Min 0	**VF - Post Min 5**
Varying Length (Con)	Pre		0.000018	0.000018	0.001048	0.000018	0.287039	0.000018	0.596933	0.995813	0.000018	0.955245	0.950723
Varying Length (Con)	Post-Min 0	0.000018		0.011067	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018
Varying Length (Con)	Post-Min 5	0.000018	0.011067		0.000018	0.000904	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018
Varying Length (Ecc)	Pre	0.001048	0.000018	0.000018		0.000018	0.000018	0.936282	0.000018	0.056711	0.820436	0.000018	0.164594
Varying Length (Ecc)	Post-Min 0	0.000018	0.000018	0.000904	0.000018		0.034131	0.000018	0.006884	0.000018	0.000018	0.000370	0.000018
Varying Length (Ecc)	Post-Min 5	0.287039	0.000018	0.000018	0.000018	0.034131		0.000018	0.999999	0.013521	0.000018	0.992627	0.003021
Pure Static	Pre	0.000018	0.000018	0.000018	0.936282	0.000018	0.000018		0.000018	0.000145	1.000000	0.000018	0.000822
Pure Static	Post-Min 0	0.596933	0.000018	0.000018	0.000018	0.006884	0.999999	0.000018		0.060145	0.000018	0.999931	0.016832
Pure Static	Post-Min 5	0.995813	0.000018	0.000018	0.056711	0.000018	0.013521	0.000145	0.060145		0.000048	0.331097	1.000000
Varying Force	Pre	0.000018	0.000018	0.000018	0.820436	0.000018	0.000018	1.000000	0.000018	0.000048		0.000018	0.000236
Varying Force	Post-Min 0	0.955245	0.000018	0.000018	0.000018	0.000370	0.992627	0.000018	0.999931	0.331097	0.000018		0.139375
Varying Force	Post-Min 5	0.950723	0.000018	0.000018	0.164594	0.000018	0.003021	0.000822	0.016832	1.000000	0.000236	0.139375	

Table 35: Analyses for Peak Torque (%change)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 18.5224

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	90266.54	1	90266.54	263.1074	0.000000
Error	11664.68	34	343.08		
EXERTION	10514.17	3	3504.72	17.3300	0.000000
Error	20627.86	102	202.23		
TIME	4365.98	1	4365.98	35.3389	0.000001
Error	4200.57	34	123.55		
EXERTION*TIME	224.65	3	74.88	0.9279	0.430169
Error	8231.70	102	80.70		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 202.23, df = 102.00

EXERTION	Varying Length (Con) 28.108	Varying Length (Ecc)	Pure Static 13.545	**Varying Force 12.720**
Varying Length (Con)		0.000261	0.000139	0.000139
Varying Length (Ecc)	0.000261		0.370039	0.207400
Pure Static	0.000139	0.370039		0.986089
Varying Force	0.000139	0.207400	0.986089	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 123.55, df = 34.000

TIME	Post - Min 0 21.904	Post - Min 5 14.006
Post - Min 0		0.000121
Post - Min 5	0.000121	

(d) Analyses for Fatigue Index

Table 36: Analyses for Fatigue Index

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 18.5224

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	90266.54	1	90266.54	263.1074	0.000000
Error	11664.68	34	343.08		
TIME	4365.98	1	4365.98	35.3389	0.000001
Error	4200.57	34	123.55		
EXERTION	10514.17	3	3504.72	17.3300	0.000000
Error	20627.86	102	202.23		
TIME*EXERTION	224.65	3	74.88	0.9279	0.430169
Error	8231.70	102	80.70		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 202.23, df = 102.00

Cell No.	EXERTION	**VL (con) 28.108**	VL (ecc) 17.448	PS 13.545	VF 12.720
1	Varying Length (concentric)		0.000261	0.000139	0.000139
2	Varying Length (eccentric)	0.000261		0.370039	0.207400
3	Pure Static	0.000139	0.370039		0.986089
4	Varying Force	0.000139	0.207400	0.986089	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 123.55, df = 34.000

Cell No.	TIME	**Post-Min 0 21.904**	Post-Min 5 14.006
1	Post-MIn 0		0.000121
2	Post-Min 5	0.000121	

(e) Analyses for Time to Peak Torque

Table 37: Analyses for Time to Peak Torque (sec)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 1.5425

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	716.6967	1	716.6967	301.2351	0.000000
Error	76.1342	32	2.3792		
EXERTION	256.2778	3	85.4259	89.3669	0.000000
Error	91.7665	96	0.9559		
TIME	4.5596	2	2.2798	5.1916	0.008139
Error	28.1045	64	0.4391		
EXERTION*TIME	2.3830	6	0.3972	0.7794	0.587033
Error	97.8417	192	0.5096		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .95590, df = 96.000

EXERTION	Varying Length (Con)	Varying Length (Ecc)	Pure Static 2.0247	**Varying Force**
Varying Length (Con)		0.011827	0.000140	0.000140
Varying Length (Ecc)	0.011827		0.000140	0.000140
Pure Static	0.000140	0.000140		0.422492
Varying Force	0.000140	0.000140	0.422492	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .43913, df = 64.000

TIME	Pre 1.3266	Post Min 0 1.2242	**Post Min 5 1.4851**
Pre		0.425919	0.135154
Post Min 0	0.425919		0.006121
Post Min 5	0.135154	0.006121	

Table 38: Analyses for Time to Peak Torque (% change from Pre)

Repeated Measures Analysis of Variance (Copy of Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 212.1720

Effect	**SS**	Degr. of Freedom	MS	F	p
Intercept	7233201	1	7233201	160.6772	0.000000
Error	1440543	32	45017		
EXERION	962659	3	320886	5.9093	0.000961
Error	5213027	96	54302		
TIME	400587	1	400587	26.5019	0.000013
Error	483694	32	15115		
EXERION*TIME	622909	3	207636	14.3924	0.000000
Error	1384970	96	14427		

Cell No.	Tukey HSD test; variable DV_1 (Copy of Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = 54302., df = 96.000 EXERION	**VL(con)** **267.86**	VL(ecc) 151.75	PS 119.97	VF 122.52
1	Varying Length (con)		0.026207	0.002499	0.003066
2	Varying Length (ecc)	0.026207		0.861912	0.888788
3	Pure Static	0.002499	0.861912		0.999922
4	Varying Force	0.003066	0.888788	0.999922	

Cell No.	Tukey HSD test; variable DV_1 (Copy of Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = 15115., df = 32.000 TIME	**Post-Min 0** **126.57**	Post-Min 5 204.48
1	Post-Min 0		0.000144
2	Post-Min 5	0.000144	

(f) Analyses for Work

Table 39: Analyses for Work (J)

Effect	Repeated Measures Analysis of Variance (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 45.9753 SS	Degr. of Freedom	MS	F	**p**
Intercept	547210.1	1	547210.1	258.8842	0.000000
Error	71866.7	34	2113.7		
EXERTION	6028.9	1	6028.9	38.8421	0.000000
Error	5277.4	34	155.2		
TIME	17847.8	2	8923.9	68.4273	0.000000
Error	8868.2	68	130.4		
EXERTION*TIME	460.8	2	230.4	3.8914	0.025112
Error	4026.3	68	59.2		

Tukey HSD test; variable DV_1 (Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = 155.22, df = 34.000 EXERTION	**VL(Con)** **45.689**	VL(Ecc) 56.405
Varying Length (Con)		0.000120
Varying Length (Ecc)	0.000120	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 130.41, df = 68.000

TIME	Pre 62.613	Post-Min 0 40.053	**Post-Min 5 50.474**
Pre		0.000112	0.000112
Post-Min 0	0.000112		0.000114
Post-Min 5	0.000112	0.000114	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 59.211, df = 68.000

EXERTION	TIME	VL(Con) - Pre	VL(Con) - Post Min 0	VL(Con) - Post Min 5	VL(Ecc) - Pre	VL(Ecc) - Post Min 0	**VL(Ecc) - Post Min 5**
Varying Length (Con)	Pre		0.000129	0.000129	0.000129	0.000129	0.999184
Varying Length (Con)	Post-Min 0	0.000129		0.002080	0.000129	0.008893	0.000129
Varying Length (Con)	Post-Min 5	0.000129	0.002080		0.000129	0.997102	0.000129
Varying Length (Ecc)	Pre	0.000129	0.000129	0.000129		0.000129	0.000129
Varying Length (Ecc)	Post-Min 0	0.000129	0.008893	0.997102	0.000129		0.000129
Varying Length (Ecc)	Post-Min 5	0.999184	0.000129	0.000129	0.000129	0.000129	

Table 40: Analyses for Work (%change)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 25.9058

Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	95241.10	1	95241.10	141.9159	0.000000
Error	22817.72	34	671.11		
EXERTION	4.05	1	4.05	0.0134	0.908610
Error	10296.87	34	302.85		
TIME	9921.27	1	9921.27	68.9160	0.000000
Error	4894.70	34	143.96		
EXERTION*TIME	268.72	1	268.72	1.8915	0.178030
Error	4830.40	34	142.07		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 143.96, df = 34.000

TIME	Post-Min 0 34.501	**Post-Min 5 17.664**
Post-Min 0		0.000120
Post-Min 5	0.000120	

(g) Analyses for Power

Table 41: Analyses for Power (W)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 29.1489

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	212332.4	1	212332.4	249.9040	0.000000
Error	28888.3	34	849.7		
EXERTION	3936.1	1	3936.1	60.0161	0.000000
Error	2229.8	34	65.6		
TIME	8713.9	2	4356.9	60.8063	0.000000
Error	4872.4	68	71.7		
EXERTION*TIME	818.2	2	409.1	12.8636	0.000018
Error	2162.7	68	31.8		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 65.583, df = 34.000

EXERTION	Varying Length (Con) 27.469	**Varying Length (Ecc) 36.127**
Varying Length (Con)		0.000120
Varying Length (Ecc)	0.000120	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 71.653, df = 68.000

TIME	Pre 40.209	Post-Min 0 24.563	**Post-Min 5 30.621**
Pre		0.000112	0.000112
Post-Min 0	0.000112		0.000306
Post-Min 5		0.000306	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 31.804, df = 68.000

EXERTION	TIME	VL(Con) - Pre	VL(Con) - Post Min 0	VL(Con) - Post Min 5	VL(Ecc) - Pre	VL(Ecc) - Post Min 0	**VL(Ecc) - Post Min 5**
Varying Length (Con)	Pre		0.000129	0.000194	0.000129	0.000199	0.928233
Varying Length (Con)	Post-Min 0	0.000129		0.032401	0.000129	0.030602	0.000129
Varying Length (Con)	Post-Min 5	0.000194	0.032401		0.000129	1.000000	0.000130
Varying Length (Ecc)	Pre	0.000129	0.000129	0.000129		0.000129	0.000129
Varying Length (Ecc)	Post-Min 0	0.000199	0.030602	1.000000	0.000129		0.000130
Varying Length (Ecc)	Post-Min 5	0.928233	0.000129	0.000130	0.000129	0.000130	

Table 42: Analyses for Power (%change from pre)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 24.3259

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	181553.3	1	181553.3	306.8072	0.000000
Error	20119.5	34	591.8		
EXERTION	71608.7	1	71608.7	123.9019	0.000000
Error	19650.2	34	577.9		
TIME	32987.9	1	32987.9	81.8094	0.000000
Error	13709.8	34	403.2		
EXERTION*TIME	273.6	1	273.6	0.5988	0.444393
Error	15537.0	34	457.0		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 577.95, df = 34.000

EXERTION	VL(Con) -13.40	**VL(Ecc) -58.63**
Varying Length (Con)		0.000120
Varying Length (Ecc)	0.000120	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 403.23, df = 34.000

TIME	Post-Min 0 -20.66	**Post-Min 5 -51.36**
Post-Min 0		0.000120
Post-Min 5	0.000120	

(h) Analyses for Electromyography - Amplitude

Table 43: Analyses for EMG Amplitude (mV)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.1774

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	8.960478	1	8.960478	284.6064	0.000000
Error	1.101931	35	0.031484		
EXERTION	0.735238	2	0.367619	30.2319	0.000000
Error	0.851197	70	0.012160		
TIME	0.472035	1	0.472035	90.3580	0.000000
Error	0.182842	35	0.005224		
EXERTION*TIME	0.070217	2	0.035108	13.4981	0.000011
Error	0.182069	70	0.002601		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .01216, df = 70.000

EXERTION	Varying Length	Pure Static .17333	**Varying Force**
Varying Length		0.000111	0.000111
Pure Static	0.000111		0.494085
Varying Force	0.000111	0.494085	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .00522, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.000123
Last 4 cycles	0.000123	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .00260, df = 70.000

EXERTION	TIME	VL - First 4 cycles	VL - Last 4 cycles	PS - First 4 cycles	PS - Last 4 cycles	VF - First 4 cycles	**VF - Last 4 cycles**
Varying Length	First 4 cycles		0.000128	0.000128	1.000000	0.000128	0.088305
Varying Length	Last 4 cycles	0.000128		0.000128	0.000128	0.000128	0.000128
Pure Static	First 4 cycles	0.000128	0.000128		0.000128	0.958999	0.002313
Pure Static	Last 4 cycles	1.000000	0.000128	0.000128		0.000128	0.101131
Varying Force	First 4 cycles	0.000128	0.000128	0.958999	0.000128		0.000230
Varying Force	Last 4 cycles	0.088305	0.000128	0.002313	0.101131	0.000230	

Table 44: Analyses for EMG Amplitude (%change)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 63.9175

Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	395295.8	1	395295.8	96.75718	0.000000
Error	142990.5	35	4085.4		
EXERTION	12082.0	2	6041.0	2.80859	0.067099
Error	150563.0	70	2150.9		

(i) Analyses for Electromyography – Power Frequency

Table 45: Analyses for EMG Median Frequency (Hz)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 12.3731

Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	861738.6	1	861738.6	5628.846	0.000000
Error	5358.3	35	153.1		
EXERTION	24740.4	2	12370.2	115.856	0.000000
Error	7474.1	70	106.8		
TIME	5670.8	1	5670.8	109.527	0.000000
Error	1812.1	35	51.8		
EXERTION*TIME	658.6	2	329.3	8.118	0.000675
Error	2839.7	70	40.6		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 106.77, df = 70.000

EXERTION	Varying Length	Pure Static 74.065	**Varying Force**
Varying Length		0.000111	0.000111
Pure Static	0.000111		0.000312
Varying Force	0.000111	0.000312	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 51.776, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.000123
Last 4 cycles	0.000123	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 40.567, df = 70.000

EXERTION	TIME	VL - First 4 cycles	VL - Last 4 cycles	PS - First 4 cycles	PS - Last 4 cycles	VF - First 4 cycles	**VF - Last 4 cycles**
Varying Length	First 4 cycles		0.000128	0.000128	0.000128	0.000128	0.001853
Varying Length	Last 4 cycles	0.000128		0.000128	0.000128	0.000128	0.000128
Pure Static	First 4 cycles	0.000128	0.000128		0.000724	0.002751	0.000128
Pure Static	Last 4 cycles	0.000128	0.000128	0.000724		0.998092	0.000130
Varying Force	First 4 cycles	0.000128	0.000128	0.002751	0.998092		0.000128
Varying Force	Last 4 cycles	0.001853	0.000128	0.000128	0.000130	0.000128	

Table 46: Analyses for EMG Median Frequency (%change)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 14.1705

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	24754.19	1	24754.19	123.2767	0.000000
Error	7028.07	35	200.80		
EXERTION	5943.03	2	2971.51	16.7035	0.000001
Error	12452.84	70	177.90		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 177.90, df = 70.000

EXERTION	Varying Length	Pure Static -7.607	**Varying Force**
Varying Length		0.000112	0.000511
Pure Static	0.000112		0.260082
Varying Force	0.000511	0.260082	

(j) Analyses for Performance Accuracy – Cycle time

Table 47: Analyses for Cycle Time for Varying Length & Varying Force (sec))

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.3589

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	29.02695	1	29.02695	225.3319	0.000000
Error	4.50865	35	0.12882		
EXERTION	0.48674	1	0.48674	4.1538	0.049154
Error	4.10127	35	0.11718		
TIME	0.22976	1	0.22976	2.0813	0.158006
Error	3.86381	35	0.11039		
EXERTION*TIME	0.22785	1	0.22785	1.9911	0.167050
Error	4.00508	35	0.11443		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .11718, df = 35.000

EXERTION	Varying Length	**Varying Force**
Varying Length		0.049257
Varying Force	0.049257	

(k) Analyses for Performance Accuracy – Force

Table 48: Analyses for Force (absolute deviation from target force (kg)) – Varying Length & Pure Static

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.5382

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	64.65132	1	64.65132	223.1933	0.000000
Error	10.13828	35	0.28967		
EXERTION	4.52918	1	4.52918	29.8202	0.000004
Error	5.31591	35	0.15188		
TIME	0.08351	1	0.08351	8.8222	0.005347
Error	0.33130	35	0.00947		
EXERTION*TIME	0.01938	1	0.01938	3.1274	0.085702
Error	0.21694	35	0.00620		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .15188, df = 35.000

EXERTION	Varying Length .84740	**Pure Static .49270**
Varying Length		0.000126
Pure Static	0.000126	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .00947, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.005485
Last 4 cycles	0.005485	

Table 49: Analyses for Force (%deviation from target force) – Varying Length & Pure Static

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 6.4878

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	42070.69	1	42070.69	999.5170	0.000000
Error	1473.19	35	42.09		
EXERTION	3546.50	1	3546.50	68.7752	0.000000
Error	1804.83	35	51.57		
TIME	36.66	1	36.66	4.8664	0.034042
Error	263.69	35	7.53		
EXERTION*TIME	21.59	1	21.59	3.9481	0.054796
Error	191.36	35	5.47		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 51.567, df = 35.000

EXERTION	Varying Length	**Pure Static**
Varying Length		0.000123
Pure Static	0.000123	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 7.5339, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.034148
Last 4 cycles	0.034148	

Table 50: Analyses for Force (%deviation from target force) – Varying Force

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 8.2474

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	16439.70	1	16439.70	241.6887	0.000000
Error	2380.71	35	68.02		
TIME	32.96	1	32.96	2.1616	0.150430
Error	533.68	35	15.25		
MAX/MIN	779.59	1	779.59	15.9834	0.000314
Error	1707.13	35	48.78		
TIME*MAX/MIN	26.52	1	26.52	1.9855	0.167635
Error	467.56	35	13.36		

Tukey HSD test; variable DV_1 (Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = 48.775, df = 35.000		
MAX/MIN	Max Force 8.3580	**Min Force 13.012**
Max Force		0.000424
Min Force	0.000424	

(l) *Analyses for Performance Accuracy – Goniometry*

Table 51: Analyses for Range of Motion (deviation from target angle (deg)) – Pure Static & Varying Force

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.9270

Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	536.5979	1	536.5979	624.3909	0.000000
Error	30.0788	35	0.8594		
EXERTION	232.8803	1	232.8803	256.4345	0.000000
Error	31.7852	35	0.9081		
TIME	1.3954	1	1.3954	6.2143	0.017552
Error	7.8589	35	0.2245		
EXERTION*TIME	0.6568	1	0.6568	3.1533	0.084471
Error	7.2899	35	0.2083		

Tukey HSD test; variable DV_1 (Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = .90815, df = 35.000		
EXERTION	Pure Static .65868	**Varying Force**
Pure Static		0.000123
Varying Force	0.000123	

Tukey HSD test; variable DV_1 (Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = .22454, df = 35.000		
TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.017669
Last 4 cycles	0.017669	

Table 52: Analyses for Range of Motion (deviation from target ROM (deg)) –Varying Length

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 12.8583

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	284016.1	1	284016.1	1717.809	0.000000
Error	5786.8	35	165.3		
TIME	476.5	1	476.5	24.237	0.000020
Error	688.1	35	19.7		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 19.660, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.000140
Last 4 cycles	0.000140	

Table 53: Analyses for Joint angles (deviation from target angles (deg)) –Varying Length

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 8.6985

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	18514.70	1	18514.70	244.6950	0.000000
Error	2648.26	35	75.66		
TIME	380.66	1	380.66	29.8338	0.000004
Error	446.57	35	12.76		
FLEX/EXT	1652.25	1	1652.25	17.1533	0.000207
Error	3371.30	35	96.32		
TIME*FLEX/EXT	262.10	1	262.10	19.8131	0.000083
Error	463.01	35	13.23		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 12.759, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.000126
Last 4 cycles	0.000126	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 96.323, df = 35.000

MAX/MIN	Flexion 7.9517	**Extension 14.726**
Flexion		0.000319
Extension	0.000319	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 13.229, df = 35.000

TIME	MAX/MIN	Flexion - First 4	Extension - First 4	Flexion - Last 4	**Extension - Last 4**
First 4 cycles	Flexion		0.000333	0.916436	0.000160
First 4 cycles	Extension	0.000333		0.001366	0.000160
Last 4 cycles	Flexion	0.916436	0.001366		0.000160
Last 4 cycles	Extension	0.000160	0.000160	0.000160	

(m) Analyses for Performance Accuracy – Accelerometry

Table 54: Analyses for Accelerations ($m.s^{-1}$) – Varying Length

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 3.9769

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	21066.35	1	21066.35	1332.009	0.000000
Error	553.54	35	15.82		
TIME	21.47	1	21.47	7.073	0.011723
Error	106.26	35	3.04		
MOVEMENT	268.14	1	268.14	13.061	0.000938
Error	718.52	35	20.53		
TIME*MOVEMENT	13.75	1	13.75	2.996	0.092305
Error	160.63	35	4.59		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 3.0360, df = 35.000

Cell No.	TIME	**First 4 cycles**	Last 4 cycles
1	First 4 cycles		0.011840
2	Last 4 cycles	0.011840	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 20.529, df = 35.000

Cell No.	MOVEMENT	**Flexion 13.460**	Extension 10.731
1	Flexion		0.001063
2	Extension	0.001063	

(n) Analyses for Cycle-to-Cycle Variability – Cycle Time

Table 55: Analyses for Cycle Time range (sec) for Varying Length & Varying Force

	Repeated Measures Analysis of Variance (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 0.3589				
Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	29.02695	1	29.02695	225.3319	0.000000
Error	4.50865	35	0.12882		
EXERTION	0.48674	1	0.48674	4.1538	0.049154
Error	4.10127	35	0.11718		
TIME	0.22976	1	0.22976	2.0813	0.158006
Error	3.86381	35	0.11039		
EXERTION*TIME	0.22785	1	0.22785	1.9911	0.167050
Error	4.00508	35	0.11443		

Tukey HSD test; variable DV_1 (Data for Statistica 2025) Approximate Probabilities for Post Hoc Tests Error: Within MSE = .11718, df = 35.000		
EXERTION	Varying Length	**Varying Force**
Varying Length		0.049257
Varying Force	0.049257	

Table 56: Analyses for Cycle Time standard deviation (sec) for Varying Length & Varying Force

	Repeated Measures Analysis of Variance (Data for Statistica 2025) Sigma-restricted parameterization Effective hypothesis decomposition; Std. Error of Estimate: 0.1585				
Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	5.798155	1	5.798155	230.9005	0.000000
Error	0.878887	35	0.025111		
EXERTION	0.086276	1	0.086276	3.6880	0.062982
Error	0.818772	35	0.023393		
TIME	0.038553	1	0.038553	1.7042	0.200259
Error	0.791784	35	0.022622		
EXERTION*TIME	0.039549	1	0.039549	1.6115	0.212653
Error	0.858945	35	0.024541		

Table 57: Analyses for Cycle Time coefficient of variation (%) for Varying Length & Varying Force

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 5.5501

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	8234.236	1	8234.236	267.3106	0.000000
Error	1078.140	35	30.804		
EXERTION	182.005	1	182.005	6.2141	0.017554
Error	1025.125	35	29.289		
TIME	50.217	1	50.217	1.8817	0.178873
Error	934.051	35	26.687		
EXERTION*TIME	44.608	1	44.608	1.5763	0.217606
Error	990.456	35	28.299		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 29.289, df = 35.000

EXERTION	Varying Length	**Varying Force**
Varying Length		0.017671
Varying Force	0.017671	

(o) Analyses for Cycle-to-Cycle Variability – Force

Table 58: Analyses for Force range (%deviation) for Varying Length & Pure Static conditions

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 3.2470

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	2115.971	1	2115.971	200.7027	0.000000
Error	368.998	35	10.543		
EXERTION	158.705	1	158.705	20.4797	0.000067
Error	271.229	35	7.749		
TIME	12.089	1	12.089	2.1688	0.149777
Error	195.091	35	5.574		
EXERTION*TIME	27.659	1	27.659	3.5008	0.069719
Error	276.530	35	7.901		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 7.7494, df = 35.000

EXERTION	Varying Length	**Pure Static**
Varying Length		0.000183
Pure Static	0.000183	

Table 59: Analyses for Force standard deviation (%deviation) for Varying Length & Pure Static conditions

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 1.4671

Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	420.8930	1	420.8930	195.5436	0.000000
Error	75.3349	35	2.1524		
EXERTION	30.4149	1	30.4149	19.5719	0.000090
Error	54.3904	35	1.5540		
TIME	2.7100	1	2.7100	2.3094	0.137577
Error	41.0709	35	1.1735		
EXERTION*TIME	5.4931	1	5.4931	3.3689	0.074948
Error	57.0694	35	1.6306		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 1.5540, df = 35.000

EXERTION	Varying Length	**Pure Static**
Varying Length		0.000205
Pure Static	0.000205	

Table 60: Analyses for Force coefficient of variation (%) for Varying Length & Pure Static conditions

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.0922

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	2.098283	1	2.098283	246.8140	0.000000
Error	0.297552	35	0.008501		
EXERTION	0.494831	1	0.494831	61.3044	0.000000
Error	0.282509	35	0.008072		
TIME	0.004635	1	0.004635	0.7170	0.402869
Error	0.226245	35	0.006464		
EXERTION*TIME	0.007148	1	0.007148	0.9790	0.329242
Error	0.255545	35	0.007301		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = .00807, df = 35.000

EXERTION	Varying Length	**Pure Static**
Varying Length		0.000123
Pure Static	0.000123	

Table 61: Analyses for Force range (%deviation) for Varying Force condition

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 5.4646

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	8909.230	1	8909.230	298.3449	0.000000
Error	1045.176	35	29.862		
TIME	2.373	1	2.373	0.0687	0.794734
Error	1208.674	35	34.534		
MAX/MIN	387.179	1	387.179	19.2909	0.000099
Error	702.470	35	20.071		
TIME*MAX/MIN	31.893	1	31.893	0.9524	0.335796
Error	1171.985	35	33.485		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 20.071, df = 35.000

MAX/MIN	Max Force 6.2260	**Min Force 9.5055**
Max Force		0.000214
Min Force	0.000214	

Table 62: Analyses for Force standard deviation (%deviation) for Varying Force condition

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 2.4253

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	1780.495	1	1780.495	302.6908	0.000000
Error	205.878	35	5.882		
TIME	0.776	1	0.776	0.1166	0.734775
Error	232.884	35	6.654		
MAX/MIN	85.151	1	85.151	21.8469	0.000043
Error	136.417	35	3.898		
TIME*MAX/MIN	7.869	1	7.869	1.2424	0.272603
Error	221.684	35	6.334		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 3.8976, df = 35.000

MAX/MIN	Max force 2.7473	**Min force 4.2853**
Max Force		0.000160
Min Force	0.000160	

Table 63: Analyses for Force coefficient of variation (%) for Varying Force condition

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 16.0776

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	51373.94	1	51373.94	198.7475	0.000000
Error	9047.10	35	258.49		
TIME	52.54	1	52.54	0.3779	0.542687
Error	4866.04	35	139.03		
MAX/MIN	49114.02	1	49114.02	189.5509	0.000000
Error	9068.75	35	259.11		
TIME*MAX/MIN	58.77	1	58.77	0.4208	0.520748
Error	4887.52	35	139.64		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 259.11, df = 35.000

MAX/MIN	Max Force 37.356	**Min Force .42011**
Max Force		0.000123
Min Force	0.000123	

(p) Analyses for Cycle-to-Cycle Variability – Goniometry

Table 64: Analyses for Joint angle range (%deviation) for Varying Length condition

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 3.2730

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	2684.103	1	2684.103	250.5604	0.000000
Error	374.934	35	10.712		
TIME	0.008	1	0.008	0.0016	0.968229
Error	182.749	35	5.221		
MOVEMENT	189.292	1	189.292	31.4983	0.000003
Error	210.336	35	6.010		
TIME*MOVEMENT	9.353	1	9.353	1.2519	0.270807
Error	261.494	35	7.471		

Table 65: Analyses for Joint angle standard deviation (%deviation) for Varying Length condition

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 1.4564

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	544.6425	1	544.6425	256.7877	0.000000
Error	74.2344	35	2.1210		
TIME	0.0213	1	0.0213	0.0204	0.887114
Error	36.5292	35	1.0437		
MOVEMENT	38.1160	1	38.1160	31.1596	0.000003
Error	42.8138	35	1.2233		
TIME*MOVEMENT	2.3826	1	2.3826	1.5619	0.219684
Error	53.3927	35	1.5255		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 1.2233, df = 35.000

MOVEMENT	Flexion 2.4593	**Extension 1.4303**
Flexion		0.000125
Extension	0.000125	

Table 66: Analyses for Joint angle coefficient of variation (%) for Varying Length condition

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 28.4085

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	132242.6	1	132242.6	163.8609	0.000000
Error	28246.5	35	807.0		
TIME	3269.9	1	3269.9	7.1707	0.011207
Error	15960.3	35	456.0		
MOVEMENT	26644.4	1	26644.4	33.3404	0.000002
Error	27970.8	35	799.2		
TIME*MOVEMENT	21.0	1	21.0	0.0308	0.861772
Error	23932.3	35	683.8		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 456.01, df = 35.000

TIME	First 4 cycles	**Last 4 cycles**
First 4 cycles		0.011326
Last 4 cycles	0.011326	

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 799.16, df = 35.000

MOVEMENT	Flexion 43.907	**Extension 16.702**
Flexion		0.000124
Extension	0.000124	

Table 67: Analyses for Joint angle range (%deviation) for Pure Static & Varying Force conditions

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.6596

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	102.5156	1	102.5156	235.6391	0.000000
Error	15.2269	35	0.4351		
EXERTION	0.0117	1	0.0117	0.0256	0.873861
Error	16.0608	35	0.4589		
TIME	1.3417	1	1.3417	2.6606	0.111832
Error	17.6508	35	0.5043		
EXERTION*TIME	0.4556	1	0.4556	0.7233	0.400839
Error	22.0469	35	0.6299		

Table 68: Analyses for Joint angle standard deviation (%deviation) for Pure Static & Varying Force conditions

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.3105

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	21.23152	1	21.23152	220.2211	0.000000
Error	3.37435	35	0.09641		
EXERTION	0.02193	1	0.02193	0.2185	0.643099
Error	3.51241	35	0.10035		
TIME	0.31158	1	0.31158	2.8626	0.099547
Error	3.80960	35	0.10885		
EXERTION*TIME	0.12283	1	0.12283	0.9262	0.342458
Error	4.64162	35	0.13262		

Table 69: Analyses for Joint angle coefficient of variation (%) for Pure Static & Varying Force conditions

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 17.8233

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	157024.1	1	157024.1	494.2996	0.000000
Error	11118.4	35	317.7		
EXERTION	55026.0	1	55026.0	170.9891	0.000000
Error	11263.3	35	321.8		
TIME	1295.2	1	1295.2	2.6263	0.114082
Error	17260.0	35	493.1		
EXERTION*TIME	1997.5	1	1997.5	3.4872	0.070237
Error	20048.1	35	572.8		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 321.81, df = 35.000

EXERTION	PS 52.570	**VF 13.474**
Pure Static		0.000123
Varying Force	0.000123	

(q) Analyses for Cycle-to-Cycle Variability – Accelerometry

Table 70: Analyses for Acceleration Variability for Varying Length flexion (concentric) and extension (eccentric) – Range (sec)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 2.1304

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	1935.855	1	1935.855	426.5348	0.000000
Error	158.850	35	4.539		
TIME	2.224	1	2.224	0.3509	0.557405
Error	221.857	35	6.339		
MOVEMENT	108.312	1	108.312	20.4223	0.000068
Error	185.626	35	5.304		
TIME*MOVEMENT	0.839	1	0.839	0.1442	0.706405
Error	203.656	35	5.819		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 5.3036, df = 35.000

MOVEMENT	Flexion 4.5338	**Extension 2.7993**
Flexion		0.000184
Extension	0.000184	

Table 71: Analyses for Acceleration Variability for Varying Length flexion (concentric) and extension (eccentric) – Standard deviation (sec)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 0.9389

Effect	SS	Degr. of Freedom	MS	F	p
Intercept	384.5484	1	384.5484	436.2723	0.000000
Error	30.8504	35	0.8814		
TIME	0.5027	1	0.5027	0.4078	0.527255
Error	43.1473	35	1.2328		
MOVEMENT	20.9050	1	20.9050	21.7766	0.000044
Error	33.5992	35	0.9600		
TIME*MOVEMENT	0.2122	1	0.2122	0.1872	0.667930
Error	39.6874	35	1.1339		

Tukey HSD test; variable DV_1 (Data for Statistica 2025)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 124.55, df = 35.000

MOVEMENT	Flexion 15.798	**Extension 12.156**
Flexion		0.058393
Extension	0.058393	

Table 72: Analyses for Acceleration Variability for Varying Length flexion (concentric) and extension (eccentric) – Coefficient of Variation (%)

Repeated Measures Analysis of Variance (Data for Statistica 2025)
Sigma-restricted parameterization
Effective hypothesis decomposition; Std. Error of Estimate: 8.6005

Effect	SS	Degr. of Freedom	MS	F	**p**
Intercept	28132.19	1	28132.19	380.3301	0.000000
Error	2588.87	35	73.97		
TIME	116.26	1	116.26	1.4476	0.236985
Error	2810.77	35	80.31		
MOVEMENT	477.36	1	477.36	3.8327	0.058273
Error	4359.24	35	124.55		
TIME*MOVEMENT	139.50	1	139.50	1.6764	0.203882
Error	2912.48	35	83.21		