The acute and chronic effects of resistance exercise with and without blood flow restriction on fall risk factors in older adults

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Abstract

Blood flow restriction (BFR) is a novel training method in which periods of mild ischemia are applied to the upper thighs using specialized tourniquets, often combined with low-intensity resistance exercise (LI). Anecdotally, this training method alters a series of haemodynamic, metabolic, mechanical, and neurological pathways mimicking the effects observed in traditional high-intensity resistance exercise (HI). However, older adults (60≥ years) adhering to HI resistance exercise can be challenging as this type of training requires a high mechanical load to develop strength, hypertrophy, and neuromuscular adaptations. The importance of maintaining lower limb strength in older adults is evident within the literature as there is a strong link between lower limb strength and fall risk. The purpose of this doctoral thesis is to explore the acute and chronic effects of BFR training through a series of pilot studies on measures associated with postural control. Chapter 4 examined the acute responses to BFR and no-blood flow restriction (NBFR) conditions on quiet standing (double leg and single leg) and gait stability (walking) after performing a fatiguing exercise protocol to either the ankle or knee in healthy young adults. There were minimal changes to quiet standing and gait stability post-exercise for either the ankle or knee irrespective of whether BFR or NBFR. However, muscle torque activation reduced considerably for the tibialis anterior and vastus lateralis after muscle fatigue for the ankle and knee respectively. As there were minimal changes in postural control for both quiet standing and gait stability in young healthy adults, it was of interest to identify whether changes would be similar in older adults. Therefore, Chapter 5 repeated the methods conducted in Chapter 4 with a focus on older adults. The results observed in older adults were similar to those in young adults. There were minimal changes to quiet standing and in gait stability after exercise, regardless of whether BFR or NBFR was applied to the ankle or knee. Although, muscle torque and activation reduced considerably for the tibialis anterior and vastus medialis after muscle fatigue for the ankle and knee, respectively. Collectively, the two acute pilot studies provided crucial information on the effects of postural control and can confirm that when fatiguing either the ankle or knee musculature, there were minimal changes to balance. The findings from Chapters 4 and 5 suggest BFR could be safely applied in young and older adults as there appeared to be minimal effect on postural control. Therefore, combining low-intensity resistance exercise with BFR (LIBFR) may be a suitable training method for clinicians or rehabilitation specialists for early-stage rehabilitation or part of a fall prevention programme in older adults. Chapter 6 investigated the effects of a 6-week LIBFR (35% ~ 1 repetition maximum [1RM]) or HI [high intensity ~ 70% 1RM) resistance programme using the box squat and calve raise exercises to examine a host of quiet standing, gait stability and functional balances tests in addition to neuromuscular, proprioception and indices of strength measured at baseline, mid and post 6-week testing points. The results from this study showed an increase in squat and calf strength post training with an increase in knee extension maximal voluntary isometric contraction (MVIC). The functional tests comprising of the timed up and go, sit-to-stand for time and power, also led to some improvement post-training. Additionally, the error rate for joint position sense and the ability to match strength (force sense) also improved, suggesting that participants improved their accuracy post-training. However, negligible changes in quiet standing and gait stability were observed in both LIBFR and HI groups. It appears that LIBFR exercise could be a good alternative to HI resistance exercise based on the improvements noted in strength and neuromuscular measures over a 6-week training period. However, further investigation is required to determine this point as there were several limitations to each pilot study regarding sample size. Although, despite the low sample size it appears that LIBFR training does not cause any detrimental effects on either quiet standing or gait stability, regardless of whether it is performed in an acute or chronic setting. This type of exercise presents an excellent opportunity to motivate sedentary older adults to engage in resistance training at a low-intensity level, which could lead to better adherence to resistance training programmes and overall physical activity.

Presentation

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Abbreviations

1RM ....One-Repetition Maximum.

ADL................................Activities of Daily Living.

ANOVA ....Analysis of Variance.

AOP ....Arterial Occlusion Pressure.

BFR: ....Blood Flow Restriction.

CR10: ....Category-Ratio 10.

COP:...............................Centre of Pressure.

COP_{AP}:............................Centre of Pressure – Anterior-Posterior.

COP_{ML}: ....Centre of Pressure – Medio-Lateral.

COP_{VL}:............................Centre of Pressure – Velocity.

CV..................................Coefficient of Variation.

DBP ....Diastolic Blood Pressure.

DLS:...............................Double leg stance.

DS:.................................Double Support.

DOMS ....Delayed Onset of Muscle Soreness.

EC..................................Eyes Closed.

EMG: ....Electromyography.

EMG_{GM}............................Electromyography – Gastrocnemius (Medial Head).

EMG_{TA} ....Electromyography – Tibialis Anterior.

EMG_{VM}............................Electromyography – Vastus Medialis.

EMG_{VL} ....Electromyography – Vastus Lateralis.

EO ....Eyes Open

EXT:...............................Extension and/ or extensor muscle.

DBP: ....Diastolic Blood Pressure.

FLEX: ....Flexion and/ or Flexor muscle.

FS10 ....Force 10% maximum voluntary isometric contraction.

FS20 ....Force 20% maximum voluntary isometric contraction.

g.....................................Acceleration due to gravity.

GM:................................Gastrocnemius (Medial Head).

HI ....High intensity resistance exercise.

ICC ....Intraclass correlation coefficient.

IGF-1: ....Insulin Like Growth Factor – Isoform-1.

JPS:................................Joint Position Sense.

LI.....................................Low Intensity.

LIBFR ....Low Intensity resistance exercise with Blood Flow Restriction.

MAPK: ....Mitogen Activated Protein Kinase.

MVC:................................Maximal Voluntary Contraction.

MVIC: ....Maximal Voluntary Isometric Contraction.

MVC_{ISOK}:..........................Maximal Voluntary Contraction Isokinetic.

mmHg:.............................Millimetres of Mercury.

mTOR:.............................Mechanistic Target of Rapamycin.

NBFR:..............................No Blood Flow Restriction.

NHS:................................National Health Service.

PF:...................................Plantarflexors.

ROM: ....Range of motion.

RPE: ....Rating of Perceived Exertion.

VAS: ....Visual Analogue Score.

VL: ....Vastus Lateralis.

VM ....Vastus Medialis.

TA:...................................Tibialis Anterior.

TAN_L................................Tandem Stance – Left Leg.

TAN_R................................Tandem Stance – Right Leg.

TUG:................................Timed up and go.

SBP: ....Systolic Blood Pressure.

SD:...................................Standard Deviation.

SL: ....Stride Length.

SLS_L: ....Single leg stance – Left Leg.

SLS_R: ....Single leg stance – Right Leg.

SPSS ....Statistical Package for the Social Sciences.

ST:...................................Stride Time.

STS:.................................Sit to Stand.

STS^P: ....Sit-to-Stand – Power.

STS^T: ....Sit-to-Stand – Time.

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Chapter 1: Introduction

Falls in older adults continues to be the leading cause of injury-related hospitalisation and the reason for reducing independence, quality of life and the increase in injury-related mortality rates (Stevens *et al.,* 2006; Sturnieks *et al*., 2018). Aside from the human cost, falls among older adults places a considerable financial strain on public health services (Scuffham, *et al*., 2003; Heath *et al*., 2007). The soaring financial cost is an ongoing issue for the NHS, with an estimated annual expenditure of £4.4 billion to treat fall-related injuries (Public Health England, 2017). As it stands, the current population of older adults in the UK is projected to surpass 20 million by 2030 (Age UK, 2019) and is due to a continual increase in the number of individuals aged 65 and above in developed countries and longer life expectancies (Christenson et al., 2009; Marmot, 2020). Consequently, it has become more important to focus on preventing and reducing the risk of falls in older adults.

The risk of falls in older adults is a major health concern and has been associated with changes in several risk factors (Ganz *et al.,* 2007; Fabre *et al.,* 2010; Ambrose *et al*., 2013; Bull *et al*., 2020; Jehu *et al*., 2021; Sherrington *et al*., 2021). Although some risk factors (chronological age, sex, and ethnicity) are non-modifiable, several physiological risk factors can be modified with appropriately targeted exercise interventions. For example, age-related reductions in muscle strength/mass and balance, which contribute to an increase in fall risk, can be effectively targeted to reduce the risk of falls. Indeed, the National Institute for Health and Care Excellence guidelines (NICE, 2013) recommend that strength and balance training should be offered as first-line therapy to prevent falls. Accordingly, there is a robust Cochrane meta-analytical review level one A evidence (108 randomised control trials) that combining balance and resistance exercises as a single intervention reduces the rate of falls in community-dwelling older adults by 23% (Sherrington *et al*., 2019).

It is now firmly established that strength (ability to produce maximal force) and power (the rate of force produced at the onset of a muscular contraction [force x velocity]) (Hess and Woollacott, 2005; Goodpaster *et al*., 2006; Van Roie *et al*., 2013; Guizelini *et al.,* 2018) play an important role in many activities of daily living (ADL) (Aagaard *et al.,* 2010). Alterations in specific physiological pathways can contribute to the steady decline in strength and power (Reid and Fielding, 2012), such as muscle atrophy (sarcopenia) (Doherty, 2003; Aagaard *et al*., 2010), low muscle quality (Cruz-Jentoft and Sawyer, 2019), a reduction in the size/ number of muscle fibres (Goodpaster *et al*., 2006) which are crucial for balance control (Deschenes, 2004; Paillard, 2012). Postural control is widely defined as the ability to maintain the body's centre of gravity over the base of support and is defined as balance (Pollock *et al*., 2000). Traditional conceptualisations view the control of an upright stance as a closed-loop feedback process via sensory inputs (somatosensory, visual, and vestibular) to the central nervous system (Rogers *et al*., 2001). Sarcopenia is one condition that tends to affect the important postural muscles of the lower limb responsible for maintaining postural control (e.g., quadriceps extensors and ankle plantarflexors) (Tokuno *et al.,* 2007; Mignardot *et al*., 2015). Therefore, preserving strength and preventing muscle atrophy is essential in older adults.

Traditionally, high-intensity (HI) resistance exercise is prescribed to preserve muscle mass and function in older adults (Moreland *et al.,* 2004; Orr *et al.,* 2006; Yardley *et al*., 2007; Penzer, *et al*., 2015). Although extremely potent (Yardley *et al*., 2007; Penzer *et al*., 2015), many older adults find this type of exercise strenuous and therefore, adherence is often extremely low (Ambrose *et al*., 2013; Jehu *et al*., 2021). This limits the achievement of a sufficient exercise dose and subsequently reduces the effectiveness of this approach to maintaining muscle strength and mass. In recent years, a novel approach is to prescribe low-intensity resistance exercise with blood flow restriction (Takarada *et al.,* 2000a; Kubota *et al*., 2008; Wernbom *et al*., 2008; Manini and Clark, 2009; Loenneke and Pujol, 2009 Fry *et al.,* 2010; Patterson and Ferguson, 2011; Loenneke *et al*., 2012a; Abe *et al.,* 2012 Yasuda *et al*., 2014; Baker *et al.,* 2020).

Low-intensity resistance exercise (~20% - 50% 1RM) combined with blood flow restriction (LIBFR) is a relatively new modality which mimics the physiological effects and responses gained from HI resistance exercise (Wernbom *et al*., 2008; Manini and Clark, 2009; Loenneke and Pujol, 2009; Loenneke *et al*., 2012a; Abe *et al.,* 2012; Baker *et al.,* 2020). Blood flow restriction (BFR) in its purest form is the application of a specially made tourniquet (or cuff) which is applied over an area of interest (upper arm or proximal thigh) to reduce venous return but not to occlude the artery of the target limb (Loenneke *et al.,* 2013; Mattocks *et al.,* 2017). The tourniquet is inflated to either an absolute (e.g., millimetres of mercury - mmHg) or relative (e.g., 40% of an individual resting systolic blood pressure) systolic blood pressure by either an automated or manual sphygmomanometer (Downs *et al*., 2014; Cook *et al*., 2017; Kjeldsen *et al*., 2019).

The reduction of blood flow (ischemia) creates a hypoxic environment to the exercising muscle by reducing the delivery of oxygenated blood that increases the rates of key metabolites (e.g., hydrogen ions, blood lactate) responsible for inducing muscle fatigue at a similar rate to HI resistance exercise (Wernbom *et al*., 2008). The early onset of muscle fatigue during LIBFR exercise has been reported to stimulate higher-order motor units and increase the recruitment of fast twitch muscle fibres (Karabulut and Perez, 2013, Kjeldsen *et al*., 2019). Moreover, the LI used in BFR exercise is equivalent to normal activities of daily living (10 – 30% working capacity, Abe *et al.,* 2006). Studies observing the acute and chronic effects of LIBFR in older adults have focussed on hypertrophy and strength adaptions (Takarada *et al.,* 2000a; Kubota *et al*., 2008; Fry *et al.,* 2010; Patterson and Ferguson, 2011; Yasuda *et al*., 2014). Yet, few studies have looked at functional changes related to quiet standing and gait stability in young or older adults as post-operative, rehabilitation, and dynamic balance (e.g., Time up and go) have previously been reviewed (Yokokawa *et al.,* 2008; Hylden *et al*., 2015; Ferraz *et al*., 2018; Linero *et al.,* 2021). Given the lack of information on LIBFR on postural control, this is a clear void to fill for gaining a more detailed understanding of the potential effects this alternate type of exercise may have in older adults.

Although resistance exercise is highly effective in reducing falls, it can also lead to short-term muscle fatigue which may increase fall risk (e.g., iatrogenic falls: Frels *et al*., 2002). There is a considerable body of work showing that muscle fatigue induced by repeated contractions (e.g., ankle and knee) can cause transient impairment in postural control in young and older adults (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Baudry, *et al*., 2007; Lin *et al*., 2009; Bisson *et al*., 2010; Papa *et al*., 2015). Hence, there is a need to establish not only the chronic effects of LIBFR but also the potential consequence of acute exposure to this new approach to resistance exercise on postural control.

1.1. Aims and Objectives

The aim of this thesis is to determine whether the use of LIBFR could be used to improve quiet standing balance and gait stability in older adults. This thesis will look to achieve this by following the 3 objectives:

- Establish a healthy able-bodied model by examining the acute effects of lower limb muscle fatigue with and with no BFR (NBFR) exercise on quiet standing balance and gait stability in young adults.
- Explore the acute effects of lower limb muscle fatigue with and with NBFR exercise on quiet standing and gait stability in older healthy adults.
- To examine the effects of a chronic exposure to LIBFR and HI resistance exercise with NBFR in older adults on muscular strength, proprioception, postural sway, and gait stability.

1.2. Thesis Structure

This thesis was designed to explore the acute and chronic effects of LIBFR and with NBFR through a series of pilot studies on postural control and gait stability (e.g., intrinsic risk factors for falls). This idea was developed from previous research on LIBFR in older adults, which has shown that this training technique can lead to increased muscular strength and hypertrophy. The framework of this thesis started with gaining a better understanding of the area through a comprehensive literature review (Chapter 2) exploring postural control, gait stability and the responses to resistance exercise with and with NBFR. This was followed by developing a general method highlighting the common methodologies to be adopted in the main pilot study chapters. Chapters 4 and 5 observed the effects of muscle fatigue with or with NBFR in young (Chapter 4) and older (Chapter 5) adults to determine if there are any acute changes to the characteristics of postural control and gait stability. The final pilot study explores the chronic effects of LIBFR compared with HI resistance exercise (NBFR) in older adults (Chapter 6) for strength and proprioception in addition to postural measuring the characteristics of postural control and gait stability. The last Chapter (Chapter 7) synthesises the findings of the thesis, highlights the limitations, and provides recommendations for future research within the field of LIBFR.

Chapter 2: Literature Review

2.1. Postural Control

Postural control is classically defined as the interaction and maintenance of balance (in relation to the centre of gravity and base of support) and posture (orientation of the body's segments to the environment) to maintain stability during quiet standing or dynamic movements (Woollacott, 2000). The relationship between stability, the base of support, the line of gravity and the centre of gravity enables humans to maintain balance during quiet standing and dynamic scenarios (Pollock *et al*., 2000). Humans can maintain an upright position despite a high centre of gravity and a small base of support (Winter, 1995; Pollock *et al*., 2000). Given that the human upright stance is intrinsically unstable, in quiet standing, people naturally sway to compensate for the small postural deviations (Horak, 2006). Sheldon (1963) classically describes "sway" as a phenomenon whereby a small deviation in movement occurs constantly as the body attempts to adjust and correct these movements. Although the origins of postural sway are not fully understood, Peterka and Loughlin (2002) propose that the small deviations in movement occur due to an interplay of gravity-induced torque and correction torque acting on the body. Consequently, measures of postural sway during quiet standing are often used to characterise postural control (Prieto *et al*., 1996).

2.2. Sensory Mechanisms

Maintaining postural control is a complex task that requires the central nervous system to integrate and process sensory (afferent) information for the central nervous system to produce coordinated musculoskeletal responses based on the sensory information provided (Riemann *et al*.,1999). Bottom-up sensory information (somatosensory, proprioception, vestibular, and visual systems) is used to detect modifications in movement patterns and respond accordingly through a closed feedback loop (Runge *et al*., 1999; Riemann and Lephart, 2002a). The information processed by the brainstem and cerebellum provides a motor (efferent) response allowing voluntary movement to occur (Lepers, 1997).

When a human adopts an upright position, there are constant adjustments produced through gravity-induced torque acting on the body, causing instability (Perterka and Loughlin, 2002). These constant adjustments to control postural sway have been classed as "sensory re-weighing" (Nashner and Berthoz, 1978), whereby sensory information is processed through a closed feedback control mechanism dependent on the environmental conditions (Nashner and Berthoz, 1978; Horak, 2006). For example, when an individual begins to walk from a quiet standing position, there is continuous feedback to the central nervous system through sensory input to initiate the movement (Reimann and Lephart, 2002a, Ivanenko and Gurfinekl, 2018). Moreover, the sensory information received by the central nervous system is essential for postural tone, which is derived from other sensory inputs such as the somatosensory, proprioception, visual and vestibular systems (Riemann and Lephart, 2002a; Paillard, 2012; Han *et al*., 2016). It is the sensory inputs and the role of these important systems (e.g., proprioception; mechanoreceptors derived from the cutaneous surface of the foot) which contribute towards postural control (Kelly *et al*., 2012; Ridge *et al*., 2022). Therefore, the body is dually tasked by simultaneously distributing muscle activity (postural tone) in response to either an internal or external perturbation to provide the necessary feedback to maintain postural control (Ivanenko and Gurfinkel, 2018).

2.3. Motor Mechanisms

Motoric responses are necessary to provide coordinated and corrective torques to help maintain stability (Woollacott, 2000). The sensory information is processed by motor systems involved in higher-level planning, force production and coordinated movements when an individual attempt to move a limb or make a postural adjustment based on the information received from the environment (Riemann and Lephart, 2002a). The afferent information derived from the joint and muscle receptors (mechanoreceptors) is required for the planning stages of a movement and is essential for determining the body's position in space (proprioception) (Riemann and Lephart, 2002b; Han *et al*., 2016). An efferent response is conveyed in the brain stem and cerebellum sending a modulated or regulated response through the descending alpha and gamma motor neurons (facilitated by the spinal cord)

(Riemann and Lephart, 2002b). The motoric response is produced from this sequence of events, and the summation of a voluntary muscular contraction through the recruitment of motor units to reproduce or match a given force. In this context, if postural control is disturbed (e.g., sudden perturbation), then the muscles may contract in a different way to restore or maintain postural control (Riemann and Lephart, 2002b). Therefore, the contribution of muscle tone (Massion, 1994; Ivanenko and Gurfinkel, 2018) postural muscle tone (Takakusaki *et al.,* 2003) and intrinsic stiffness (Winter *et al*., 1998; Loram and Lakie, 2002) are three components recognised for ensuring postural control is maintained during standing and locomotion.

Normal quiet standing requires a level of voluntary muscle contraction and muscle tone, which refers interchange between to the force produced and resistance to the muscle lengthening (Massion, 1994; Ivanenko and Gurfinkel, 2018; Wright, 2019). As seen during quiet standing, muscle tone is integral to the maintenance of postural control where there is a fluctuation between muscle tone and muscle stiffness to help to stabilse the body (e.g., postural sway) (Massion, 1994; Takakusaki *et al.,* 2003; Paillard and Borel, 2013; Ivanenko and Gurfinkel, 2018). Furthermore, the constant level of muscle tone is often categorised into phasic (voluntary muscle activity) and tonic (involuntary muscle activity; anti-gravity) muscle groups. In the lower limb, there are several tonic muscles recruited during quiet standing (e.g., gastrocnemius, tibialis anterior), and when challenged (e.g., muscle fatigue trials or perturbation), there is a change in muscle tone and recruitment pattern to prevent a sudden perturbation (Riemman and Lephart, 2002b; Paillard, 2012; Wright, 2019). A change in muscle tone and recruitment pattern in the lower limb is dependent on the environment involving a strategy at either the ankle, knee, or hip to compensate or prevent movement. Diener *et al.* (1988) identified a distal to proximal relationship in muscle activation of the lower limb when postural was challenged. The results from the study identified the gastrocnemius responded quickest, closely by the hamstrings and paraspinal muscles. More recently, Donath *et al*. (2016) refers to both young and older adults having a similar distal to proximal muscle activation pattern when performing simple and challenging balance task (eyes open vs. closed; double leg vs. single leg stance). The authors reported a change in muscle recruitment pattern, with young adults recruiting the gastrocnemius compared to the older adults with the tibialis anterior. The findings suggest that a change in muscle recruitment patterns following balance tasks are largely age and balance task dependent, and may contribute to the change in muscle stiffness, tone, and contraction to help maintain postural control.

2.3.1. Intrinsic Muscle Stiffness

Intrinsic muscle stiffness is a change in mechanical stiffness derived from active musculotendinous structures (Loram and Lakie, 2002). Maintaining an upright double-leg stance position requires a degree of intrinsic muscle stiffness predominantly at the ankle (Winter *et al*., 2001; Loram and Lakie, 2002, Vlutters *et al.,* 2015). The musculotendinous structures of the foot help to maintain stability and prevent the risk of a fall (Winter *et al*., 2001; Loram and Lakie, 2002, Vlutters *et al.,* 2015) and has been suggested by Loram and Lakie, (2002) that stiffness generated at the ankle is not neurally controlled but rather biomechanical. The contribution of muscle stiffness from musculotendinous structures of the ankle has been referred to as a “spring” like mechanism where the stiffness of the Achilles tendon and gastrocnemius aid in supporting double limb standing (Winter *et al*., 2001). Moreover, muscle stiffness at the ankle may contribute to stability at the ankle when postural sway is suddenly challenged through the recruitment of specific tonic and phasic muscles of the musculoskeletal system (Vlutters *et al.,* 2015). Therefore, it could be assumed that postural control is maintained by (1) passive ankle stiffness derived from the musculature and (2) a relationship between distal to proximal tonic muscles initiated during double leg standing.

2.4. Movement Strategies

2.4.1. Ankle Strategy

The ankle strategy is adopted to control postural sway when there is little external influence (e.g., double leg stance in quiet standing). According to Gatev *et al.* (1999), the ankle strategy is adopted when a small and slow unexpected disturbance in postural control occurs and is characterised by a swaying movement at the ankles (Sakanaka *et al*., 2021). In addition, the centre of mass is repositioned along the sagittal plane and is often referred to as a single-segment inverted pendulum (Nashner and

McCollum, 1985; Winter *et al*., 1988; Runge *et al.,* 1999; Gage *et al*., 2004). As the centre of mass sits approximately 5cm in front of (anteroposterior direction) the base of support (Iqbal, 2011), a constant co-contraction of the dorsiflexors (e.g., anterior tibialis) and plantarflexors (e.g., gastrocnemius) muscles of the ankle are required to prevent toppling (Tokuno *et al*., 2007; Iqbal, 2011). This subsequently leads to the stiffening of these muscles to prevent excessive movement and enhancement of ankle torque (Kouazki and Shinohara, 2009). Moreover, the plantarflexors are an important agonist muscle in controlling ankle torque. Therefore, the constant regulation of muscle activity during quiet standing to the ankle is vital for controlling postural sway.

2.4.2. Hip Strategy

The hip strategy is another sagittal plane-dominated movement involving the fast displacement of hip movement with active trunk rotation (Nashner and McCollum, 1985; Runge *et al*., 1999), and results in a larger angular acceleration at the hip and an increase in ankle and hip torque (Winter *et al*., 1998; Amin and Harrington, 2013). In contrast to the ankle strategy, the hip strategy has been described as a double-segment inverted pendulum (Runge *et al*., 1999) whereby movement at the ankle and hips occur, projecting the centre of mass to the edge of the base of support. This strategy is often adopted when there is a narrowing of the base of support (e.g., single leg standing) or when there is a reduction in proprioceptive acuity (e.g., due to normal aging) (Warnica *et al*., 2014; Craig *et al*., 2016). It may be possible that the early activation of the hip and trunk musculature could be due to a reduction in the somatosensory system in older adults, which has been associated with peripheral neuropathies and ischemia to the cutaneous surface of the foot (Amiridis *et al*., 2003). Thus, a greater degree of hip flexion is produced resulting in the recruitment (e.g., co-contraction) of the proximal muscles of the thigh when adopting the hip strategy (Winter *et al*., 1996; Nagy *et al*., 2007). Therefore, when there is a compromise at the ankle (e.g., an increase in stiffness or reduction in proprioceptive acuity) the hip strategy may be adopted.

2.4.3. Compensatory Strategy

Muscle co-contraction is the simultaneous recruitment of an agonist and antagonistic across a joint and is an important mechanism adopted by the central nervous system to increase joint stiffness and stability at the expense of an energetic cost (e.g., muscle fatigue) (Donath *et al*., 2016; Iwamoto *et al*., 2017; Li *et al*., 2021). Several studies have indicated that older adults have a larger muscle co-contraction compared to their younger counterparts (Melzer *et al*., 2004; Nagai *et al*., 2011; Donath *et al*., 2016) and is referred to as a compensatory strategy in older adults for the maintenance of postural control (Iwamoto *et al*., 2017). An impairment to the sensorimotor system may lead to an increase in muscle co-contraction (e.g., ankle dorsiflexors/ plantarflexors) by the loss of lower limb muscle strength and/or proprioception (e.g., force and joint position sense) to maintain balance when performing quiet standing balance tasks (Melzer *et al*., 2004). Although, muscle co-contraction is often considered an effective strategy to decrease postural sway, it is now well established that this strategy is more maladaptive than compensatory (e.g., muscle co-contraction) and does not necessarily lead to a reduction in postural sway and may even reduce/ or impede adaptive responses to postural perturbations (Reynolds, 2010; Craig *et al*., 2016).

2.5. Measuring Postural Sway

Posturography is the most widely used quantitative method to assess postural control and sway (Pinsault and Vuillerme, 2009; Golriz *et al*., 2012). A widely used method for evaluating postural sway involves testing an individual's capacity to stay upright and still on a force platform. Excursions of the centre of pressure (COP) express the location of the resultant and ground reaction force acting on the supporting surface (force platform) (Duarte *et al*., 2010). More detail about the force platform and measurements are covered in Chapter 3; Section 3.4.

2.5.1. Static Posturography

Static posturography refers to maintaining balance while standing upright position on a fixed surface (Carpenter *et al*., 2001; Visser *et al*., 2008). This method of assessing postural steadiness is often performed on a force platform using either a fixed or compliant surface (Visser *et al*., 2008). There are several sensory manipulations (eyes open and closed) and different stance positions (double leg, single leg, or tandem stance) which can be performed on the force platform to render postural tasks more difficult (College *et al*., 1994; Carpenter *et al*., 2001; Visser *et al*., 2008; Duarte *et al*., 2010).

2.6. Reliability of Postural Sway Measures

Measuring postural sway can be difficult due to postural sway being a highly “chaotic” system with high variability, with several authors reporting different methodologies that can make the interpretation of the results difficult. The most common method of measuring postural sway is through the analysis of COP using a force platform (Santos *et al*., 2008). However, there are reliability and validity issues with measuring COP using a force platform due to biological variability (Santos *et al*., 2008). Factors such as the duration of the trial (Pinsault and Vullierme, 2009), foot position (Pinsault and Vuillerme, 2009), verbal commands to participants (Zok *et al*., 2008), sampling frequency (Carpenter *et al*., 2001) and the number of trials performed (Santos *et al*., 2008; Wheat *et al*., 2012) can influence COP outcome measures. The measure of COP parameters during trials should be consistent to reflect or eliminate the possibility of systematic and random errors (Carpenter *et al*., 2001). Moreover, it is important to identify if the changes in postural sway are either true changes or biological (e.g., normal variation due to systematic noise in the neuromuscular system) (Duarte *et al*., 2010).

The duration of a trial can influence the reliability and quality of the COP data being collected. Doyle *et al.* (2007) reported acceptable levels of reliability (Intraclass correlation coefficient; ICC ≥0.70) in anterior-posterior, medial-lateral and average velocity COP measures suggesting a minimum of five 60-second trials. However, performing quiet balance tasks for long durations may not be a viable option when performing multiple trials in older adults. In comparison, LeClair, and Riach, (1996) and Pinsault and Vuillerme (2009) recommended that a 30-second trial is an ideal duration for test-retest reliability of COP measures and acceptable to conduct repeated 30-second trials when measuring COP (ICC ≥0.75). Indeed, the most widely used durations seem to be either 30 or 60 seconds which is consistent with the literature (Pinsault and Vuillerme 2009; Duarte *et al*., 2010; Van der Kooij *et al*., 2011). There appears to be no universally accepted duration of time a participant should maintain an upright standing position. Therefore, maintaining an upright posture may lead to the accumulation of muscle fatigue, which can potentially impact an individual's ability to sustain the same position over time. (Pinsault and Vuillerme 2009; Van der Kooij *et al*., 2011).

2.7. Postural Sway Characteristics in Older Adults

Normal aging is accompanied by a change in the function of the musculoskeletal and sensorimotor systems, reflected in the ability to maintain or re-establish postural control (Kanekar and Aruin, 2014). The main characteristics identified in older adults are an increase in neuromuscular noise (Singh *et al*., 2012), a loss in muscle size (Laughton *et al*., 2003), strength (Runge *et al*., 1999; Morcelli *et al*., 2014), motor output (Kouzaki and Shinohara, 2010), reduction of fast twitch muscles fibres (Evans and Lexell, 1995) and loss of proprioception. Consequently, the previously named characteristics are responsible for a faster velocity of COP (Prieto *et al*., 1996), centre of mass acceleration to COP and centre of gravity (Masani *et al*., 2007). In addition, older adults tend to adopt a hip strategy as older individuals often exhibit a decrease in reaction time, a delay in feedback (e.g., latencies in postural muscles) and reliance on co-contraction of agonist and antagonist muscles (Kanekar, Aurin, 2014: Donath *et al*., 2016). Overall, the two key age-related changes observed in older adults are (1) a transition from an ankle to hip strategy and (2) muscle co-contraction. Older adults tend to adopt a hip or stepping strategy when postural sway amplitude and velocity exceed the threshold the ankle can withstand. In addition, mediolateral postural sway amplitude and velocity appear to increase in older individuals because of a decrease in the level of strength and torque produced at the hip (Winter *et al*., 1996; Nam *et al*., 2013; Kanekar and Aruin, 2014). Several prospective studies have suggested that medial-lateral parameters of postural sway are a good predictor of the risk of falls in older adults (Meltzer *et al*., 2004; Hausdorff, 2007; Nam *et al*., 2013). Whereby a combination of hip and compensatory strategies (section 2.4.3) appears present in older adults to prevent excessive postural sway and reduce the risk of falls.

2.8. Functional Balance

There are many ways to assess and measure functional balance in older adults (Alonso *et al.,* 2014). These are popular because they are easy, quick and require minimal equipment or expertise to administer the tests. Although, the reliability, practicality and external validity of functional balance assessments have often been questioned. The most used tests are the Berg Balance Scale (Berg *et al*., 1992), Timed Up and Go (Shumway-Cook *et al*., 2000) and the Sit-To-Stand test (Bohannon, 1995; Goldberg *et al.,* 2012). The functional balance tests employed in this thesis to assess the physical capabilities of younger and older individuals are widely accepted as valid and reliable, as evidenced by existing literature. A more comprehensive understanding of the specific functional tests is reviewed in Chapter 3, section 3.8.

2.9. Age-related Effects

Young adults are better at maintaining postural sway and balance compared to older adults suggesting there is an age-related decline in postural sway with age (Lin *et al.*, 2009). Several studies have reported differences in postural sway during unperturbed quiet standing, walking (Hausdorff, 2007), joint torque (Wojcik *et al*., 2011), force production (McClenaghan *et al*., 1996) and muscle fatigue (reviewed in Chapter 2.10.1; section 2.10.2 and 2.10.3). Lin *et al.* (2009) investigated the effects of unilateral localised muscle fatigue on postural sway immediately after fatiguing the musculature of the ankle, knee, hip, and lumbar spine in young and older adults. The authors identified a greater impairment in older adults than younger adults, as older adults adopted a hip strategy to maintain postural sway. Moreover, recent evidence by Laughton *et al.* (2003) and Nagai *et al*. (2011) have reported a significant increase in muscle activation and co-contraction of lower limb muscles in older adults compared to young adults during quiet standing. Indeed, the increase in postural sway after a bout of muscle fatigue on quiet standing appears to change from anteroposterior to mediolateral with age (Nam *et al*., 2013). It could be considered that with an increase in age, older adults may develop a strategy whereby co-contraction of the musculature around the ankle is adopted as another strategy to prevent falls.

2.10. Postural Sway Responses to Acute Exercise

It is well-established that postural sway increases in response to acute exercise (Lepers *et al*., 1997; Lin *et al*., 2009). Muscle fatigue is an exercise-induced reduction in muscle force or power that temporarily impairs motor performance (Enoka and Stuart, 1992; Gandevia, 2001; Deschenes, 2004). Although the mechanisms are not fully understood (González-Izal *et al*., 2012), the main processes of muscle fatigue associated with a disturbance to postural control are described below and summarised in Table 2.1:

- Metabolic responses (e.g., blood pressure, heart rate or ventilatory) and by-products (hydrogen ions, lactate, calcium ion sensitivity).
- Muscular responses (additional motor unit recruitment, co-contraction of synergistic muscles, stretch reflex, muscle spindle sensitivity).
- Sensory disturbances of visual (moving field of vision and environment) and vestibular information (Otolithic responses e.g., waking, standing = horizontal and vertical induced by head movements/ displacement).
- Proprioception disturbance (joint position and force sense).

Table 2.1: Physiological characteristics and proposed mechanisms responsible for the disturbance of postural sway.

Characteristic	Physiological Disturbance
Metabolic response:	
Blood pressure:	**Blood Pressure:** Standing position results in shift in ~300 - 1000ml blood from central to lower regions. ↓SBP ↔ DBP due to a shift in the sympathetic, parasympathetic nervous system and baroreceptor function. Quiet standing – DLS. ↓SBP = 2.9 - 3.2cm change in postural sway excursion (Murata *et al*., 2012).
Heart Rate:	**Cardiac activity force:** Normal hemodynamic (normal cardiac circadian rhythm) generates (~0.2 - 0.4Nm) of torque at the ankle. Quiet standing - DLS - small deviations in AP and ML postural sway amplitude. (Conforto *et al*., 2001) **Continuous/ steady state exercise:** Intensity below lactate threshold = ↔ disturbance to postural control (Mello *et al*., 2010). **Maximal exercise:** ↑HR (~93% heart rate max) ↑ postural sway (Pendergrass *et al*., 2003).
Ventilatory:	**Hyperventilation:** resulting from ↑CO_2 and ↓pH = ↑ in peripheral nerve excitability. ↑O2 demand. Speed and duration of activity can lead to postural disturbance resulting from an ↑ VO_2 = ↑COP (Mello *et al*., 2010).
Calcium Ion, Hydrogen Ion, Lactate production:	Metabolic by-products cause an ↑ in peripheral muscle fatigue → disturbing postural control. Decrease in postural control related when individuals reach anaerobic threshold. A decrease in muscle pH level raises metabolites. Resulting in an ↑ in muscle spindle afferent activity and myotatic loop activity to regulate postural control (Miller and Bird, 1976; Ruzic *et al*., 2014).
Muscular responses:	
Motor unit recruitment:	An ↑ in EMG activity identified disturbing postural sway and recovery after exercise. Dependent on the limb fatigued (recovery faster with knee compared to ankle fatigue) and muscle histology (slow vs. fast twitch muscle composition) ↑ in EMG latency following fatigue ↑ in postural sway (Dickin and Doan, 2008).

Co-contraction:	Muscle co-contracture of lower limb - leads to an ↑ muscle stiffness to regulate postural sway (Section 2.4.3).
Muscle spindle activity:	Interaction between intrafusal and extrafusal fibres for controlling postural sway through a stretch reflex - leading to an ↑ in reciprocal inhibition under fatiguing conditions. Afferent Ia and II sensory receptors alteration leading to degradation of gamma motor neuron discharge affecting muscle spindle activity (Windhurst, 2007; Robeiro *et al*., 2007a; Ribeiro *et al*., 2013).
Sensorimotor responses:	
Visual:	EO and EC conditions provide disturbances to balance and are dependent on proprioception input and distance of the target to focus on (EO condition). EO ↑ in postural control in high frequencies (>0.5Hz) compared to decrease in lower frequencies with EC↓ (<0.5Hz). Link to greater change in muscle stiffness with EO and EC (Caron, 2004).
Vestibular:	Orientation of the head can increase otolithic receptor sensitivity if a linear acceleration of 6-m/s^{-1} is reached. Under fatigue, the labyrinthine system detects the fluid change in the inner ear through to the cerebellum. Repetitive exercise (causing vertical and horizontal displacement), ↑ postural sway e.g., Forward lean results in arm extension and lower limb extension through stimulation of vestibulospinal reflex in preparation for a fall (Brandt and Strupp, 2005; Paillard, 2012).
Proprioception:	**Force Sense:** ↓Golgi tendon organs activation, ↓their capacity to maintain force output ↑ force match process (Relph and Herrington, 2016). **Joint Position:** Muscle spindle activation disturbs the afferent pathway, resulting in ↑ latency in muscle contracture and alteration of alpha and gamma muscle contracture (Windhorst, 2007; Torres *et al*., 2010).

Key: EO = DLS = Double leg stance, SLS = Single leg stance, Eyes open, EC = Eyes closed, SBP = Systolic blood pressure, DBP = Diastolic blood pressure, HR = Heart rate, VO_2 = Maximal oxygen consumption, COP = Centre of pressure, CO_2 = Carbon dioxide, pH = potential of hydrogen, ↑ = Increase/greater, ↓ = Decrease/lesser, ↔ = No change

2.10.1. Ankle Muscle Fatigue

Several studies have observed an increase in postural sway using different ankle fatiguing protocols by isolating the dorsiflexors and plantarflexors of the ankle due to their importance in postural control (Harkins *et al*., 2005; Bisson *et al*., 2010). The effects of muscle fatigue on the ankle provide information on the 'potential' movement strategies adopted by young and older adults post-exercise (Lin *et al*., 2009). Moreover, the musculature around the ankle is responsible for supporting and stabilising the body during upright standing and walking, as inaccuracies in ankle proprioception can cause an increase in postural sway (Wright and Arnold, 2021). In addition, a decrease in force production and an increase in force sense error when muscle fatigue is induced in the ankle can alter muscle contracture (Carson *et al*., 2002), affecting postural control. Furthermore, an increase in muscle spindle discharge can disrupt efferent feedback and joint position sense (Gribble and Hertel, 2004).

In young adults, it has been extensively reported that after muscle fatigue trials (e.g., isometric or isokinetic exercise) to the dorsiflexors and plantarflexors that there is an increase in anterior-posterior postural sway direction (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Dickin and Doan, 2008; Gimmon *et al*., 2011; Boyas *et al*., 2012). Whereas in older adults there is a disturbance in mediolateral postural sway after ankle fatiguing trials. It appears that the fatiguing protocol (e.g., mode: isokinetic and isometric) and type of contraction (e.g., concentric/ eccentric) applied to the ankle musculature is a key factor in studies observing differences in the direction of postural sway in young and older adults (Christie and Kamen, 2009; Avin and Law, 2011; Paillard, 2012). For example, in older adults, the difference in postural sway direction compared to young adults could be age-related (Connolly *et al*., 1999). Whereby older adults are more resistant to fatigue because of a higher proportion of slow-twitch muscle fibres compared to younger adults, who have an equal distribution of slow and fast-twitch muscle fibres (Evans and Lexell, 1995; Christie and Kamen, 2009). The difference in muscle fibre composition may influence the movement applied. For example, older adults may use an adaptive strategy to offset the fatigue experienced to the ankle dorsiflexors and plantarflexors through the recruitment of the invertors and evertors of the foot to increase stability, which may increase mediolateral postural sway.

Yaggie and McGregor, (2002) reported an increase in mediolateral postural sway while maintaining a single-leg stance position after performing unilateral fatigue to the plantarflexors. A similar response was identified by Berger *et al*. (2005) whereby unilateral fatigue was performed, followed by assessing double leg stance position where mediolateral postural sway was altered despite performing the exercise to the dorsiflexors and plantarflexors (responsible for sagittal plane movement). The author acknowledges that inducing ankle fatigue not only causes anterior-posterior postural sway but can also cause mediolateral postural sway. When selecting tasks for assessing quiet standing and gait stability after a fatiguing protocol, it is crucial to consider the variability and conflicting reports found in the literature. These discrepancies may be attributed to the methodologies employed to induce ankle fatigue and should be carefully examined.

2.10.2. Knee Muscle Fatigue

Evidence on the use of knee fatiguing protocols has primarily reported a change in mediolateral postural sway compared to anterior-posterior for ankle fatiguing protocols (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Nandi *et al*., 2008: Bisson *et al*., 2010). Studies performing isokinetic fatiguing protocols to the knee extensor and flexor muscles followed by completing the postural sway task post-exercise have reported changes in strategies to maintain postural control. Gribble and Hertel, (2004) described an increase in mediolateral direction after isokinetically fatiguing the knee and hip musculature on single-leg balance with eyes open. Similar findings were observed by Bizid *et al*. (2009) using a similar fatiguing protocol to Gribble and Hertel, (2004), whereby anterior-posterior and mediolateral postural sway increased more when the knee was fatigued as opposed to the ankle fatiguing protocols. Moreover, Chaubet and Paillard, (2012) identified an increase in mediolateral postural sway after using 20% of an individual’s maximal voluntary contraction (MVC) to elicit muscle fatigue to the knee extensors. It is apparent that after an acute bout of fatigue applied to the extensor and flexor muscles of the knee, mediolateral postural sway is affected more often than anterior-posterior direction. The mediolateral postural sway is adopted more often in older adults than young adults and does not seem to vary between double-leg or single-leg stance positions.

2.10.3. Time Course Changes to Fatigue

Assessing postural sway immediately following a maximal or submaximal fatiguing protocol is a common method for observing the acute effects of muscle fatigue. The duration of these protocols is determined by various factors, including the type of muscular contraction, exercise duration, and loss of maximum voluntary contraction as a result of muscular fatigue. A review by Paillard, (2012) identifies a relationship between the percentage of MVC loss and disturbance in mean COP velocity. The loss of MVC after a short but intense bout of exercise impacts the COP, resulting in the disturbance of postural control post-exercise. Harkins *et al.* (2005) reported an increase in postural sway when using two ankle fatiguing models (30% and 50% MVC) to the plantar and dorsiflexors of the ankle. Postural sway was impaired for 75 seconds for 30% MVC and 35 seconds for 50% MVC before returning to baseline measures. Similar findings have been observed by Yaggie and McGregor, (2002), applied 50% MVC to the invertors, evertors, plantar and dorsiflexors of the ankle. A return to baseline postural sway measures after the localised fatiguing trials took 20 minutes. Other examples include Lin *et al.* (2009), who indicated young and older adults took 11 and 2 minutes to return to the pre-fatigue state for anteroposterior velocity respectively, and Boyas *et al.* (2011) whereby sway area and anteroposterior velocity, recovered within 2 minutes after dorsiflexor and plantarflexor fatigue on unilateral balance.

2.11. Advances in Resistance Training

Resistance training is a traditional method of exercise commonly performed by sport-specific athletes to improve hypertrophy, strength, or power (Kraemer and Ratamess, 2004; Suchomel *et al*., 2018). The anabolic effect resistance training has on skeletal muscle is well documented (Schoenfeld *et al*., 2017: Grgic *et al*., 2018). Marked increases in protein synthesis (MacDougall *et al.,* 1995), myofibrillar breakdown, and muscle fibre cross-sectional area (Schoenfeld, 2013; Ogborn and Shoenfeld, 2014), insulin growth like factor response (Loenneke, 2012b) and neuromuscular function are well recognised adaptations from resistance training. The American Society of Sport and Exercise Medicine recommend >70% of an individual’s 1-repetition maximum (1RM) as an exercise intensity to increase strength or hypertrophy (Wernbom *et al*., 2007). However, it is widely accepted that the physical demand for an older adult to perform resistance training at 60-80% of an individual's 1RM is not a realistic intensity to sustain (Siguin and Nelson, 2003). Novel training techniques and innovative methods are required to ensure resistance exercise is sustainable in older adults. Blood flow restrictions (BFR) is a relatively new training method appropriate for older adults to use as it allows the individual to train at lower training intensity when performing resistance exercise (20-50% 1RM) (Abe *et al*., 2005). This novel training method isolates a targeted muscle group in the upper or lower limb by applying an inflated pneumatic cuff, tourniquet or KAATSU system to perform resistance exercise. The acute and chronic physiological responses and adaptation to LIBFR in young (Abe *et al*., 2005; Manini *et al*., 2012) and older adults (Fry *et al*., 2010; Yasuda *et al*., 2016) are like those observed in normal strength and hypertrophy exercise (Loenneke and Pujol, 2009). The application of this training method will be discussed in the following section detailing the rationale, acute responses, and chronic adaptations to exercise in young and older adults.

2.12. Rationale for Low Intensity Resistance Exercise with Blood Flow Restriction

Low-intensity resistance exercise with BFR (LIBFR) is a training method consisting of the partial reduction of venous blood pressure to the upper arm or upper thigh without compromising arterial blood pressure (Kawada, 2005; Loenneke and Pujol, 2009). The application of this method of training is by use of either a pneumatic blood pressure cuff (Takarada *et al*., 2000a), specially made tourniquets (Takarada *et al*., 2000a), or KAATSU system (Takarada *et al*., 2000, Abe *et al*., 2006) inflated to a percentage or target resting brachial artery systolic blood pressure (SBP) (Loenneke *et al*., 2010). Previous literature has indicated that LIBFR could be a beneficial mode of resistance exercise for improving activities of daily living in older adults, which is due to the lower intensity (20-50% 1RM) required for this modality (Abe *et al.* 2006; Loenneke *et al*., 2010; Park *et al*., 2015; Ozaki *et al*., 2016). Furthermore, it is considered that training at 20% 1RM is equivalent to 10-30% of maximal work capacity (Abe *et al.,* 2006; Loenneke *et al*., 2012c). This would imply that training at a lower intensity would provide a more tolerable dosage for older adults to perform resistance exercise. In addition, the benefits of LIBFR have been seen in older adults including participants with osteoarthritis (Buford *et al*., 2015), cardiovascular disease (Down *et al*., 2014), hypotension (Pinto and Polito, 2015); during concurrent training (libardi *et al.,* 2015) and following injury (Ohta *et al*., 2003). The physiological responses and adaptations to this type of training reported from the literature are anecdotal. However, based on the literature available, the use of LIBFR has demonstrated an increase in muscle hypertrophy and strength in young (Manini *et al*., 2012; Kim *et al*., 2012) and older (Fry *et al*., 2010; Karabulut *et al.,* 2010; Yasuda *et al*., 2014; Bigdeli *et al*., 2020) populations. Yet, it remains unclear whether the strength benefits achieved from LIBFR can contribute to reducing the risk of falls and improving activities of daily living among older adults.

2.13. Physiological Responses to Low Intensity Resistance Exercise with Blood Flow Restriction exercise

The proposed physiological responses following LIBFR involve complex signalling and stimulation of systemic and local pathways (Loenneke *et al*., 2012b; Wernbom and Aagaard, 2019). For example, Abe *et al*. (2005) reported a 24% increase in systemic insulin growth factor-1 response after two weeks of LIBFR compared to a group completing low-intensity resistance exercise. The results from Abe *et al.* (2005) are comparable to the findings reported during high-intensity resistance training (Goto *et al*., 2005). This is partly due to the combination of lower mechanical load and higher metabolic stress, which appears to be the catalyst during this type of training. However, a deeper understanding of the exact physiological responses to LIBFR remains unclear or unanswered.

More recent accounts of LIBFR refer to a mechanistic approach to help understand the physiological responses when performing this type of training but are unfortunately not well defined. Pearson and Hussain (2015) describe the mechanisms involved in LIBFR as a cascade of effects and have been termed primary mechanisms followed by the activation of secondary mechanisms. The two primary mechanisms are metabolic stress and mechanical tension, which work synergistically and contribute to the secondary mechanisms involved in muscle induced by hypertrophy with LIBFR. The secondary mechanisms are classed as a signal response providing activation of systemic/local endocrine and neuromuscular pathways to increase autocrine/paracrine actions to promote muscle hypertrophy (Kawada, 2005; Karabulut *et al*., 2007; Abe *et al*., 2012; Pearson and Hussain, 2015). A more detailed account of the acute responses and adaptations to LIBFR will be discussed in the following section with an emphasis on the mechanisms responsible during this mode of exercise.

2.14. Acute Physiological Responses to Low Intensity Resistance exercise with Blood Flow Restriction

The acute physiological responses to LIBFR in young adults similar to the responses identified in older adults (Baker *et al*., 2020). For example, in an acute bout of LIBFR training there is a significant increase in growth hormone release post exercise in young and older adults (Sato *et al*., 2005; Manini *et al*., 2012; Seo *et al*., 2016). More specific responses to acute BFR exercise will be discussed in the following subsections.

2.14.1. Growth Hormone

An increase in the growth hormone (GH) response during resistance exercise is essential for skeletal muscle growth (Abe *et al*., 2012). During an acute bout of LIBFR resistance exercise (Abe *et al*., 2005; Fujita *et al*., 2008, Leite *et al*., 2015), GH release appears to be elevated in young and older adults (Fry *et al*., 2010). Takarada *et al*. (2000a) demonstrated an increase in GH 15 minutes after the cessation of LIBFR exercise (20% 1RM) in young adults. Similarly, Manini *et al.* (2012) reported an increase in post-exercise (45 minutes) growth hormone release in young and older adults performing LIBFR (20% 1RM) compared to high-intensity resistance knee extension exercise (80% 1RM). However, the results from Manini *et al.* (2012) identified a significant increase in post-exercise growth hormone release in young compared to older adults. Many other authors have reported a similar increase in growth hormone release in young (Sato and Abe, 2005; Leitie *et al.,* 2015) and older adults (Seo *et al*., 2016) after the cessation of BFR exercise.

2.14.2. Lactate Accumulation

There appears to be consistency within the literature suggesting lactate accumulation may increase the response of growth hormone release in young and older adults. It has been theorised that an increase in systemic blood lactate accumulation could affect the stimulation of central and peripheral nervous system response after BFR exercise (Eiken *et al*., 1991; Viru, 1998; Takarada *et al*., 2000b). Ishii and Nishada, (2013) reported that there is an increase in the stimulation of the somatosensory cortex after upper limb BFR, which is linked to higher levels of blood lactate and stimulation of the metaboreflex. Furthermore, the study reported that exercising at 50% MVC increased group III and IV muscle afferents. Therefore, when a muscle becomes fatigued through repeated muscle contractions following LIBFR, the reduction in blood flow to the exercising muscles causes localised ischemia, which may increase metabolites such as potassium, lactic acid and bradykinins (Pollak *et al*., 2014). These metabolites activate nociceptive stimuli causing pain, while group III and IV are mechanically sensitive and increase blood pressure, heart and ventilatory rate. Older adults predominately have a lower threshold of fast twitch muscle fibres than young adults (Evans and Lexell, 1995). Therefore, creating a hypoxic environment in the lower limb in older adults may encourage the recruitment of fast twitch muscle fibres by these metabolites becoming more readily available to enhance the recruitment of group III and IV muscles afferents. Moreover, an increase in motor units in the cutaneous surface of the foot may improve the ability of older adults to improve their ability to maintain postural control and reduce the risk of falls.

2.14.3. Insulin like Growth Factor-1

A localised hormone responsive to mechanical tension during LIBFR is an increase in insulin growth factor-1 (IGF-1). Numerous studies have observed increases in IGF-1 immediately after the cessation of BFR exercise in young (Abe *et al*., 2005), middle age (Seo *et al*., 2016) and older adults (Fry *et al*., 2010). Abe *et al*. (2005) reported an increase of 23.8% in serum IGF-1 after two weeks of LIBFR training when performed twice a day in young adults. Similar findings have been observed in young (Takarada *et al*., 2000a), middle-aged (Seo *et al*., 2016) and older adults (Fry *et al*., 2010) when performing LIBFR. The activation of IGF-1 is crucial for muscle protein synthesis. Mechanical tension created by resistance exercise appears to stimulate the isoform IGF-1c (mechano-growth factor [MGF]) (Abe *et al*., 2005). This isoform appears to initiate post-exercise muscle hypertrophy response after resistance exercise. Previous studies have proposed that IGF-1c would encourage the activation of mitogen-activated protein kinase (MAPK) and mechanistic target of rapamycin (mTOR) signalling pathways, encouraging muscle hypertrophy and preserving muscle. However, the extent to this theorised response to BFR training is scarce.

2.14.4. Mechanistic Target of Rapamycin and Mitogen Activated Protein Kinase-1

Stimulation of key anabolic pathways such as mechanistic target of rapamycin (mTOR; Fry *et al.,* 2010), MAPK (Pearson and Hussain, 2015), ribosomal S6 kinase 1 (Loenneke and Pujol, 2009) are elevated after an acute bout of BFR exercise. This in turn initiates protein synthesis, proliferation of satellite cells (Morgan and Partridge, 2003) and mRNA transcription which are important for muscle hypertrophy in traditional resistance exercise (Manini *et al*., 2012). Fry *et al.* (2010) reported an elevation in mammalian target of rapamycin and ribosomal S6 kinase 1 up to 3 hours' post LIBFR exercise (20% 1RM) in older adults. In contrast, Ozaki *et al.* (2016) conducted a walking protocol combined with blood flow restricted exercise in young adults and observed an increase (3-hour post) in ribosomal S6 kinase 1 but no difference in mammalian target of rapamycin. In this context, it is possible to assume aerobic exercise is not a sufficient stimulator in anabolic signalling compared to LIBFR.

2.15. Chronic Adaptations to Low Intensity Blood Flow Restriction Exercise in Young and Older Adults.

General strength and muscle hypertrophy adaptations to LIBFR is widely reported in young adults, with emerging evidence in older adults (Baker *et al*., 2020; Lim and Goh, 2022). It is apparent that acute responses and adaptations to resistance exercise with LIBFR increases strength and muscle hypertrophy, which may be due to neurological and morphological adaptations. Indeed, over a prolonged period (2-16 weeks), increases in strength and muscle hypertrophy has been observed in young (Fatela *et al*., 2008; Barcelos *et al*., 2015; Lixandrão *et al*., 2018) and older adults (Karabulut *et al*., 2010; Yasuda *et al.,* 2016; Cook *et al*., 2017; Baker *et al*., 2020). Therefore, increasing strength in fundamental movements that reflects everyday activities in older adults is probable. However, it remains unclear as to whether the neuromuscular adaptations that occur from LIBFR improve postural sway and balance in older adults.

2.15.1. Strength Adaptations

Previous studies examining the physiological adaptations of LIBFR have consistently shown improvements in strength following isometric, isotonic, and isokinetic resistance exercise (Takarada *et al*., 2000b; Abe *et al*., 2005; Fatela *et al*., 2008; Barcelos *et al*., 2015; Seo *et al*., 2016; Lixandrão *et al*., 2018). For example, Manimmanakorn *et al*. (2013) observed an increase in maximum voluntary isometric contraction (MVIC) strength for 3 seconds (s) (13.3%) and a sustained MVIC (30s) for muscular endurance (9.3%) after 5-weeks of bilateral LIBFR (20% 1RM) in young adults. Kim *et al.* (2009) reported similar findings to Manimmanakorn *et al*. (2013) in young adults, with LIBFR at 20% 1RM after 3-weeks of bilateral isotonic knee extension (2.2%), knee flexion (3.7%) and leg press (1.6%) exercise. The adaptations to strength can occur over a relatively short or long period with LIBFR in young adults. In addition, several meta-analyses have consistently reported LIBFR as a potent stimulus for the development of strength in young adults (Loenneke *et al.,* 2012a; Lixandrão *et al*., 2018) and are also evident in older adults (Centner *et al*., 2019). Karabulut *et al*. (2010) reported an increase in leg extension (19.1%) and leg press exercise (19.3%) after 6-weeks of LIBFR (20% 1RM) in older adults performed. Similarly. Yasuda *et al*. (2014) reported an increase in knee extensor (26.1%) and leg press (33.4%) after 12 weeks of LIBFR (20-30% 1RM). It appears that the adaptations in strength, independent of age (young vs older adults), can increase lower limb strength, and this is evident in a comparative study when using the same LIBFR protocols in young and older adults (Manini *et al.,* 2012). Moreover, increasing strength may benefit older adults by potentially reversing the effects of sarcopenia. In turn, the strength adaptations observed in young and older adults from LIBFR may improve neuromuscular function, which is essential for older individuals (Karabulut *et al*., 2007; Cook *et al*., 2017). Based on the evidence available on the strength adaptations from LIBFR, this modality might improve physical function (e.g., rising from a chair) among older people.

2.15.2. Neurological Adaptations

The function of the neuromuscular system and adaptations following LIBFR at relatively low mechanical loads (20-50% 1RM) are similar to traditional strength training (Shinohara *et al*., 1997; Abe *et al*., 2010; Karabulut *et al*., 2010; Karabulut *et al*., 2013; Manimmanakorn *et al*., 2013). An early investigation by Shinohara *et al*. (1997) reported an increase in strength and maximal rate of torque development after 4-weeks of unilateral LIBFR performing isometric training at 40% maximal voluntary contraction. The authors suggested that there is a greater capacity to recruit higher threshold motor unit activation as opposed to increasing the size of the muscle following LIBFR. This view on the enhancement in motor unit recruitment following LIBFR exercise is consistent within the literature and referred to as an increase in force production that may be related to an accelerated shift from slow to fast twitch muscle fibre recruitment (Kubo *et al*., 2006; Loenneke *et al*., 2012c; Karabulut and Perez, 2013, Wernbom and Aagaard, 2019; Kjeldsen *et al*., 2019). This suggests that earlier recruitment in motor unit activation can occur following LIBFR. Furthermore, a comprehensive meta-analysis conducted by Cerqueira *et al*. (2022) on the myoelectric activity following LIBFR over a training period of ~6-12 weeks reported long-term improvements in muscle activation. However, the studies (n = 6) included for the analysis of long-term adaptation were in young adults, performing exercise to failure, where performing to failure is not always optional for all individuals (e.g., older adults). In addition, a scarce amount of information on the neurological adaptations following LIBFR interventions on myoelectric activity in older adults, warrants further attention. In summary, the current body of evidence suggests an accelerated transfer from slow to fast twitch muscle fibre recruitment may occur following LIBFR. The utilization of LIBFR exercise in older adults who present with a higher proportion of slow twitch muscle fibres (Evans and Lexell, 1995) should be given careful consideration due to its potential benefits. This particular modality has demonstrated the ability to significantly enhance the response of fast twitch muscle fibres, thereby resulting in increased strength and torque production.

2.16. Potential Improvements in Postural Control with Low Intensity Resistance Exercise with Blood Flow Restriction Training

The practical application of LIBFR to increase lower limb muscular strength in young and older adults has been well documented in the literature. The increase in strength following LIBFR has also been shown to increase anabolic markers and signalling processes commonly associated with HI resistance exercise. However, despite the vast amount of evidence on the strength and hypertrophy benefits following LIBFR, there appears to be a paucity of research on balance.

Currently, there is a scarce amount of evidence on whether the strength benefits when performing regular bouts of LIBFR would transfer to an improvement in either quiet standing or gait stability. An early account by Diener *et al*. (1984) on postural stability following acute ischemia applied above the ankle resulted in destabilisation and a loss in proprioception input to the cutaneous surface of the foot. This early account used high SBP pressures to anesthetise the foot to assess proprioception and postural stability. In addition, the placement of the tourniquet was above the ankle and is not an ideal location for the current guidelines on LIBFR. Despite the issues raised in the study by Diener *et al*. (1984), applying ischemia to the lower limb led to a disturbance in postural control. There is a possibility that using LIBFR in a controlled setting with an appropriate BFR pressure could be of benefit to older adults. Moreover, there is a high prevalence of falls in older adults related to the steady decline in muscular strength and slower reaction time due to a greater proportion of slow to fast twitch muscle fibres. Therefore, LIBFR may be a modality which could allow older adults to perform low-intensity resistance exercise but gain the benefits of HI resistance exercise. This may be an opportunity to improve the recruitment of fast twitch muscle fibres responsible for fast reactive responses, which could help reduce postural sway in young and older adults.

2.17. Summary

Based on the literature reviewed within this Chapter there are several questions which need solving to gain a further understanding on the acute and chronic effects of LIBFR exercise in young and older adults. The strength and potential neurological adaptations from LIBFR exercise could be a good alternative to traditional HI exercise for older adults. Furthermore, using a LI exercise in this population may help improve exercise tolerance and adherence to resistance exercise programmes. One of the most challenging aspects when performing resistance exercise is the high intensity required to achieve strength adaptations, which is not always possible in older adults. It is in our interest to identify whether using LIBFR can improve postural control in this population.

Chapter 3: General Methods

3.1. Introduction

This Chapter will provide a detailed description of the equipment, and procedures used within the three experimental studies (Chapter 4, 5 and 6) including the current methods used to perform blood flow restriction (BFR), blood pressure, strength, and postural sway. Furthermore, the assessment and procedures used when performing quiet standing and dynamic balance tests when assess postural sway to measure centre of pressure (COP) on a force platform will be explained.

3.2. Data Protection

All participant information collected throughout the course of the thesis was anonymised and kept confidential. Personal information was stored away in a locked filing cabinet and only accessible to the lead investigator. Data collected and stored electronically on the University of Northampton computers was coded, password protected and transferred onto a personal computer/storage device and deleted immediately from University of Northampton computers. All procedures carried out were in accordance with the Data Protection Act. 1988.

The experimental procedures conducted within this thesis were reviewed and ethically approved by the University of Northampton Research Ethics Committee (Appendix A1) and conformed to the Declaration of Helsinki (1964) guidelines. To the authors' knowledge, there was no confliction of interest or personal relationship with anyone who could influence the outcome of the ethics committee.

3.3. Participants

Healthy young (age; 18-30 years) and old (age; 60-79 years) male and female adults were recruited for all investigations (Chapter 4; section 4.2.1, Chapter 5; section 5.2.1 and Chapter 6; section 6.2.1). Young healthy participants were recruited to understand how this population responded to BFR exercise and the effect it may have on quiet standing and gait stability prior to exposing older adults to the same protocols. All participants were recruited through poster advertisement, word of mouth and local community clubs within Northamptonshire. All participants were made aware of the procedures and measures and had the opportunity to ask questions and could voluntary withdrawal at any time. Prior to the start of each study, the participants were provided with a participant information sheet (Appendix A2) outlining the purpose of the study. All participants completed a physical activity readiness form (PAR-Q; Appendix A3) containing health, medical and exercise related questions to determine the participant's eligibility to perform the protocol stated in the participant information sheet. This was followed by the signing of an informed consent form (Appendix A4).

3.3.1. Inclusion Criteria

All participants were physically active (minimum of 150 minutes of moderate aerobic activity per week (e.g., walking to the shops; Sparling *et al.,* 2015; Bull *et al.,* 2020) or had physically active jobs (Haskell *et al*., 2007) and was determined from the PAR-Q. None of the participants recruited had participated in resistance exercise <6 months prior to the study (Yasuda *et al*., 2016). In addition, all participants were free from any musculoskeletal disease, neurological condition, or functional impairments.

3.3.2. Exclusion Criteria

Young and older adults recruited for each study were asked to continue with their normal daily diets and consumed less than the recommended weekly units of alcohol (14 units per week), and a non-smoker (<6 months) (Wood *et al*., 2018; Firth *et al*., 2020). All participants were reminded to avoid drinking alcohol, caffeine, and sports drinks and exercise 24 hours before testing (Yasuda *et al*., 2016). Each experimental trial was separated by 24-48 hours of recovery and performed at the same time of day to avoid any influences of circadian rhythm (Van Dongen and Dinges, 2000). As each study consisted of BFR during exercise, all participants with a high resting systolic (SBP) and diastolic (DBP) blood pressure (140mmHg ≥ SBP / 90mmHg ≥ DBP) were excluded from the study (Chobanian *et al*., 2003; Alyabsi *et al*., 2020). It was important to consider this for older adults as SBP is a more potent cardiovascular risk factor in older adults over 50 years of age (Chobanian *et al*., 2003). In addition, older adults are more at risk of disease and medical conditions, and an additional exclusion criterion used to define the individual as 'healthy' was used based on Greig *et al*. (1994) criteria. Healthy participants were free from disease and did not report any recent balance, uncorrected vision, vestibular disorders, diabetes, musculoskeletal injury (e.g., ankle or knee-related injuries) or neuropathic conditions (e.g., altered sensation or radiating pain).

3.4. Force Platform

3.4.1. Postural Sway Measurements

A single square shaped (50.2 [L] x 50.2cm [W] x 4.5cm [D]) portable Hall effect force platform (AMTI, Accusway, Watertown, MA) was used to measure posturographic ground reaction forces. Three orthogonal force components (Fx, Fy and Fz) and three moment components (Mx, My and Mz) is simultaneously recorded. All ground reaction forces were sampled at 100Hz and filtered using a 4th order analogue low pass (6Hz) Butterworth filter (AMTI, BioAnalysis, Version 2.2, Watertown, MA). The force platform was operated using a single power supply (220V) connected to a 6-digital channelled interface box. Once pressure is exerted to the surface of the force platform, the magnetic element (Hall element) transfers the vertical and horizontal forces applied to the surface producing an electrical signal amplified by the Hall effect sensor. The Hall effect sensors in all four corners of the force platform respond once pressure is exerted on the surface through a built-in amplifier located within the force platform. A digital external synchronised signal transmits a unique data set from the internal processor located within the platform to the laptop for real time data and analysis. Real time data were recorded (AMTI, Netforce, Watertown, MA) and transformed into centre of pressure data using the accompanying software (AMTI, BioAnalysis, Version 2.2, Watertown, MA). To quantify postural control, the amplitude of velocity of the COP was calculated. The amplitude of the COP displacement is the distance between the maximum and minimum COP coordinate, which was calculated in the anteroposterior (COP_{AP}) and mediolateral (COP_{ML}) direction (cm) to quantify postural performance (Paillard and Noe, 2005). Lower amplitude COP displacements typically reflects better postural control, with larger amplitude displacements indicative of poorer postural control (i.e., COP moving closer to limits of the base of support). In addition, mean COP velocity (COP_{VL}) (cm/s) was calculated (e.g., total distance travelled – calculated by dividing the COP excursion by the duration of the trial (Paillard and Noe, 2005), which is thought to reflect the rate at which postural adjustments are made with COP lower velocity indicative of a smoother and more controlled adjustments. On the other hand, faster adjustments indicate more frequent/ rapid corrections to correct postural errors. These three COP parameters are commonly used to assess postural control (Zok *et al*., 2008) with previous studies reporting good reliability and validity in both young (Pinsualt and Vullereme, 2009; Golriz *et al*., 2012) and older adults (Moghadam *et al*., 2011).

3.4.2. Postural Sway

All participants were asked to maintain an upright double leg stance (DLS), single leg stance (SLS; left or right leg) or tandem (TAN; left or right leg) stance position conducted with either eyes open or closed on a fixed or foam surface (Figure 3.1). All stance positions were randomly assigned to each participant, who performed each stance in silence (quiet standing) while barefooted on a portable force platform. A medium-density foam balance pad (Balance-pad Plus, Alcan Airex AG, Switzerland: apparent 55kg/m^3) was used for the compliant surface tasks and removed for firm surface tasks. To ensure continuity between trials, the foot position was standardised using a customised 'V' foot template outlined on the force platform ([feet abducted at 30° as measured at the great toe with heels separated by a distance of 3cm between the medial malleolus of the tibia] Figure 3.2; Duarte *et al*., 2010). The participants were instructed to remain as still as possible throughout each trial with their arms left to hang freely beside their body. All participants were asked to fixate their vision onto a wall-mounted black cross 2.00m (distance cross dimensions 25cm x25cm). The location of the cross was adjusted to the participant's eye line to avoid vestibular disturbance (Vuillerme and Nafati, 2007). Each trial was completed 3-times and interspersed with a 30-second break. Evidence suggests that 3 x 30s trials have a high level of reliability for COP measures (Le Clair and Riach, 1996; Pinsault and Vuillerme, 2009).

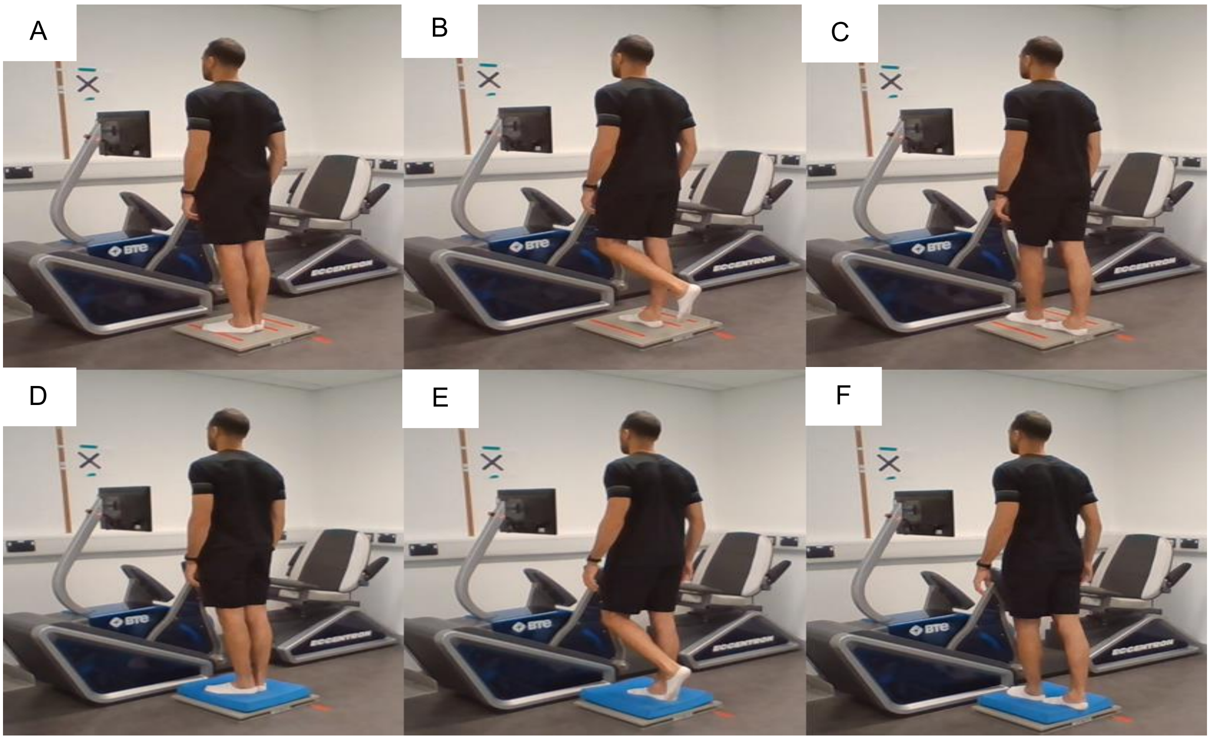

Figure 3.1: Stance position adopted for the assessment of postural sway amplitude and velocity. (A) Double leg stance - Firm, (B) Single leg stance - Firm, (C) Tandem stance (left) - Firm, (D) Double leg stance – Foam, (E) Single leg stance – Foam, (F) Tandem stance (left) – Foam).

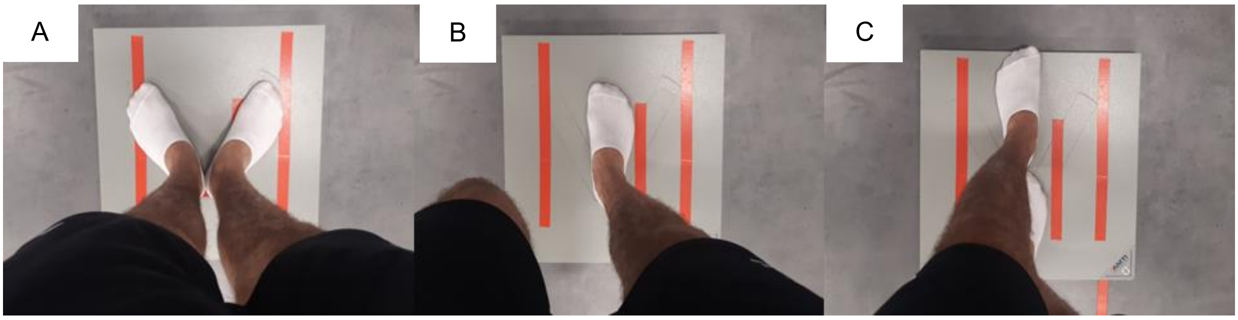

Figure 3.2: Transverse view of stance position adopted for the assessment of postural sway amplitude and velocity. (A) Double leg stance, (B) Single leg stance, (C) Tandem stance (left) on a firm surface.

3.5. Pilot Study 1: Intra and Inter-Session Reliability of centre of pressure measures to assess postural control during quiet stance on a fixed compliant surface

3.5.1. Introduction

Postural control during unperturbed quiet stance is commonly assessed with COP measures recorded using a force platform. At this point it is important to note that the reliability represents an essential prerequisite for any outcome measure to ensure that observed difference in COP measures (e.g., due to acute and training induced stresses) reflect a true change in postural control rather than systematic or random error (Hopkins *et al.,* 2009; Shrout and Fleiss, 1979). Force platform measurements can be subject to measurement errors that include two types of variability: intra-session (within a single session), inter-session retest (between sessions). Although several studies have investigated the reliability of COP measures, it is difficult to make specific and direct comparisons across these studies because of differences in the experimental setup and protocol (e.g., sampling time, sampling frequency, instructions issued to subjects, number of trials recorded). Therefore, the main objectives of this pilot study were to (1) determine intra-and-inter session reliability of COP based on measures of postural stability of a normal young and older adult population and (2) establish a standardised and repeatable protocol to assess quiet standing balance.

3.5.2. Methods

3.5.2.1. Participants

Eight young (age: 27.5 ± 1.5 years) and five older (age: 67.8 ± 6.9 years) men (8) and women (5) were recruited through a convenience sampling method and were required to attend the laboratory on three occasions, separated by 24 hours. All participants read a participant information sheet and completed a PAR-Q and signed an informed consent form. Participants were excluded from the study if they did not meet the exclusion criteria describe in Chapter 3; section 3.3.1 and 3.3.2.

3.5.2.2. Procedure

Balance performance was recorded over three separate days (intersession), each separated by 24 hours. During each visit, postural sway was assessed a total of three times (intrasession) for each postural condition. To examine postural sway amplitude and velocity, each participant performed balance tasks while standing quietly on a force platform following the instruction described in Chapter 3.4.1 and 3.4.2. Three standing balance tasks were chosen based on varying level of difficulty and for comparative purposes with previous literature (Vuillerme and Nafai, 2007; Pinsualt and Vuillerme, 2009) These tasks were DLS: eyes open (EO) and close (EC) [firm surface]; DLS: EO and EC [foam surface]; SLS left leg (SLS_L) and SLS right leg (SLS_R): EO and EC [firm surface]; SLS_L and SLS_R: EO and EC [foam surface]; Tandem left leg (TAN_L) and Tandem right leg (TAN_R): EO and EC [firm surface]; TAN_L and TAN_R stance: EO and EC [foam surface] (Figure 3.3). A total of three trials were recorded for each condition and a mean of these trials were used for analysis from the data collection session. Throughout all tests, the investigator stayed close to the participants to prevent falling without interring with the balance performance. Intrasession reliability was calculated using data from the final visit.

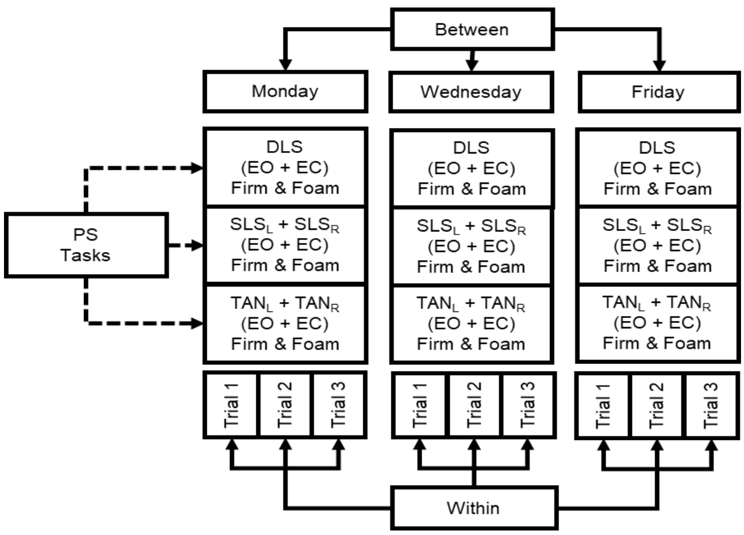

Figure 3.3: Intrasession and intersession reliability schedule for measuring postural sway amplitude and velocity in young and older adults. Key: DLS = Double leg stance, SLS_L, Single leg stance – left leg,

SLS_R = Single leg stance – right leg, TAN_L = Tandem – left leg, TAN_R = Tandem – right leg, EO = Eyes open, EC = Eyes closed.

3.5.3. Statistical Analysis

Data was analysed using a statistical package for the social sciences (SPSS version 22. Chicago, IL, USA). A Shapiro-Wilk test was used to assess normal distribution of the data set and a Mauchly's test of sphericity to verify homogeneity of variance. The reliability of COP measures was determined using intra-class correlation coefficient (ICC) and typical error expressed as the coefficient of variation (CV). According to Hopkins *et al*. (2009) an ICC value of 0.8 – 1.0 should be used as a threshold indicating excellent reliability. In addition, the following thresholds were used for anything below 0.9, <0.40 = ''poor'', 0.40–0.75 = ''fair to good'', >0.75 = ''excellent'' reliability (Carpenter, 2001, Ruhe, Fejer and Walker, 2010). The ICC has previously been used to report relative reliability of COP (Carpenter, 2001; Pinsault and Vuillerme, 2009; Ruhe *et al*., 2010). The coefficient variance was collected from each individual and calculated using the standard deviation and the mean of three COP measures (CV = [(SD/mean) x 100]. Mean ± Standard deviation (SD) for the within and between session variation are also reported. A 95% confidence interval include upper and lower limits was calculated using a specially formatted Excel spreadsheet. A One-way ANOVA was performed on between-session variation only. For the between-session variation, mean ± SD are reported as an average of three trials. Within session variation is reported as individual trials.

3.5.4. Results

Within session reliability for COP_{AP}, COP_{ML} and COP_{VL} were consistent amongst young and older adults but varied from near perfect scores (>0.81) to fair agreement (0.40 – 0.75). No difference in between session reliability was observed on COP_{AP}, COP_{ML} or COP_{VL} in either the young or older adults across all conditions (DLS, SLS_L and SLS_R and TAN_L and TAN_R on either firm or foam surfaces with EO and EC) (P = <0.05) over 3 separate days of testing (Appendix A6). The full intrasession and intersession reliability data set for all stance positions (young and older adults) can be viewed in Appendix A6.

The experimental studies conducted throughout this thesis adopted 3 stance positions (DLS: EC [Firm surface], SLS_L and SLS_R with EO [Firm surface]) to assess quiet standing in young and older adults. To reduce the number of tables for all stance positions assessed in this pilot study, the results provided below are based on the 3 stances position previously mentioned (Table 3.1, 3.2 and 3.3). The supplementary data can be viewed for all stance position can be viewed in Appendix A6.

Within session reliability for COP_{AP}, COP_{ML} and COP_{VL} (Figure, 3.4, 3.5; Table 3.1, 3.2 and 3.3) were consistent amongst young and older adults but varied from near perfect to poor scores. Young and older adults reported “excellent” scores for BL on COP_{AP} and COP_{VL} but not for COP_{ML}. Older adults reported poor results for both SLS_L and SLS_R stances position for COP_{AP}, COP_{ML} and COP_{VL} compared to young adults who were between “fair to good” and “Excellent”. No difference in between session reliability was observed on COP_{AP}, COP_{ML} or COP_{VL} in either the young or older adults across (P = <0.05) over 3 separate days of testing (Table 3.1, 3.2 and 3.3).

Table 3.1: Intrasession and intersession COP_{AP} reliability data set for young and older adults

Task	Group	Day	CV (%)	ICC Between	ICC Within	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
DLS EC Firm	YA	1	59.4	0.59	0.62	Fair to Good	2.0	2.30	6.32	
		2	60.1		0.48	Fair to Good	1.8	2.03	5.68	$F_{(2,18)}$ = 0.693, P = 0.513
		3	49.3		0.79	Excellent	2.1	3.34	7.56	
	OA	1	41.0	0.53	0.15	Poor	1.6	2.90	6.20	
		2	30.5		0.93	Excellent	1.0	2.00	4.10	$F_{(2,15)}$ = 1.184, P = 0.333
		3	49.0		0.80	Excellent	1.5	1.70	4.60	
SLS_L EO Firm	YA	1	29.1	0.43	0.31	Poor	1.2	4.00	6.37	
		2	50.3		0.46	Fair to Good	1.0	2.77	6.40	$F_{(2,18)}$ = 1.766, P = 0.199
		3	33.0		0.50	Fair to Good	1.5	4.15	7.06	
	OA	1	57.9	0.58	0.55	Fair to Good	3.3	2.70	9.30	
		2	49.2		0.83	Excellent	2.6	2.20	7.50	$F_{(2,15)}$ = 0.576, P = 0.574
		3	33.6		0.12	Poor	1.1	2.80	5.00	
SLS_R EO Firm	YA	1	49.1	0.5	0.56	Fair to Good	3.2	3.17	7.76	
		2	44.1		0.71	Good	3.9	3.90	8.02	$F_{(2,18)}$ = 2.158, P = 0.144
		3	52.6		0.97	Excellent	2.5	3.50	8.43	
	OA	1	64.5	0.34	0.23	Poor	4.0	2.70	10.7	
		2	87.1		0.14	Poor	5.7	0.30	11.7	$F_{(2,15)}$ = 1.060, P = 0.370
		3	42.5		0.62	Fair to Good	1.8	3.10	6.60	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

Table 3.2: Intrasession and intersession COP_{ML} reliability data set for young and older adults

Task	Group	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
DLS EC Firm	YA	1	99.2	0.24	0.33	Poor	3.2	0.90	7.35	
		2	77.0		0.28	Poor	1.4	0.80	3.60	$F_{(2,18)}$= 4.422, P = 0.270
		3	53.2		0.20	Poor	1.3	1.90	4.56	
	OA	1	86.6	0.48	0.40	Fair to Good	1.6	0.60	3.86	
		2	87.5		0.35	Poor	1.6	0.30	3.42	$F_{(2,15)}$ = 1.990, P = 0.179
		3	48.7		0.28	Poor	0.8	0.90	2.46	
SLS_L EO Firm	YA	1	54.0	0.49	0.62	Fair to Good	3.2	4.30	10.75	
		2	79.8		1.00	Excellent	1.6	1.00	4.17	$F_{(2,18)}$ = 0.302, P = 0.743
		3	64.7		0.83	Excellent	2.7	2.60	7.94	
	OA	1	51.2	0.23	0.66	Fair to Good	1.7	1.80	5.32	
		2	72.1		0.60	Fair to Good	2.7	0.70	6.19	$F_{(2,15)}$ = 4.534, P = 0.034
		3	41.3		0.53	Fair to Good	1.3	2.00	4.57	
SLS_R EO Firm	YA	1	54.0	0.74	0.26	Poor	1.8	5.70	9.40	
		2	66.1		0.71	Good	1.5	3.50	6.49	$F_{(2,18)}$ = 2.512, P = 0.109
		3	81.9		0.29	Poor	1.1	1.90	4.17	
	OA	1	53.6	0.61	0.30	Poor	1.9	2.20	5.88	
		2	55.9		0.50	Fair to Good	2.1	1.30	5.55	$F_{(2,15)}$ = 0.526, P = 0.600
		3	59.7		0.90	Excellent	2.2	1.70	6.15	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

Table 3.3: Intrasession and intersession COP_{VL} reliability data set for young and older adults

Task	Group	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
DLS EC Firm	YA	1	15.1	0.40	0.53	Fair to Good	0.3	2.06	2.61	
		2	32.9		0.21	Poor	2.3	1.36	2.31	$F_{(2,18)}$ = 3.381, P = 0.057
		3	30.1		0.96	Excellent	2.3	1.40	2.27	
	OA	1	40.2	0.29	0.90	Excellent	0.7	1.30	2.70	
		2	32.2		0.70	Fair to Good	0.6	1.0	2.20	$F_{(2, 15)}$ = 0.910, P = 0.428
		3	30.2		0.76	Excellent	0.4	1.10	2.01	
SLS_L EO Firm	YA	1	15.4	0.3	0.14	Poor	0.5	3.78	4.82	
		2	16.0		0.89	Excellent	0.5	3.19	4.11	$F_{(2,18)}$ = 0.520, P = 0.603
		3	17.7		0.76	Excellent	0.5	2.95	3.91	
	OA	1	50.9	0.89	0.04	Poor	2.2	2.30	6.70	
		2	37.5		0.32	Poor	1.5	2.20	5.30	$F_{(2, 15)}$ = 1.425, P = 0.279
		3	40.3		0.42	Fair to Good	1.4	2.20	4.90	
SLS_R EO Firm	YA	1	19.0	0.4	0.74	Fair to Good	0.7	3.70	5.00	
		2	19.6		0.81	Excellent	0.6	3.35	4.57	$F_{(2,18)}$ = 0.477, P = 0.628
		3	14.4		0.36	Poor	0.4	2.88	3.60	
	OA	1	41.5	0.70	0.04	Poor	1.5	2.70	5.60	
		2	48.8		0.65	Fair to Good	2.1	1.90	6.10	$F_{(2, 15)}$ = 0.888, P = 0.437
		3	30.3		0.35	Poor	1.2	2.90	5.30	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

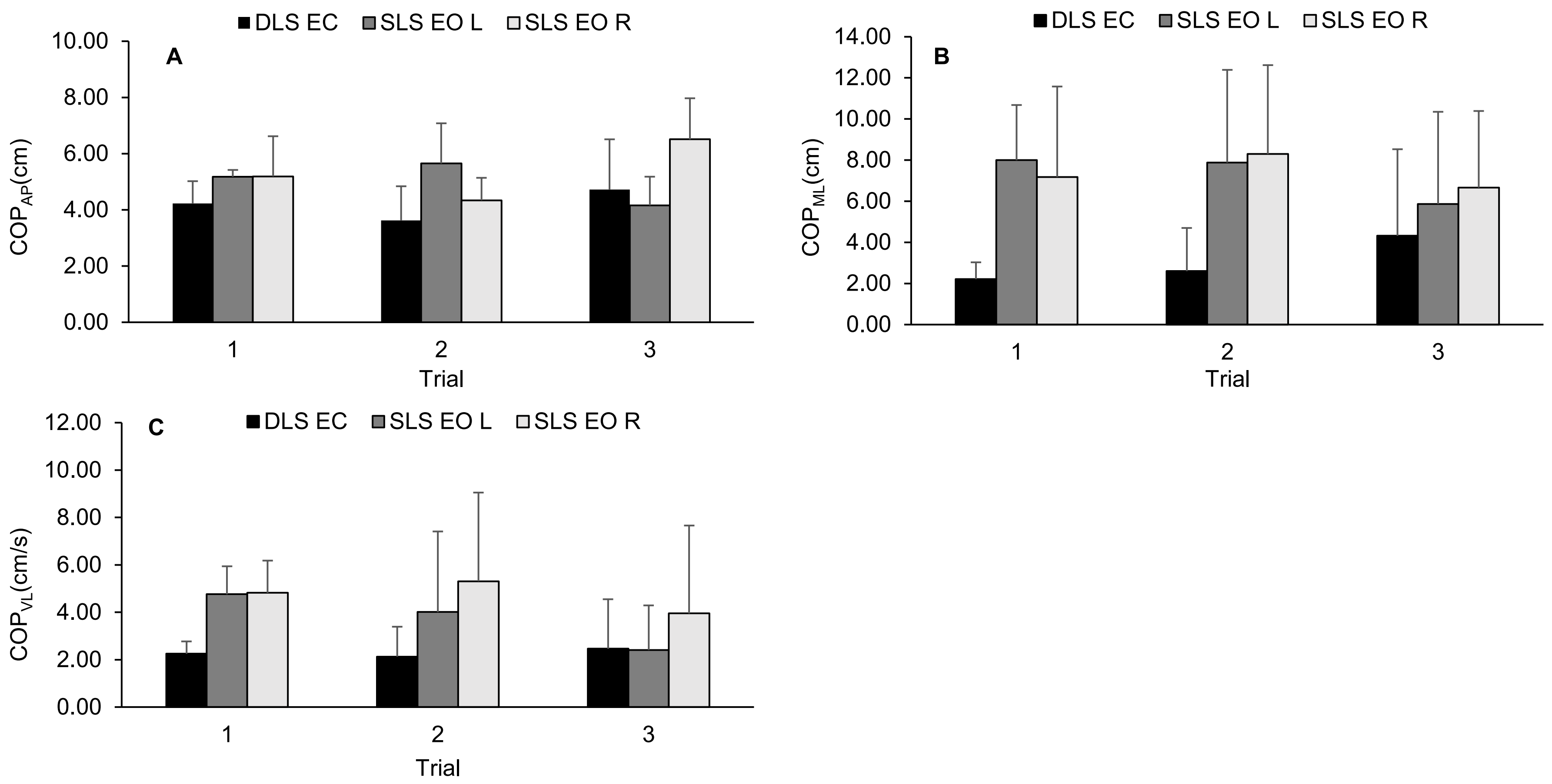

Figure 3.4: Within session reliability for young adults. (A) Centre of pressure – Anterior posterior, (B) Centre of pressure – Mediolateral, (C) Centre of pressure – Velocity. Key: DLS = Double leg stance, SLS-L = Single leg stance – Left leg, SLS-R = Single leg stance – Right leg, EC = eyes closed, EO = Eyes open.

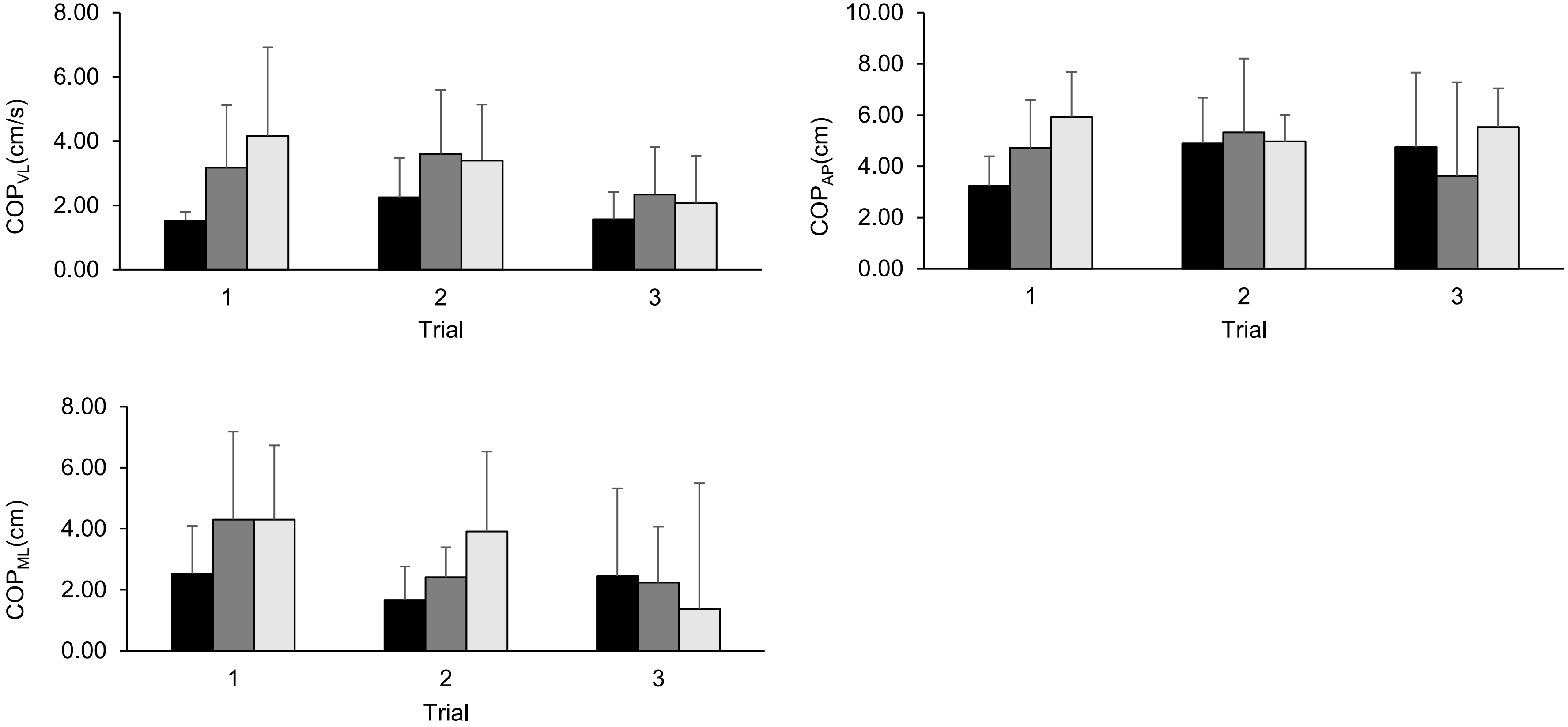

Figure 3.5: Within session reliability for older adults. (A) Centre of pressure – Anterior posterior, (B) Centre of pressure – Mediolateral, (C) Centre of pressure – Velocity. Key: DLS = Double leg stance, SLS-L = Single leg stance – Left leg, SLS-R = Single leg stance – Right leg, EC = eyes closed, EO = Eyes open.

3.5.5. Discussion

The purpose of this pilot study was to determine intra-and-inter session reliability for COP_{AP}, COP_{ML} and COP_{VL} in young and older adults based on popular clinical measures of balance. In general, most COP variables recorded "fair to good" and "excellent" ICC scores for intrasession compared to intersession reliability for young and older adults. The most reliable variable was the intrasession reliability for COP_{VL}, which is consistent with the findings of Lin *et al*. (2008), where the COP_{VL} variable was reported as the most reliable measure in young and older adults. Also, this variable reported the smallest CV across the 3-days compared to COP_{AP} and COP_{ML}. The SLS_L and SLS_R with EO stance positions were the most challenging for young and older adults compared to DLS EC. The SLS_L stance positions were the most reliable compared to SLS_R in both populations. Although all unilateral stance positions were similar in ICC for both populations, the older adults found these stance positions more challenging to perform. This is consistent with the findings from Da Silva *et al*. (2013) where unilateral positions were more challenging in older adults and reported a high ICC for COP_{VL}. For each stance position, the participant had to hold the position for 30 seconds and perform 3 trials per stance. The results here are in line with previous work from Pinsault and Vuillerme, (2009) and Wheat *et al*. (2012), where 3 trials and 30 seconds durations were deemed to be the most reliable for COP measures. Our study also identified an acceptable level of reliability based on ICC and CV, with COP_{VL} being one of the most reliable variables for all stance positions. Therefore, the following stance positions would be acceptable for future studies.

3.6. Isokinetic Dynamometry

A Biodex System 3 Pro, Rev 2 isokinetic dynamometer (Biodex medical systems Inc. NY, USA) was used to assess the dynamic function of a single joint movement through a fixed angular position (Isometric) or velocity (isokinetic or isovelocity). The dynamometer can be operated using active or passive mechanisms at pre-set angular velocities (Baltzopoulos, 2008). A clinical data station containing a Windows™ PC and operating system with Biodex advantage software (Version 4) package. The dynamometer contains a potentiometer, strain gauge and range of motion set switches that send an instant analogue or digital output to the host computer to be analysed. An auxiliary RS 232 for real-time kinetic interface and remote access for analogue signals is located at the rear of the host computer. The two interfaces allow individual signal ground, analogue position, analogue torque, and analogue velocity to be transferred to a data acquisition system (Biopac MP150).

3.6.1. Experimental Protocol

Isokinetic and Isometric ankle and knee strength was assessed using the right limb of the participant. The isokinetic dynamometer was calibrated before performing a trial (Chapter 3; section 3.6.2). All participants were required to sit upright with their back flat against the dynamometer chair at a hip joint angle of 85° (Abe *et al*., 2009) with both legs resting at 90° of knee flexion. Two restraining straps were placed diagonally across the trunk and securely locked into place to ensure that the force generated comes from the tested limb, preventing additional movement from proximal musculature. All procedures conducted during the setup of the isokinetic dynamometer were in accordance with the Biodex user's guide. The participant's chair set-up and data collected were stored and analysed using the Biodex advantage software (version 4) (Biodex Medical Systems Inc., New York, USA) for future use and access to the data stored securely on the isokinetic dynamometers computer system.

3.6.1.1. Ankle Protocol

A single ankle attachment was aligned with the red dot on the shaft of the dynamometer and securely fastened. The participant's right foot was placed into the ankle attachment ensuring the talus and lateral malleolus were lined up with the axis of rotation of the dynamometer shaft. The participant's chair position was adjusted to a tilt angle of 70° with the participant's knee placed at 120° knee and 85° hip flexion. The participant's thigh was placed onto a knee support and height adjusted to ensure 120° of knee flexion and 85° of hip flexion was achieved (Hartmann *et al*., 2009). Once in position, the straps on the footplate were checked and tightened to ensure no excess movement before conducting a trial (Figure 3.6).

Figure 3.6: Isokinetic dynamometer set up for ankle exercise.

3.6.1.2. Knee Protocol

A single-leg knee attachment was aligned with the red dot on the shaft of the dynamometer and securely fastened. The chair was adjusted to ensure that the participants the lateral femoral epicondyle of the right knee was aligned with the axis of rotation of the dynamometer lever arm, as misalignment of the knee joint during the set-up of the participant can cause errors in joint torque (Herzog, 1988; Anderson *et al.,* 2010). The distal portion of the right leg was secured with a leg brace and applied 2cm superior to the medial and lateral malleolus at the end of a solid metal bar. The participant’s right thigh was securely fastened to ensure no excessive movement from proximal musculature occurred. A joint angle of approximately 20° - 30° of knee flexion was set for the thigh support to prevent muscles of the knee and hip from contributing to the ankle movement (Figure 3.7; Hartmann *et al.*, 2009).

Figure 3.7: Isokinetic dynamometer set up for knee exercise.

3.6.2. Calibration

Calibration of the dynamometer was performed before data collection. Once the dynamometer and host computer are turned on, the axis of rotation located on the dynamometer moves and conducts an initial calibration (~10 seconds). The acqknowledge software for the dynamometer is opened on the host computer to start the calibration process. A calibrated 4.55kg countermeasure weight was attached to the metal input arm and fastened. The countermeasure was then moved into a horizontal position, with the weight furthest away from the dynamometer head (Approximately 72.5cm long). A spirit level was used to confirm the horizontal position (90°) and adjusted accordingly. The countermeasure is sensitive to vibrating movement, and a hand was placed on the countermeasure to ensure no external noise affected the raw signal input to the host computer, as this would alter the voltage transferred to the MP150WS data acquisition system (Biopac, CA, USA). The data collected on the dynamometer was exported to a laptop (Toshiba, Satellite Pro C650-18U, Japan) set up with an MP150WS data acquisition system which was used to synchronise torque produced from the dynamometer using analogue and digital signals. The information gained from the host computer was inputted into the acqknowledge (Version 4) software located on a laptop to create a graph template. The graph template (gtl*) produced on the acqknowledge software was used to record moment data and torque forces from the dynamometer. The template was duplicated for future participants and used in conjunction while assessing electromyography.

3.6.3. Gravitational Moment Correction

To account for the effects of the participant limb weight on the torque produced during data collection, the weight of the limb was recorded (Winter *et al*., 1981; Herzog, 1988; Baltzopoulos and Brodie, 1989). However, errors when measuring torque and angular velocity produced against the lever arm of the dynamometer can reduce the reliability of these measures (Drouin *et al.,* 2004). Nelson and Duncan (1983) introduced a correction factor, which considered the weight of the participant's limb once a predetermined angular position of the dynamometer arm was locked into position. The Biodex System 3 Pro performs a gravitational moment correction by measuring the weight of the tested limb on a fixed angular position (Lund *et al.,* 2015). A calculation using the gravitational moment by applying a cosine function to the gravitational moment and angular position is automatically applied through the acqknowledge software (Equation 3.1). An ICC of 0.99 for Intra and Inter-session reliability of velocity, torque and position has been reported by Drouin *et al.* (2004) and has shown acceptable mechanical reliability and validity.

$$T_g\theta_1 = Fg\mathrm{Cos}\theta 1(r).$$

Equation 3.1: The leg falls passively at any of angle of knee flexion or ankle plantarflexion (θ1), the torque (Tgθ1) caused by the force gravity (Fg) is equal to the component of the gravitational force that is perpendicular to the moment arm (FgCosθ1) times the le length of the moment arm (r) (Extracted from Nelson and Duncan, 1983).

Once the participants were set up for either ankle (Chapter 3; section 3.6.1.1) or knee (Chapter 3; section 3.6.1.2) strength assessment on the dynamometer. The arm of the dynamometer was passively moved into the participants' neutral position for the ankle (0° +/- 5°) or fully extended position for the knee (90° +/- 5°). The dynamometer arm was locked into position, and the participant was encouraged to remain as relaxed as possible. The angular position of the participant's limb was recorded using the acqknowledge software followed by the weighing of the limb. A correction factor option was selected on the software, and the weight of the limb was recorded for the participant.

3.6.4. Validation

Test-retest reliability for isometric and isokinetic movement in knee extension and flexion and ankle dorsiflexion and plantarflexion has been reported as being highly reliable for the isokinetic dynamometer (Baltzopoulos and Brodie, 1989; Drouin *et al*., 2004; Hartmann *et al*., 2009; Webber and Porter, 2010). This is determined by assessing the force that can be applied at a constant velocity through a set joint range of motion. Measures of position (°), torque (Nm) and velocity (°/s) [ranging from 20°/s – 300°/s] have been identified as being mechanically reliable (Feiring *et al*., 1990; Drouin *et al*., 2004). Drouin *et al*. (2004) reported an intra and inter-session reliability of 'near perfect' ICC scores when different calibrated weight was placed at the end of the dynamometer lever arm and assessing torque and velocity through 70° arc (30°/s – 300°/s) and isometrically. Earlier investigations have reported similar findings for ankle dorsiflexion and plantarflexion (Hartmann *et al*., 2009; Webber and Porter, 2010) and knee extension and flexion (Feiring *et al*., 1990; Brown *et al*., 1993; Gleeson and Mercer, 1996). These studies provide evidence that the isokinetic dynamometer is a mechanically reliable and valid method to quantify strength for ankle dorsiflexion and plantarflexion and knee extension and flexion.

3.6.5. Maximal Voluntary Contraction

All participants performed a series of unilateral (right leg) maximal voluntary contraction isometrically (MVIC) or maximal voluntary contraction isokinetically (MVC_{Isok}) for the ankle (dorsiflexion and plantarflexion) and for the knee (extension and flexion) performed on the isokinetic dynamometer. All participants were correctly positioned on the chair with the ankle or knee of the right limb following the previously described set-up protocol (section 3.6.1). The dynamometer lever arm was set at a joint angle of 5° (0.09 radians) for dorsiflexion and 25° (0.44 radians) for plantarflexion (Harkins *et al*., 2005) and a joint angle of 75° (1.31 radians) for knee extension and at 60° (1.05 radians) for knee flexion (Abe *et al.*, 2009) for both warm-up and experimental trials. The warm-up consisting of 3 submaximal MVIC for the ankle (dorsiflexion) and plantarflexion and contractions (~50% of maximal torque) interspersed with a 2-minute rest period between sets before completion of the MVIC trials (Fujita *et al*., 2008). This provided an opportunity for the participant to become familiar with the movement patterns and an opportunity for any adjustments to be made before the MVIC trials. A minimum of three MVIC trials was performed until two consecutive trials were within 5% of each other (Cook *et al.*, 2010). The highest force produced during the MVIC was recorded as the MVIC (Wong and Ng, 2006). The participant was instructed to perform the tests to their maximum with verbal encouragement given for motivation purposes by the investigator.

3.6.6. Force Sense

Force sense (FS) was assessed on the right leg (knee) using the same set-up position previously described for the isokinetic dynamometer (Chapter 3; section 3.6.1) to establish the participant's MVIC (section 3.6.1). After performing the MVIC knee extension, a force-matching process was conducted using 10 and 20% of the participant's MVIC and performed in a randomised counterbalanced order (Docherty and Arnold, 2008). The participants' leg was placed into a single-leg attachment and securely fixed to the dynamometer head. A spirit level (Stanley, New Britain, USA) was placed along the shaft of the dynamometer and used to assess the participants resting knee joint position (90° knee flexion). The position of the knee was recorded on the dynamometer computer and set at 90° before the participants were moved into 70° of flexion. The participants were instructed to extend their legs against the fixed single-leg attachment to achieve a target force (10 or 20% MVIC) based on the digital readout display on the dynamometer. All participants were asked to maintain the contraction for 5 seconds before relaxing for 30 seconds. To prevent a learning effect, all participants had no visual feedback of the torque generated (Wright and Arnold, 2012) and were required to attain the same target force as previously produced (Docherty and Arnold, 2008; Wright and Arnold, 2012). All participants were asked to inform the investigator once they perceived that the target force was reached. Following this, three practice trials were completed to ensure participants fully understood the procedure. Once the practice trials were finished, the participants were asked to perform five trials at 10% and then 20% of the individuals MVIC (contraction held for 5 seconds) interspersed with 30 seconds of rest between trials. The data collected was analysed using the Biodex advantage software which consisted of taking an average of each 5-second trial performed at each percentage (e.g., 5 contractions at 10% and 5 contractions at 20%).

3.6.7. Joint Position Sense

The participant's joint position sense (JPS) was conducted using the same seated position as previously described (section 3.6.1). The participant's right leg was securely fastened to a single-leg attachment. A blindfold was placed over the participant's eyes to prevent any visual cues to the participant's joint position (Kim, Choi, and Kim, 2014). All participants were assessed at 45° and 70° of knee extension and flexion. For knee extension, the participant's limb was passively moved from the resting position (90° knee flexion) by the investigator to either 45° or 70° of knee extension in a randomised order (Olsson *et al*., 2004; Ribeiro *et al*., 2007b). For knee flexion, the participant's limb was passively moved from the resting position (0° knee extension) to either 45° or 70° of knee flexion in a randomised order. All participants performed 3-practice trials before completing the protocol, with the participant's limb placed back in the resting position for 5 seconds before attempting to reproduce one of the four joint angles. Each trial was separated with 5 seconds of rest, with three trials performed for each joint angle (Olsson *et al*., 2004). A handheld remote was given to the participant in their dominant hand, and was instructed to press the button on the remote once the participant felt the target joint angle was achieved. Once the button was pressed, the dynamometer arm stopped, and the joint angle was recorded with the average of the three trials used for analysis (Figure 3.8).

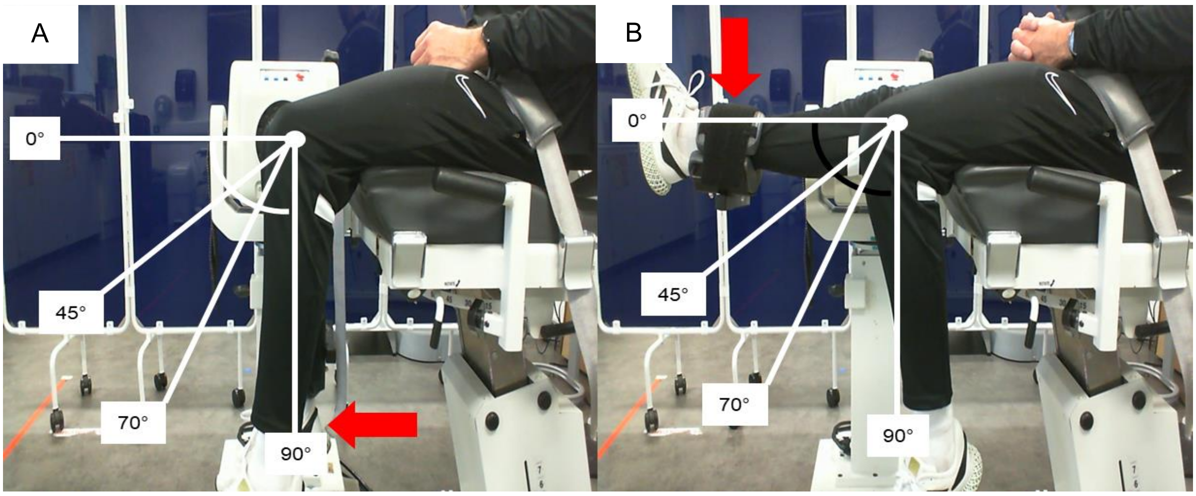

Figure 3.8: Dynamometer set up for knee extension (A) and Flexion (B) for joint position sense. Red arrow denotes start position before repositioning task.

3.7. Electromyography

A Biopac MP150WS data acquisition system with EMG100c (Biopac Systems Inc., Goleta, CA, USA) was used to measure and record surface electromyography (sEMG) signals produced during muscle activity. A dual wireless transmitter (BioNomadix; Biopac Systems Inc., Goleta, CA, USA) was used to record two independent muscle activities simultaneously (Signal detection: Chapter 3; section 3.7.2). The Biopac system processes the data using an algorithm and is converted using an analogue-to-digital conversion. The analogue input has a voltage range of ±10V (force sensor resistor of ± 0.003%) with an input impedance of 10MΩ. The amplitude of the signal is detected and converted to a digital signal that generates a numerical sequence (analogue-digital conversion). The analogue signal was sampled once this process was completed.

3.7.1. Site Preparation

All participants were informed before attending a data collection session (Chapter 4 and Chapter 5) that the shaving of body hair and abrasion to the skin maybe necessary to improve skin contact and reduce noise when applying the surface electrodes to the muscle. If this was required, the removal of body hair was conducted using a disposable razor before a slight abrasion to the skin using a scourer pad was applied and cleaned using an ethanol wipe (Isopropyl alcohol 70%; Cutisoft, Physique medical, UK) to remove excess dirt and oil. Two pre-gelled disposable stud surface electrodes (1.1cm diameter; Ag/AgCl contact; 4% Chloride salt; Figure 3.9) (Biopac EL500, Linton Instrumentation, UK) with a 4.1cm vinyl back were placed bilaterally over the belly of the muscle (anterior tibialis, medial gastrocnemius, vastus medialis and lateralis; Figure 3.10) and taped down using micropore tape (3M, 2.5cm x 9.1m; Fit4Sport, UK) to secure and reduce excessive movement. An inter-electrode distance of 4.4cm was used to minimise any crosstalk between surface electrodes (Hermens *et al.*, 2000; González-Izal *et al.*, 2012) with an additional electrode placed on a bony landmark close to the proximity of the target muscle (Figure 3.10).

All participants were encouraged to walk for 5 minutes with the electrodes attached to ensure contact to the skin was maintained during movement and whether the electromyography reading is reliable when standing still. A laptop installed with the Biopac AcqKonwledge 4.4 software (Biopac systems Inc., Goleta, CA, USA) was used to process the data.

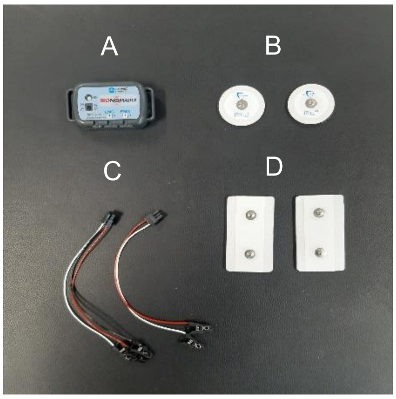

Figure 3.9: Surface electromyography equipment. (A) BioNomadix transmitter, (B) Single electrode, (C) 2 x 30cm leads and (D) 2 x stud pre-gelled electrodes.

3.7.2. Signal Detection

The Biopac data acquisition system sampled the data at a frequency of 2000Hz and was smoothed using a low to high, 20Hz to 500Hz band pass filter (Dantas *et al*., 2010) by a 16-A/D converter to a laptop. All real-time data were recorded and analysed using the acqknowledge software with the average root mean square calculated with a 250ms sample window. The application and correct positioning for sEMG followed the SENIAM (Surface electromyography for the non-invasive assessment of muscles) guidelines (Hermens *et al*., 2000).

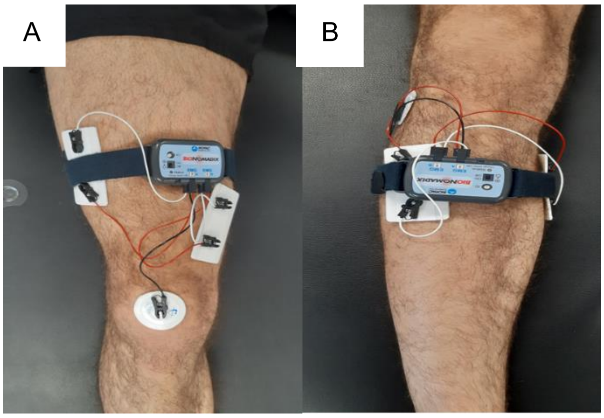

Figure 3.10: Surface electromyography set up for (A) Vastus lateralis and vastus medialis (reference electrode – patella) (B) Anterior Tibialis and Medial Gastrocnemius (reference electrode – head of fibula).

3.8. Functional Balance Tests

3.8.1. Sit to Stand

All participants were instructed to sit and rise from a chair (Height: ~43cm) as quickly as possible with arms folded across their chest for 5 x repetitions (5-STS, Bohannon, 1995; Goldberg *et al*., 2012). The time was recorded using a digital stopwatch (nearest 10th of a second) and started once the participant was instructed with the word "Go" and finished when the participant stood in an upright position on completion of the 5th repetition (Figure 3.11; Takai *et al*., 2009). A practice trial was included to familiarise themselves with the position of the chair and was performed at submaximal effort (Kanehisa and Fukunaga, 2014). The participant performed the 5-STS test twice with a 1-minute break between trials. The time (STS^T) and power (STS^P) were calculated using the following equation (Equation 3.2):

$$\mathrm{STS}^{P}\ \frac{(Leg\ length - 0.4)\ \mathrm{x\ body\ mass\ x\ g\ x\ 5}}{\mathrm{STS^{T}}}$$

Equation 3.2: Test-retest reliability for the STS has been reported with an ICC ranging from 0.64 – 0.84 demonstrating good to excellent reliability (Bohannon, 2011).

Previous research has reported that the 5-STS test has excellent test-retest reliability (ICC = 0.95-1.00; Wallmann *et al.*, 2011).

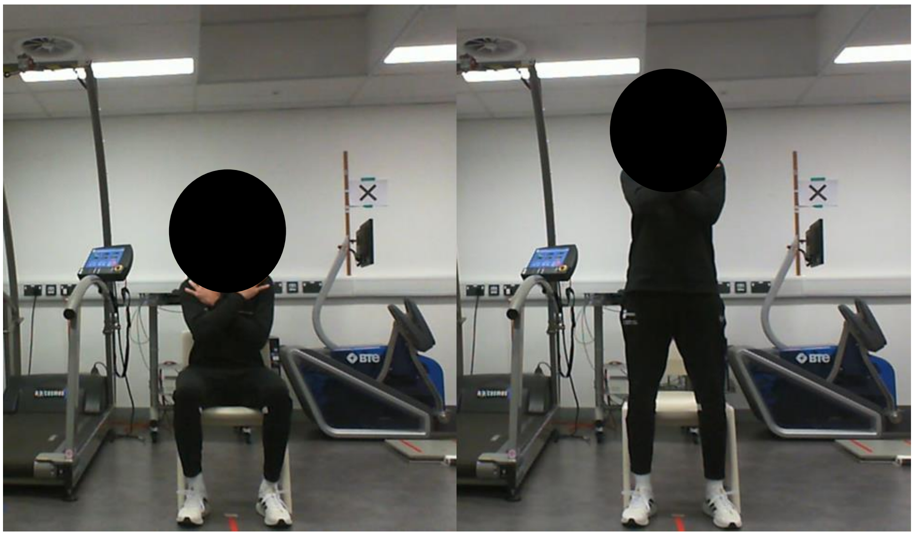

Figure 3.11: Sit to stand protocol.

3.8.2. Timed Up and Go

The participants were asked to sit upright in a chair (Height: ~43cm) with feet flat on the floor behind a taped marked line (Shumway-Cook *et al.*, 2000). A 3-metre walkway was measured from the taped line near the position of the chair, with another taped line placed on the floor at the 3-metre mark. All participants were instructed to stand up from the chair, walk to the 3-metre line, cross the taped line, turn around and walk back to the chair and sit down (Podsiadlo and Richardson, 1991; Shumway-Cook *et al.*, 2000). The timing of the trial was recorded using a conventional stopwatch (seconds) and commenced once the participant was given a verbal command of “go” and finished when the participant sat back down with feet flat on the ground behind the line (Figure 3.12). The participant had one practice trial followed by 3 x trials with a one-minute rest between trials. The average time from the 3 x trials was used for data analysis. Previous research has reported that the timed up-and-go test has excellent test-retest reliability (ICC = 0.95-0.97; Steffen *et al.*, 2002).

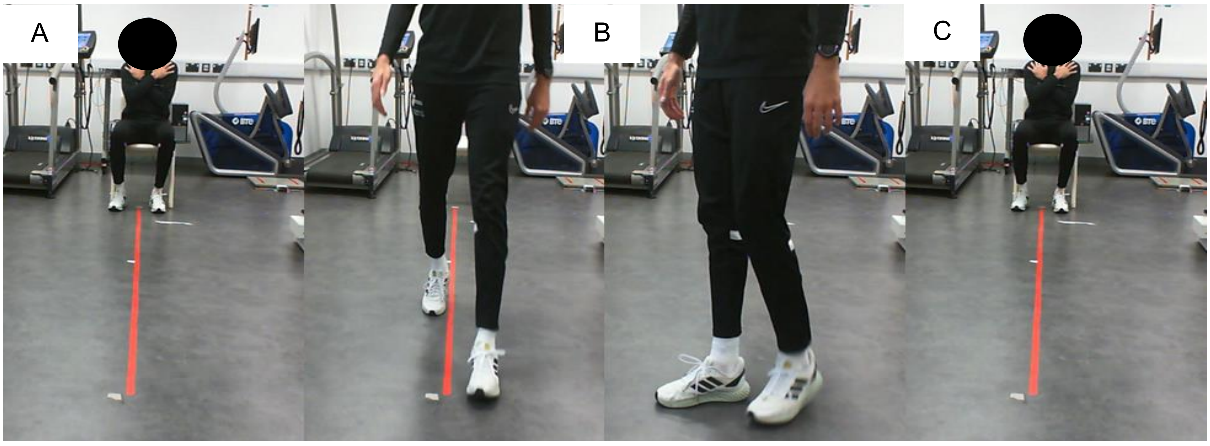

Figure 3.12: Timed up and go start (A), turn (B) and finishing position (C).

3.8.3. Gait Stability

Each participant was instructed to walk at a comfortable pace on a motorised treadmill (H/P/Cosmos Gaitway, Traunstein, Germany) with two tandem in-built force platforms (Kistler, Winterthur, Switzerland) located under a single-track rubberised belt. The force platforms were calibrated and adjusted before each trial which conformed to the manufacturer's protocol using the Gaitway® software package. The participants were asked to stand still in the middle of the first force platform at the rear of the treadmill to determine body mass before completing the walking trial. All participants were asked to remove footwear and walk barefoot in the middle of the treadmill's single-track belt. The investigator adjusted the treadmill speed in 0.2km/h increments in response to the participants' instructions of 'slower' or 'faster' who were blinded from the walking speed (Roberts *et al*., 2018). All participants were asked to look forward and avoid looking down towards the treadmill belt. The trial was discarded if the participant did not maintain pressure across the two force platforms or if both feet met one of the force platforms (Figure 3.13). All participants walked for two minutes on the treadmill, with data recorded in the final minute. Ground reaction forces were sampled at 200 Hz, and the gait indices collected from each trial were mean stride time (s), stride length (cm) and double-limb support time (s) to assess gait stability (Verghese *et al*., 2009).

3.8.4. Gait Stability Measurements

The motorised treadmill measures the vertical ground reaction force generated from the two-tandem piezoelectric force plates by using an algorithm to denote left and right foot strikes. The treadmill is supported by an eight-channel charge amplifier, foot discriminator circuit and belt sensor where data is collected via the patented Gaitway® software from the host computer. The software collects and analyses the data from the vertical ground reaction forces generated from the force plates located in the bed of the treadmill (Cojan-Carlea *et al*., 2015; Roberts *et al*., 2018). The data collected can be separated from the left and right foot strikes and averaged to provide a mean. The double support time (s) is the time taken from the heel strike of one foot up (e.g., left) to the toe-off phase of the other foot (e.g., right). The stride time is measured from the contact (time; s) of the initial contact of the left foot to the subsequent contact of the left foot. Finally, the stride length is the distance between successive points of initial contact measured from the left foot and right foot (Cojan-Carlea *et al*., 2015).

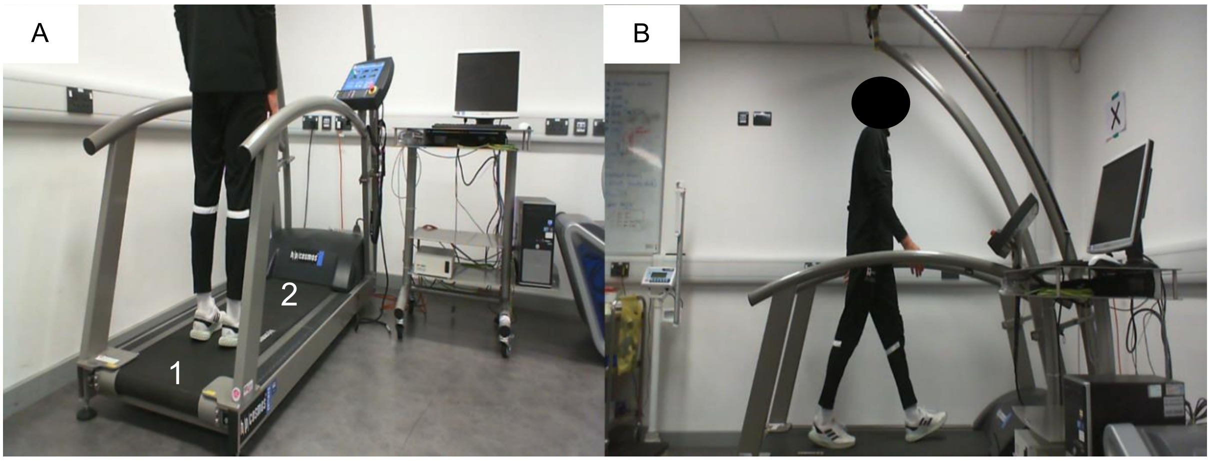

Figure 3.13: Set up for gait stability. (A-1) location of the distal force platform used to weigh participant prior to performance, (A-2) location of second force platform located at the proximal end of the treadmill, (B) participants were asked to straddle between both force platforms (A-1 and A-2) while performing the walking task.

3.9. Physiological Measures

3.9.1. Blood Pressure

The Omron M3 (HEM-7134-E) electronic automatic blood pressure monitor was used to measure systolic (SBP) and diastolic blood pressure (DBP) from the brachial artery using an oscillometric method (range of 0 – 229 mmHg) and pulse rate (range 40-180 beats/min; Coleman, 2006) with an accuracy of ± 3mmHg and ± 5% beats/min. The oscillometric method works by detecting vibrations from the brachial artery wall, which can be transduced into electric signals to determine SBP and DBP (Berger, 2001; Babbs, 2012). The blood pressure monitor is controlled by a fuzzy-logic electric pump with an automatic release valve and is responsible for detecting changes in cuff pressure during inflation and deflation of the blood pressure cuff (Berger, 2001). During inflation of the blood pressure cuff (approximately 20mmHg above predicted SBP; Berger, 2001), the external pressure applied to the arm reduces arterial blood flow causing temporary occlusion (Babbs, 2012). As the blood pressure cuff deflates and blood can flow from the artery, the monitor transfers the vibrations detected from the arterial wall through the air inside the cuff. The transducer in the monitor converts the measurements from the vibrations detected in the cuff into electrical signals to determine SBP. When the blood pressure cuff deflates to a point where no vibration is detected in the arterial wall, then this would signify DBP.

3.9.2. Validation

The Omron M3 automatic blood pressure monitor has been validated (Alpert *et al*., 2010; O’Brien *et al*., 2010 Topouchian *et al*., 2011). In addition, the Omron M3 fulfils the criteria set by the European international society of hypertension international protocol revision 2010 (Akoplat *et al*., 2012). The international protocol was established by a working group of experts in measuring blood pressure to ensure the accuracy of the automatic blood pressure monitor were valid for commercial use since the demise of mercury-containing sphygmomanometers (O’Brien *et al*., 2010 Topouchian *et al*., 2014). More information describing the protocol and standards used to validate the device (post-2010) can be viewed in the article by O’Brien *et al*. (2010) and a downloadable copy from www.dableducational.org.

3.9.3. Resting Systolic Blood Pressure

Target BFR pressure was determined before each study (Chapters 4, 5 and 6) using a standardised protocol established from the pilot study in young and older adults (Chapter 3: section 3.10). To examine resting SBP, participants were required to remove all footwear and attain a seated position on a portable couch with the back raised in a dimly lit room (Hunt *et al*., 2016). A blood pressure cuff was applied around the proximal portion of the right arm before the start of the 10-minute resting period. The position of the blood pressure cuff was approximately 3cm superior to the olecranon process of the right arm (Hunt *et al*., 2016). The right arm was then abducted at 80° away from the torso (Hunt *et al*., 2016); the shoulders at 10° of flexion and the elbow at 160° of extension, measured using a manual goniometer. All participants remained quiet, with their eyes open during the rest period. A total of 3 x blood pressure measures were taken in the final 2 minutes (30 seconds each).

3.10. Pilot Study 2: Intra and inter-session reliability of resting systolic blood pressure in young and older adults

3.11. Introduction

Resting SBP is subject to change either daily (intersession: between) or through diurnal rhythm (Intrasession: within; O'Brien and Atkins, 2007). Assessing the reliability of the blood pressure monitor and the measures it provides is essential to determine resting SBP for BFR training. The application of LIBFR is based on multiplying (ranging between 1.2 – 2.0) a participant's resting SBP. The SBP reading taken from the blood pressure monitor needs to be reliable to ensure the correct pressure applied to the BFR thigh tourniquets is consistent for BFR training and to reduce the chance of occlusion to the femoral artery. Previous studies have referred to the importance of prescribing resting SBP for BFR studies by ensuring the initial set-up is consistent and repeatable to reduce systematic or random errors (Loenneke *et al.,* 2012c; Karabulut and Perez, 2013; Hunt *et al.,* 2016). Although, there is a lack of evidence on inter-session and intra-session reliability data for calculating resting SBP for BFR training. The main objectives of this pilot study were to (1) determine intra-and-inter-session reliability of resting SBP based on young and older adults from a normal population and (2) establish a standardised and repeatable protocol to assess resting SBP during rest.

3.12. Methods

3.12.1. Participants

Ten healthy young (age; 25.7 ± 3.6 years) and seven older adults (age; 72.0 ± 7.7 years) male (8) and females (9) were recruited through a convenience sampling method and were required to attend the laboratory on three occasions, separated by 24 hours. Participants were excluded from the study if they did not meet the exclusion criteria described in (Chapter 3; section 3.3.1 and 3.3.2).

3.12.2. Procedure

Resting SBP was recorded over three separate days. During each visit, rest SBP was taken three times. All participants were asked to remove all footwear and attain a seated position on a portable couch with the back raised in a dimly lit room. An automatic blood pressure cuff (Omron M3, Matsusaka Mie, Japan) was applied firmly around the proximal portion of the right arm at approximately ~3cm superior to the olecranon process of the right arm (Hunt *et al*., 2016). The participant's right arm was abducted to 80° away from the torso (Hunt *et al*., 2016) with the shoulders at 10° of flexion and elbow at 160° of extension. The participants were asked to remain quiet throughout the assessment with their eyes open. A 10-minute resting period started once the blood pressure cuff air hose was connected to the air socket of the blood pressure monitor. At the end of the 10-minute resting period, 3 x blood pressure measurements were taken with a 30-second rest interval between each measure (Hunt *et al*., 2016). The average of the 3 x SBP measurements was taken from the data collection session (Figure 3.14). Intrasession reliability was calculated using data from the final visit.

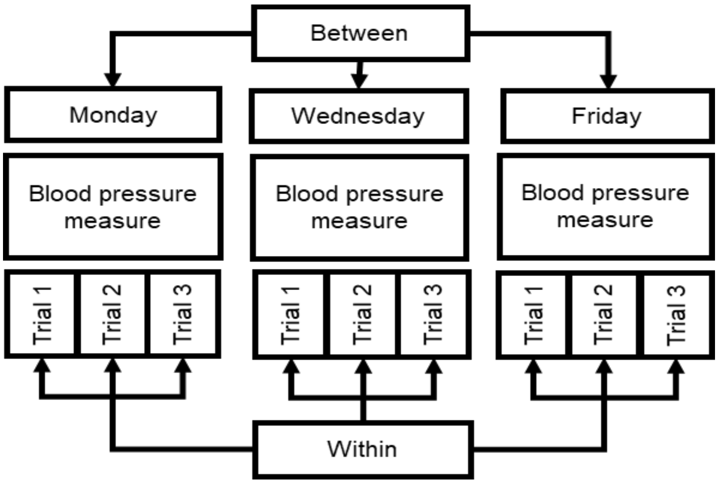

Figure 3.14: Intrasession and intersession reliability schedule for measuring resting systolic blood pressure in young and older adults.

3.12.3. Statistical Analysis

Data was analysed using SPSS version 22. (Chicago, IL, USA). A Shapiro-Wilk test was used to assess normal distribution of the data set and a Mauchley test of sphericity to verify homogeneity of variance. The reliability of resting SBP was determined using intra-class correlation coefficient (ICC) and typical error expressed as the coefficient of variation. According to Hopkins *et al*. (2009) an ICC value of 0.9 – 1.0 should be used as a threshold indicating reliability. Coefficient variance was collected from each individual and calculated using the standard deviation and the mean of three blood pressure measures (CV = [(SD/mean) x 100]. Mean ± Standard deviation (SD) for the within and between session variation are also reported. A 95% confidence interval include upper and lower limits was calculated using a specially formatted Excel spreadsheet. A One-way ANOVA was performed on between-session variation only. For the between-session variation, mean ± SD are reported as an average of three trials. Within session variation is reported as individual trials.

3.13. Results

No significant differences were observed in both young and older adults in resting brachial artery systolic blood pressure ($P \geq 0.05$) for between session variability ($F_{(2, 51)} = 1.354$, $P = 0.267$) (Figure 3.15). Minimal difference in were observed when the average of each trial was taken to display within session variation (Figure 3.16, Table 3.4).

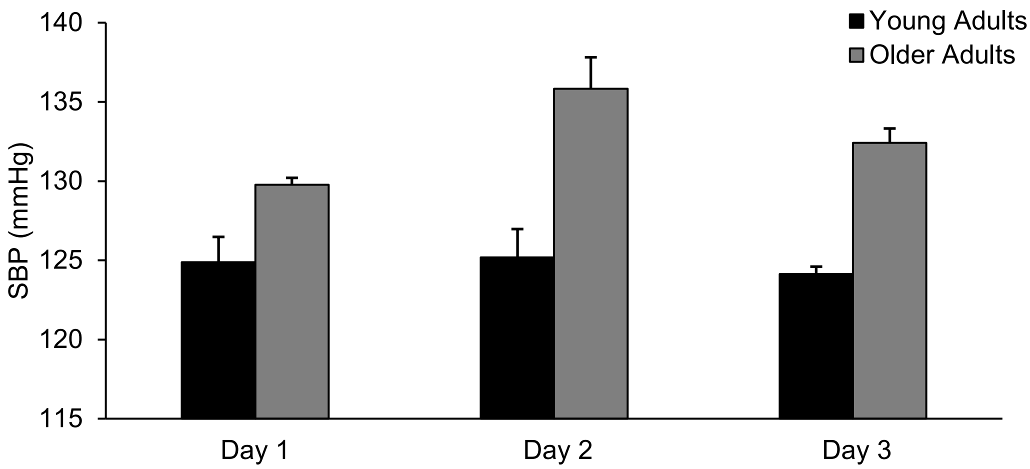

Figure 3.15: Between session reliability data for young and older adults.

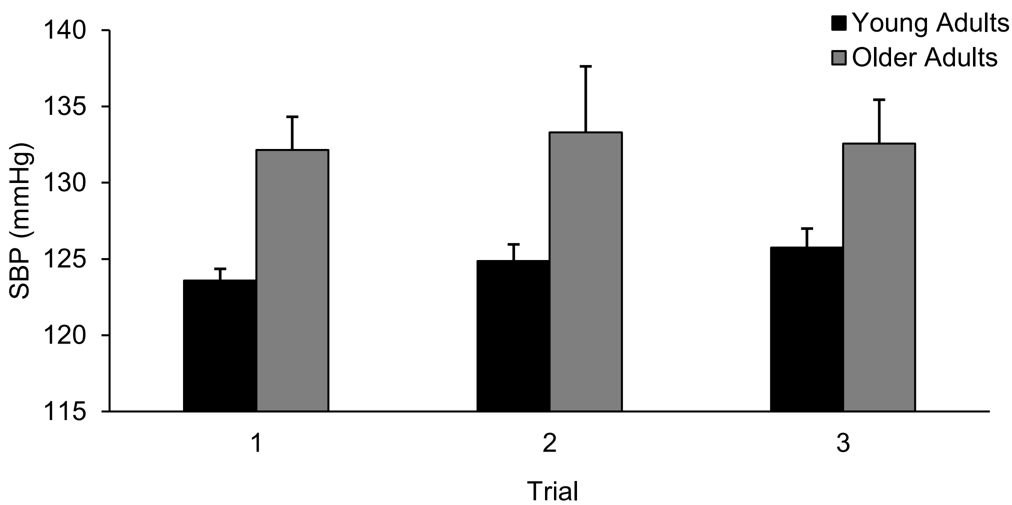

Figure 3.16: Within session reliability data for young and older adults.

Within session resting rachial artery SBP was consistently reliable for both young and older adults (Table 3.4). The ICC values ranged from 0.86 - 0.97 and 0.89 - 0.95 for young and older adults respectively. The consistency of the resting SBP was also demonstrated through a small CV ranging from 4.5 - 8.2% in young adults and 6.9 - 9.2% in older adults (Table 3.4).

Table 3.4: Intrasession reliability for young and older adults.

	Day	CV (%)	ICC	ICC Description	CI	Lower Bound	Upper Bound
Young	1	4.5	0.86	Excellent	4.5	120	129
	2	4.6	0.94	Excellent	4.6	121	130
	3	8.2	0.98	Excellent	3.8	120	128
Old	1	8.8	0.89	Excellent	0.2	129	130
	2	9.2	0.97	Excellent	1	135	137
	3	6.9	0.95	Excellent	0.5	132	133

Key: CV = Coefficient of Variance, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval. ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010, Koo and Li, 2016).

3.14. Discussion

This pilot study aimed to determine whether measuring SBP for intra-and-inter session reliability was consistent across young and older adults for prescribing lower limb BFR. Overall, the results were “excellent” for intra-and-inter session reliability, with a relatively low CV% reported. The findings from this pilot study are consistent with previous studies on intrasession reliability (Climie, *et al*., 2012; ICC = 0.98; Barrios-Fernandez *et al*., 2022; ICC = 0.90) when a seated position is adopted during a 10-minute rest period to determine BFR pressure (Hunt *et al*., 2016). The results from the study confirm that taking SBP using a set resting protocol was reliable and is appropriate for young and older adults in future studies.

3.15. Thigh Tourniquet

The Occlusion Cuff® is an aneroid sphygmomanometer consisting of an 85cm (L) x 7cm (W) nylon tourniquet with an inflatable bladder that is operated manually using a bulb pump and manometer. The tourniquet is designed for lower or upper body BFR and can inflate to a blood pressure range of 0-

300mmHg. Inflation of the bladder is controlled by a neoprene inflation bulb pump attached to the bottom of a metal valve. The valve has a knurled cylindrical shaped air release valve used to deflate the tourniquet. A glass-fronted manometer with an analogue needle display is located above the release value and is used to indicate systolic blood pressure. A protruding metal air stem connects to a rubber bladder tube attached to the bladder within the cuff. After the Occlusion Cuff® has been inflated to the desired systolic blood pressure, the rubber tube can be disconnected from the cuff. At this point, the participant is free to move while still wearing the cuff. (Figure 3.17).

Figure 3.17: Thigh tourniquet used to blood flow restriction.

3.16. Thigh Tourniquet Application

Two thigh tourniquets were applied bilaterally with the participant in a supine position. Participants were required to move their knee to 45° of knee flexion. The thigh tourniquet was applied firmly to the upper portion of the thigh at approximately 5cm inferior to the border of the femoroacetabular joint (Figure 3.18). Once the tourniquet was in place, the velcro straps on the outside of the cuffs were securely fastened. The rubber tubing was then connected to the manometer, and any residual pressure in the bladder was deflated (Figure 3.17). Adjustment to the position of the cuff was made if residual pressure in the bladder was identified, in accordance with Karabulut *et al.* (2013).

The participant was then instructed to extend their leg back to the fully extended position. The cuff pressure was set to reduce the possibility of lower limb occlusion to the femoral artery and to achieve venous restriction (Takarada *et al.,* 2000a; Abe *et al.,* 2006; and Loenneke *et al.*, 2015). To obtain target BFR pressure, the tourniquet was inflated in a staggered fashion. The target SBP for lower limb BFR was set at 1.2 x the individual's resting SBP. The tourniquet was initially inflated to 50mmHg systolic blood pressure for 30 seconds and deflated for 10 seconds. The tourniquet pressure was inflated to the participant's resting SBP for 30 seconds and deflated for 10 seconds. The tourniquet was then set to the participant's target systolic blood pressure. If the blood pressure was not maintained, the tourniquet was blown up if necessary and deflated if pain or discomfort was experienced.

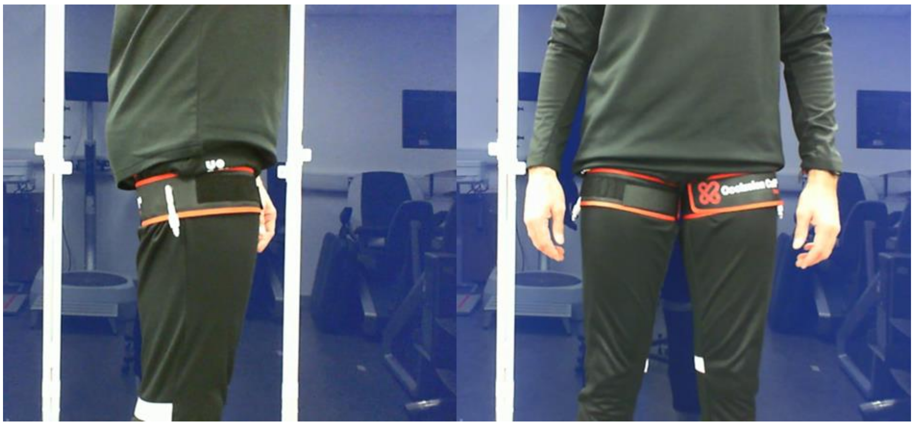

Figure 3.18: Thigh tourniquet application and position for blood flow restriction exercise.

3.17. Subjective Measures

Body mass (kilograms; Kgs) and height (centimetres; cm) was recorded before each study (Chapter 4, 5 and 6) using an electronic set of weighing scales (SECA 761, Hamburg, Germany) and portable Leicester stadiometer (Tanita, Leicester, UK).

Chapter 4:

The Effects of Lower Limb Muscle Fatigue with and without Blood Flow Restriction on Postural Sway and Gait Characteristics in Healthy Young Adults

Highlights:

- Few studies have examined the effects of muscle fatigue with blood flow restriction (BFR) on postural sway and gait stability in young adults.
- This study compared the effects of acute ankle and knee muscle fatigue with and with no blood flow restriction (NBFR) on postural sway and gait stability in young adults.
- Despite reductions in muscle torque and muscle activation, the fatigue protocols elicited minimal effects on postural sway or gait stability.

4.1. Introduction

Maintaining postural control is a complex task where the central nervous system continually integrates and re-weighs information from the visual, vestibular, and proprioceptive sensory afferents while modulating appropriate motoric commands (Assländer and Perteka, 2014). Muscle fatigue emanating from repeated contractions can result in a temporary contractile dysfunction (Gandevia, 2001), which can compromise the quality of sensory information (Lepers *et al*., 1997) by altering the sensorimotor information the central (progressive failure of voluntary neural drive) and peripheral (peripheral weakness and/or more localised muscle fatigue) nervous systems (Harkins *et al*., 2005; Christie and Kamen, 2009; Springer and Pincivero, 2009). For example, peripheral fatigue increases metabolites (e.g., blood lactate) that can decrease muscle activation and alter torque production increasing the demand to maintain a bipedal upright stance (Dickin and Doan, 2008; Bisson *et al*., 2010). Several studies investigating the effects of muscle fatigue to either the ankle plantarflexors (Harkins *et al*., 2005; Dickin and Doan, 2008) or knee extensors (Duarte *et al*., 2010; Nam *et al*., 2013) have reported an increase in the centre of pressure (COP) excursions during quiet standing (e.g., postural sway) and/or reduced gait stability.

Although the mechanisms of muscle fatigue on balance control are task-dependent (e.g., related to the intensity, duration, active muscle mass and type of muscle action [Enoka and Duchateau, 2016]), the influence of some key components of fatigue on postural control has not yet been quantified. Low-intensity resistance exercise with BFR has the potential to increase muscle hypertrophy, strength, and torque (Rolnick and Schoenfeld, 2020) but also induces neuromuscular fatigue (Cerqueira *et al*., 2017). For example, low-intensity resistance exercise with BFR elicits a similar magnitude of reduction in maximal voluntary contraction at the ankle (25-50% 1RM decline in ankle plantarflexor torque [Patterson and Ferguson, 2009]) and knee (20% 1RM [Cook *et al*., 2017]) when compared to high-intensity resistance exercise. The mechanisms responsible for the initial reductions in torque under BFR conditions remain unclear, although the addition of BFR can alter the recruitment of muscle fibres possibly attenuating muscle activation (Kjeldsen *et al*., 2019). During BFR training, there is a reduction in oxygen supply and an increase in metabolites such as lactate and hydrogen ions. This results in effects that mirror those of high-intensity resistance training and could potentially disturb postural control in fatiguing conditions. This raises an important question: what are the differences in fatigue induced by low-intensity contractions with no BFR vs. low-intensity contractions with BFR on indices of quiet standing balance and gait performance?

Fatigue experienced under traditional muscle fatiguing exercise protocols (e.g., repeated bouts of voluntary knee extension exercise performed on an isokinetic dynamometer) can affect the central and peripheral nervous systems resulting in impaired postural control (Bisson *et al*., 2010; Paillard, 2012). However, the acute effects of BFR combined with exercise on postural control remain unknown. With the popularity of low-intensity resistance training with BFR being used for rehabilitation to prevent muscle atrophy (Kubota *et al*., 2008; Loenneke *et al*., 2013; Hughes *et al*., 2017), determining the potential negative (or lack of) side effects of low intensity BFR exercise on balance in young adults is a crucial first step to understanding the potential risks and long-term benefits of BFR exercise. Based on these considerations, the present study aims to investigate the effects of lower limb muscle fatigue with BFR and with no BFR (NBFR) on postural sway and gait characteristics in young adults.

Research Hypothesis (H_1): Muscle fatigue performed using repeated maximal voluntary isokinetic contraction protocol will elicit a marked reduction in balance (increase in postural sway), gait performance (longer time during the double limb support stance, increase in stride time and shorter stride length) and muscle performance (reduction in torque and increase in EMG amplitude) at the ankle and knee following the fatiguing protocol, with these effects more pronounced in the BFR condition.

4.2. Methods

4.2.1. Participants

Eight recreationally active male (*n* = 5) and female (*n* = 3) young adults (age 26.75 ± 3.58 years, height: 173.76 ± 13.36cm, weight: 76.19 ± 16.96 kg) volunteered to participate in the study. The inclusion and exclusion criteria used for this study can be viewed in Chapter 3; section 3.3.1 – 3.3.2. A sample size calculation using *a priori* power (G*Power version 3.1.9.2, Universitat Kiel, Dusseldorf, Germany; Faul *et al*., 2007) for repeated measures identified a minimum of 28 participants would be required to detect a medium standardised effect (Cohen's f = .25 with α 0.05 and a 1-β error of 0.80) for one group and two within subject measures (Appendix 9.3.1). A *post hoc* power calculation was performed based on the 8 participants recruited in the current study (medium standardised effect of Cohen's f = .25 with α 0.05 and a 1-β error of 0.26 [Appendix 9.3.1]).

4.2.2. Experimental Procedure

The study was designed as a repeated measures study (Figure 4.1), with each participant visiting the laboratory on four separate occasions to perform the following conditions in a random order: (1) knee extensor fatigue with BFR (2) knee extensor fatigue with no BFR (NBFR) (3) ankle plantarflexor fatigue with BFR (4) ankle plantarflexor fatigue NBFR (Figure 4.1).

On the initial visit, all participants completed the necessary informed consent paperwork (Chapter 3; section 3.3) followed by the collection of anthropometric data collected using the procedures identified in Chapter 3; sections 3.9.1, 3.9.3 and 3.17. Thereafter, a double bipolar surface EMG electrode was placed over the muscle belly of the tibialis anterior (EMG_{TA}), medial head of the gastrocnemius (EMG_{GM}) for the ankle fatiguing protocol or the vastus medialis (EMG_{VM}) and vastus lateralis (EMG_{VL}) for the knee fatiguing protocol on the participant's right leg (Chapter 3; section 3.7). A single EMG electrode was placed over two bony prominences for the knee (patella) and ankle (head of the fibula). Quiet standing balance using a double leg stance (DLS; eye closed) and single leg stance (SLS; eyes open - right leg) was measured on a force platform to assess postural sway amplitude (PS), followed by gait assessments (walking gait [timed gait and ground reaction forces]). The outcome measures were assessed before and immediately after fatigue protocols and following a 10-minute passive recovery. For the BFR conditions, two nylon thigh tourniquets were applied to the upper portion of the right and left thigh whilst participants (Chapter 3; section 3.15 – 3.16). The tourniquets remained deflated until the participant was seated in the chair of the dynamometer and ready to perform the fatiguing trial.

The fatigue protocol consisted of repeated unilateral (right leg) maximal voluntary isokinetic contractions (MVC_{ISOK}; concentric/concentric) performed between 5 – 15° for ankle dorsiflexion and plantarflexion and at 90 – 120° for knee extension and flexion at a velocity of 60°/s (Gribble and Hertel, 2004). The detailed procedure for the ankle and knee exercise protocols can be found in Chapter 3; section 3.6. An initial submaximal warm-up was performed on the dynamometer to familiarise the participant with the protocol and to establish an initial maximal voluntary isometric contraction (MVIC) (3 repetitions x 3-s holds – 1 min rest between). The average of the three MVIC scores was taken, and 50% of the mean MVIC was calculated (Dickins and Doan, 2008) and used to establish baseline maximal torque production before the fatiguing trial and to normalise the EMG amplitude. For the BFR trials, the tourniquets were inflated to 1.2 x resting systolic blood pressure (SBP) (Chapter 3; section 3.15 - 3.16) before starting the fatigue trial. For each fatigue trial, all participants completed ankle (dorsiflexion/plantarflexion) or knee (flexion/extension) contractions until the MVC_{ISOK} had decreased to 50% of the baseline value. This is a standard protocol used to induce fatigue in postural control studies (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Harkins *et al.*, 2005; Dickin and Doan, 2008).

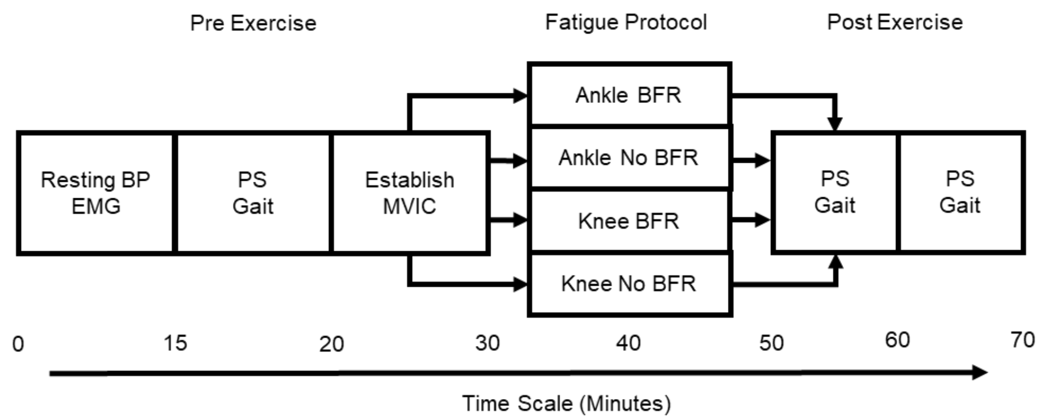

Figure 4.1: Schematic of experimental procedures. Key: BP = Blood pressure, BFR = Blood flow restriction, NBFR = No blood flow restriction, PS = Postural sway, EMG = Electromyography, MVIC = Maximal voluntary isometric contractions.

4.2.3. Blood Pressure

All participants had their resting SBP taken before each trial (0 - 10 minutes; Figure 4.1). The specific details outlining how blood pressure was taken for each participant can be viewed in Chapter 3; Section 3.9.1 – 3.9.3. The average of three resting SBP measures was taken for each participant before completing each BFR trial and used to set a target pressure (millimetres of mercury - mmHg) for BFR during either ankle or knee exercise (Table 4.1)

Table 4.1: Resting systolic blood pressure taken for each condition.

Condition	Avg. Resting SBP (mmHg)	Avg. Target SBP (mmHg)
Ankle BFR	125 ± 6; CV 4.95%	151
Ankle NBFR	125 ± 8; CV 6.33%	-
Knee BFR	124 ± 10; CV 8.16%	149
Knee NBFR	128 ± 3: CV 2.24%	-

Key: BFR = Blood flow restriction, CV = Coefficient of variance, mmHg = millimetres of mercury, NBFR = No Blood flow restriction, SBP = Systolic blood pressure. Data expression as mean ± SD, CV%.

4.2.4. Surface Electromyography

All participant's skin was shaved with a disposable razor, abraded with a scouring pad, and cleaned immediately after with an alcohol wipe. Two pre-gelled surface electrodes were applied to the vastus medialis, vastus lateralis (i.e., knee extensor trial) and the tibialis anterior and medial gastrocnemius (i.e., ankle dorsi/plantar flexor trial) musculature with an interelectrode distance of 2.5cm (Chapter 3; section 3.7.1) (Donath *et al.,* 2016). A single small electrode (EL501) was placed on the head of the fibula (ankle trial) and the surface of the patella (knee trial) as a reference electrode (Chapter 3; section 3.7.2 – Figure 3.11). The EMG system used was wireless and allowed for an easier transition when performing the fatigue protocol on the dynamometer and fitting the tourniquet around the proximal portion of the thigh. All EMG data were recorded throughout the muscle fatiguing protocol and stopped once the participants reached the 50% threshold. The specific procedure, including details of the application and calibration for the assessment of EMG, can be reviewed in Chapter 3; section 3.7.

4.2.5. Postural Sway

All participants performed DLS and SLS trials in a randomised order. Each trial lasted 30s and was repeated three times with an average of the three trials used in the subsequent analysis (a total of 6 trials pre-exercise) (Pinsault and Vuillerme, 2009; Pilot study 1). Immediately following and 10-min post-exercise, all participants completed a single DLS and SLS trial. The specific details outlining the calibration and sampling procedures for assessing postural sway can be reviewed in Chapter 3; section 3.4.

4.2.6. Gait Assessment

To assess gait each participant was instructed to walk at a comfortable pace on a motorised treadmill. The specific procedures to set up for the treadmill can be reviewed in Chapter 3; section 3.8.3.

4.2.7. Blood Flow Restriction

All participants completing the BFR fatigue protocol had a set of nylon pneumatic thigh tourniquets placed around the proximal portion of each thigh; 5cm below the femoroacetabular joint line. The specific application and protocol used to reach the target resting SBP for the blood pressure cuffs can be viewed in Chapter 3; Section 3.15 – 3.16.

4.2.8. Isokinetic Dynamometry

All participants were seated on a Biodex Rev 2 isokinetic dynamometer to perform MVIC and MVC_{Isok} (concentric/concentric) muscular contractions for the (1) ankle dorsiflexors and plantarflexors or (2) knee extensors and flexors. The specific details outlining the set-up of the dynamometer and procedures used to calibrate and adjust for gravity corrections can be viewed in Chapter 3; section 3.6.

4.2.9. Statistical Analysis

All data were analysed using SPSS 26.0 (SPSS Inc, Chicago, ILL, USA) and presented as mean ± standard deviation. The data were initially screened for outliers and checked for normality (Shapiro-Wilk Test) to confirm whether the data were normally distributed and for homogeneity of variance (Levenes test $P \geq 0.05$). A series of separate two-way repeated measures ANOVAs (time [pre, post and post 10] × condition [ankle BFR, ankle NBFR, knee BFR and knee NBFR]) were conducted on each variable. Mauchly's test of sphericity was used to determine whether sphericity was violated and when confirmed, where the epsilon (ε) was >0.75 then a Huynh-Feldt correction was used and if the ε was below <0.75

then a Greenhouse-Geisser correction was employed (Muller and Barton, 1989; Abdi, 2010). The average of three trials for COP_{AL}, COP_{ML}, COP_{VL} for each stance position (DLS and SLS_R) was determined before further analysis. For muscle activation the data were exported from Acqknowledge software to a custom-built Microsoft Excel spreadsheet to calculate the percentage from pre- and post-exercise EMG_{GM}, EMG_{TA}, EMG_{VL} and EMG_{VM} (millivolts; μV) amplitude before statistical analysis. If a significant interaction effect was detected, then follow-up simple main effects analyses using Bonferroni correction (pairwise comparisons) was performed, where no interaction effect was detected the data sets were collapsed with main effects analyses conducted. Effect sizes were calculated from the ANOVA were displayed as partial eta-squared (n_p^2) and were defined as small (<0.06), medium (0.06-0.13), and large (≥0.14), whilst Cohen's *d* was used for *post hoc* testing and were defined as negligible (<0.2), small (0.20-0.49), moderate (0.50-0.79) or large (≥0.80) (Cohen, 1988; Lakens, 2013). The alpha value was *a priori* set at P <0.05 for all tests.

4.3. Results

4.3.1. Centre of Pressure

4.3.1.1. Double Leg Stance

For COP_{AP} (Table 4.2), the two-way repeated measures ANOVA revealed no significant interaction effect ($F_{(6, 42)}$ = 0.632, P = 0.704, n_p^2 = 0.063). The main effects analyses revealed a significant effect of time ($F_{(2, 14)}$ = 4.553, P = 0.015, n_p^2 = 0.140) but not condition ($F_{(3, 21)}$ = 0.132, P = 0.940, n_p^2 = 0.019). The *post hoc* pairwise comparisons revealed a significant increase in COP_{AP} from pre- to post-exercise (collapsed data: pre: 2.69cm ± 1.14, post: 2.81cm ± 1.16, P = 0.008, d = 0.10). For COP_{ML}, no significant interaction effect ($F_{(6, 42)}$ = 0.773, P = 0.509, n_p^2 = 0.099) or main effects of time ($F_{(2, 14)}$ = 0.122, P = 0.789, n_p^2 = 0.017) or condition ($F_{(3, 21)}$ = 0.317, P = 0.617, n_p^2 = 0.043) were detected. Similarly, no significant interaction ($F_{(6, 42)}$ = 0.801, P = 0.485, n_p^2 = 0.103) or main effects of time ($F_{(2, 14)}$ = 1.397, P = 0.280, n_p^2 = 0.166) or condition ($F_{(3, 21)}$ = 2.732, P = 0.070, n_p^2 = 0.281) were detected for COP_{VL}.

Table 4.2: Mean ± SD outcome measures for double leg stance position in young adults

Condition	Pre	Post*	10-min post
Ankle BFR			
COP_{AP} (cm)	2.53 ± 0.95	2.79 ± 0.84	2.72 ± 1.09
COP_{ML} (cm)	2.35 ± 0.76	2.18 ± 0.57	2.40 ± 0.80
COP_{VL} (cm/s)	1.67 ± 0.31	1.63 ± 0.20	1.65 ± 0.34
Ankle NBFR			
COP_{AP} (cm)	2.64 ± 1.19	2.81 ± 1.37	2.51 ± 1.27
COP_{ML} (cm)	1.93 ± 0.55	2.10 ± 0.78	2.15 ± 0.88
COP_{VL} (cm/s)	1.72 ± 0.29	1.63 ± 0.44	1.52 ± 0.35
Knee BFR			
COP_{AP} (cm)	3.00 ± 1.09	2.84 ± 0.96	2.44 ± 1.19
COP_{ML} (cm)	2.51 ± 1.19	2.44 ± 0.99	2.17 ± 0.82
COP_{VL} (cm/s)	1.66 ± 0.20	1.60 ± 0.28	1.44 ± 0.27
Knee NBFR			
COP_{AP} (cm)	2.59 ± 1.44	2.83 ± 1.66	2.60 ± 1.88
COP_{ML} (cm)	2.25 ± 1.28	2.65 ± 2.53	2.60 ± 2.54
COP_{VL} (cm/s)	1.96 ± 0.59	2.23 ± 1.04	2.03 ± 1.13

Key: BFR = Blood flow restriction, NBFR = No blood flow restriction, COP_{AP} = Centre of Pressure - Anteroposterior, COP_{ML} = Centre of Pressure - Mediolateral, COP_{VL} = Centre of Pressure – Velocity. *Significant difference from pre- exercise ($P \leq 0.05$)

4.3.1.2. Single Leg Stance

For COP_{AP} (Table 4.3), there was no significant interaction ($F_{(6, 42)}$ = 0.337, P = 0.652, n_p^2 = 0.167), or main effects of time ($F_{(2, 14)}$ = 0.337, P = 0.662, n_p^2 = 0.113) or condition ($F_{3, 21)}$ = 0.704, P = 0.560, n_p^2 = 0.091). Similarly for COP_{ML}, no significant interaction ($F_{(6, 42)}$ = 1.334, P = 0.262, n_p^2 = 0.125), or main effects of time ($F_{(2, 14)}$ = 1.636, P = 0.206, n_p^2 = 0.055) or condition ($F_{(3, 21)}$ = 0.888, P = 0.464, n_p^2 = 0.113) were detected. Likewise for COP_{VL}, no significant interaction ($F_{(6, 42)}$ = 2.260, P = 0.114 n_p^2 = 0.075), or main effects of time ($F_{(2, 14)}$ = 0.893 P = 0.506, n_p^2 = 0.87) or condition ($F_{(3, 21)}$ = 1.309, P = 0.298, n_p^2 = 0.158) were detected.

Table 4.3: Mean ± SD outcome measures for single leg stance position in young adults

Condition	Pre	Post	10-min post
Ankle BFR			
COP_{AP} (cm)	4.47 ± 1.86	4.27 ± 1.40	3.85 ± 1.49
COP_{ML} (cm)	2.89 ± 0.84	2.76 ± 0.64	2.45 ± 0.10
COP_{VL} (cm/s)	4.36 ± 1.66	4.04 ± 1.44	3.73 ± 1.34
Ankle NBFR			
COP_{AP} (cm)	3.72 ± 1.01	3.39 ± 1.77	3.97 ± 1.77
COP_{ML} (cm)	2.71 ± 0.56	2.28 ± 0.84	2.68 ± 0.58
COP_{VL} (cm/s)	3.94 ± 1.10	3.24 ± 1.57	3.58 ± 1.20
Knee BFR			
COP_{AP} (cm)	3.20 ± 1.05	3.89 ± 0.86	3.37 ± 1.14
COP_{ML} (cm)	3.04 ± 0.67	2.66 ± 0.33	2.47 ± 0.86
COP_{VL} (cm/s)	4.18 ± 1.19	3.63 ± 0.84	3.42 ± 1.38
Knee NBFR			
COP_{AP} (cm)	3.58 ± 1.31	4.22 ± 2.75	4.25 ± 3.18
COP_{ML} (cm)	2.66 ± 0.37	2.98 ± 1.67	2.74 ± 1.55
COP_{VL} (cm/s)	4.08 ± 0.74	4.40 ± 2.12	4.22 ± 2.74

Key: BFR = Blood flow restriction, NBFR = No blood flow restriction, COP_{AP} = Centre of Pressure - Anteroposterior, COP_{ML} = Centre of Pressure - Mediolateral, COP_{VL} = Centre of Pressure – Velocity

4.3.2. Maximal Voluntary Isokinetic Contraction

With regards to MVC_{ISOK} performance for ankle plantarflexion, no significant interaction effect was detected ($F_{(1, 14)}$ = 0.032, *P* = 0.860, n_p^2 0.002). The main effects analyses revealed a significant effect of time ($F_{(1, 14)}$ = 2.220, *P* = <0.001, n_p^2 0.941) but not of condition ($F_{(1, 14)}$ = 0.002, *P* = 0.961, n_p^2 0.000). The *post hoc* pairwise comparisons revealed a significant decrease in isokinetic plantarflexor torque production pre- to post-exercise (collapsed data: pre = 25.68Nm ± 9.62, post = 13.88Nm ± 6.01, *P* = <0.01, *d* = 1.47). For knee extension, no significant interaction effect was detected ($F_{(1, 14)}$ = 0.166, *P* = 0.690, n_p^2 = 0.000). Main effects analyses revealed a significant effect of time ($F_{(1, 14)}$ = 17.579, *P* = <0.001, n_p^2 = 0.557) but not of condition ($F_{(1, 14)}$ = 0.033, *P* = 0.957, n_p^2 0.012). The *post hoc* pairwise comparisons revealed a significant decrease in isokinetic knee extensor torque production pre- to post-exercise (collapsed data: pre = 132.77 ± 45.88Nm, post = 96.27 ± 34.31Nm, *P* = <0.01, *d* = 0.90) (Figure 4.2).

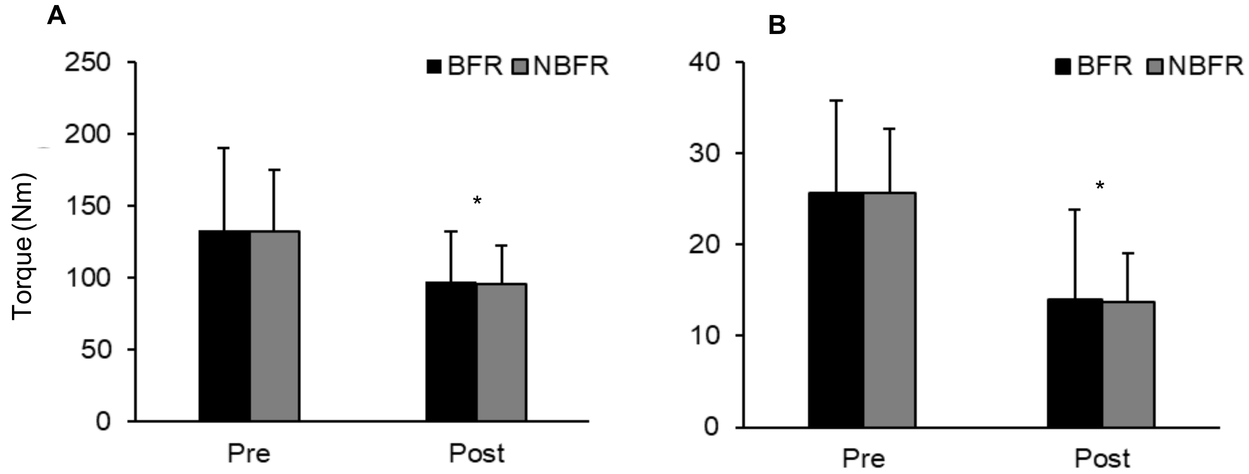

Figure 4.2: Mean ± SD for (A) knee extension and (B) ankle plantarflexion pre- and post-maximal voluntary isokinetic contraction (MVC_{isok}) exercise with blood flow restriction (BFR) and with no BFR (NBFR). *Significant difference from pre-exercise (*P* < 0.05)

4.3.3. Ankle Surface Electromyography

For EMG_{GM} (Table 4.4), no interaction effect was detected ($F_{(1, 14)}$ = 5.138, *P* = 0.445, n_p^2 = 0.042) with a main effect of time ($F_{(1, 14)}$ = 5.158, *P* = 0.039, n_p^2 = 0.269) but not for condition ($F_{(1, 14)}$ = 2.310, *P* = 0.151, n_p^2 = 0.142). The *post hoc* pairwise comparisons revealed a significant decrease in EMG_{GM} pre- to post-exercise (collapsed data: pre = 95.96 ± 8.01%, post = 86.25 ± 10.20%, *P* = 0.036, *d* = 1.06). Similarly, for EMG_{TA} no interaction effect was detected ($F_{(1, 14)}$ = 2.670, *P* = 0.125, n_p^2 = 0.160). The main effect analyses revealed a significant effect for time ($F_{(1,14)}$ = 31.326, *P* = <0.001, n_p^2 = 0.691) but not for condition ($F_{(1, 14)}$ = 0.048, *P* = 0.990, n_p^2 = 0.004). The *post hoc* pairwise comparisons revealed a significant decrease in EMG_{TA} pre- to post-exercise (collapsed data: pre = 89.92 ± 13.10%, post = 69.04 ± 15.46%, *P* = <0.01, *d* = 1.46).

Table 4.4: Mean ± SD in Electromyography (EMG) amplitude pre- and post-ankle fatiguing exercise with blood flow restriction (BFR) and no BFR (NBFR)

%	Ankle Pre BFR	Ankle Post BFR	Ankle Pre NBFR	Ankle Post NBFR
EMG_{GM}*	95.56 ± 5.82	90.58 ± 9.28	96.35 ± 10.31	81.93 ± 10.05
EMG_{TA}*	92.35 ± 8.75	66.30 ± 15.86	87.37 ± 16.21	71.79 ± 13.11

Key: GM = Gastrocnemius, TA = Tibialis Anterior. *Significant difference from pre-exercise (*P* ≤0.05)

4.3.4. Knee Surface Electromyography

For EMG_{VL} (Table 4.5), no significant interaction effect was detected ($F_{(1, 14)}$ = 0.440, *P* = 0.518, n_p^2 = 0.030). The main effect analyses revealed a significant effect of time ($F_{(1, 14)}$ = 9.610, *P* = 0.008, n_p^2 = 0.407) but not for condition ($F_{(1, 14)}$ = 1.125, *P* = 0.971, n_p^2 = 0.001). The *post hoc* pairwise comparisons revealed a decrease in EMG_{VL} pre- to post-exercise (collapsed data: pre = 94.19% ± 5.80, post = 83.38% ± 14.46, *P* = 0.006, *d* = 0.76) (Table 4.5). For EMG_{VM}, no significant interaction ($F_{(1, 14)}$ = 0.520, *P* = 0.822, n_p^2 = 0.004) or main effects of time ($F_{(1, 14)}$ = 0.650, *P* = 0.803, n_p^2 = 0.005) or condition ($F_{(1, 14)}$ = 1.531, *P* = 0.928, n_p^2 = 0.001) were detected.

Table 4.5: Mean ± SD in Electromyography (EMG) amplitude pre- and post-knee fatiguing exercise with blood flow restriction (BFR) and no BFR (NBFR)

%	Knee Pre BFR	Knee Post BFR	Knee Pre NBFR	Knee Post NBFR
EMG_{VM}	84.10 ± 28.39	84.18 ± 26.65	86.00 ± 14.34	83.22 ± 19.04
EMG_{VL}*	93.61 ± 5.97	80.49 ± 15.50	95.00 ± 5.91	86.17 ± 13.80

Key: VL= Vastus Lateralis, VM = Vastus Medialis. *Significant difference from pre-exercise (*P* ≤ 0.05)

4.3.5. Gait Stability

For stride time, there was no interaction effect ($F_{(6, 42)}$ = 2.302, P = 0.061, n_p^2 = 0.231). The main effect analyses revealed a significant effect of time ($F_{(2, 14)}$ = 4.826, P = 0.017, n_p^2 = 0.173) but not of condition ($F_{(3, 21)}$ = 3.241, P = 0.062, n_p^2 = 0.297). The *post hoc* pairwise comparisons revealed an increase in stride time pre- to post-exercise (collapsed data: pre = 1.15 ± 0.06s, post = 1.20 ± 0.08s, P = 0.040, d = 0.57). For stride length, no significant interaction effect ($F_{(6, 42)}$ = 5.738, P = 0.376, n_p^2 = 0.121) nor main effects of time ($F_{(2, 14)}$ = 0.608, P = 0.542, n_p^2 = 0.025) or condition ($F_{(3, 21)}$ = 2.024, P = 0.137, n_p^2 = 0.202) were detected. For double limb support, no significant interaction ($F_{(6, 42)}$ = 1.180, P = 0.385, n_p^2 = 0.104) nor main effects of time ($F_{(2, 14)}$ = 1.148, P = 0.325, n_p^2 = 0.039) or condition ($F_{(3, 21)}$ = 3.051, P = 0.058, n_p^2 = 0.276) were detected (Table 4.6).

Table 4.6: Mean ± SD outcome measures for gait stability in young adults

Condition	Pre	Post*	10-min post
Ankle BFR			
Stride Time (s)	1.15 ± 0.08	1.16 ± 0.10	1.14 ± 0.08
Stride Length (cm)	101.62 ± 5.00	102.72 ± 13.66	105.20 ± 13.32
Double Support (s)	0.27 ± 0.24	0.31 ± 0.27	0.24 ± 0.19
Ankle NBFR			
Stride Time (s)	1.17 ± 0.08	1.29 ± 0.14	1.20 ± 0.10
Stride Length (cm)	104.88 ± 13.92	109.47 ± 17.98	101.67 ± 24.72
Double Support (s)	0.47 ± 0.21	0.53 ± 0.22	0.38 ± 0.28
Knee BFR			
Stride Time (s)	1.14 ± 0.05	1.14 ± 0.08	1.16 ± 0.08
Stride Length (cm)	105.48 ± 12.90	103.37 ± 16.71	108.2 ± 15.65
Double Support (s)	0.32 ± 0.28	0.26 ± 0.14	0.29 ± 0.15
Knee NBFR			
Stride Time (s)	1.14 ± 0.05	1.19 ± 0.09	1.14 ± 0.06
Stride Length (cm)	87.07 ± 8.37	89.93 ± 16.78	89.61 ± 13.48
Double Support (s)	0.24 ± 0.17	0.31 ± 0.15	0.32 ± 0.21

Key: BFR = Blood flow restriction, NBFR = No Blood flow restriction. *Significant difference from pre-exercise ($P \leq 0.05$)

4.4. Discussion

The aim of this study was to explore the effects of lower limb muscle fatigue with and with no BFR on postural sway and gait characteristics in young adults. To our knowledge, this is the first study which addresses this question. The main findings from this study were as follows: (1) the muscle fatiguing protocols were successful at significantly reducing muscle torque production, confirming the presence of muscle fatigue; (2) postural sway was largely unchanged, but an increase in COP_{AP} was detected post-exercise (independent of condition, i.e. data were collapsed across conditions) during DLS trials only, and (3) gait stability (stride time) increased from pre- to post-exercise, but this was independent on condition. Overall, the findings of this study show that whilst markers of muscle function were impaired following fatigue protocols, the overall impact of fatigue on postural sway and gait characteristics were minimal. Whilst the results suggest a lack of significant evidence for the effects of fatigue and/or blood flow restriction on postural control, this does not, however, preclude the possibility of a true effect, as the low power of the analyses may have limited our ability to detect a small but meaningful difference. Therefore, a study with sufficient power is needed to investigate this issue further.

4.4.1. Characterisation of Muscle Fatigue

The exercise protocol successfully induced muscle fatigue, as evidenced by the decrease in ankle and knee MVC_{ISOK} torque post-exercise. These findings are consistent with previous protocols where a reduction in MVC_{ISOK} was observed post-exercise in young adults (Gribble and Hertel, 2004; Lindström *et al*., 2006; Dickin and Doan, 2008). Although, it is important to highlight that the decrease in MVC_{ISOK} torque (pre- to post-exercise) reported was relatively large for the ankle (~45%) and moderate for knee (~28%), some participants did not reach the 50% threshold in MVC before reassessing postural control. In general, the less than anticipated reduction in MVC_{ISOK} torque for the knee could be related to the contraction type (concentric – concentric), the velocity of movement (60°/s) and the duration of the exercise (number of repetitions performed). It is possible that the exercise intensity for some participants may have been too challenging (e.g., co-ordination when performing the movement on the dynamometer) when using a slower velocity (60°/s) compared to a faster velocity (e.g., 150°/s), as more effort is required to maintain repeated muscle contractures during slower velocities. Several review articles have suggested that younger adults are less resistant to fatigue and is dependent on the type of contraction, velocity of movement and whether the fatiguing protocol was performed using submaximal or maximal efforts (Paillard, 2012; Berchicci *et al.*, 2013). In addition to the submaximal contractions performed at a slower velocity, there may have been a different characteristic of neuromuscular fatigue (e.g., peripheral and central fatigue) experienced by the young adults in the present study. Gandevia *et al.*, (1998) and Babault *et al.*, (2006) refer to slower angular velocities of 30 - 60°/s (respectively) and repeated efforts are aligned to central fatigue. Thus, the level of fatigue experienced in the study may have been centrally driven where discomfort and emotive responses (e.g., level of motivation) could have been a factor. Further investigation on maximal and submaximal efforts of muscle contracture along with monitoring verbal responses (e.g., rate of perceived exertion) with and without BFR should be considered in the future.

In the current study there were marked reductions in EMG_{TA} (~25%), EMG_{GM} (~14%) following fatigue to the ankle, which is contrary to previous findings on EMG_{TA}, where an increase in amplitude after muscle fatiguing exercise has been reported (Næss-Schmidt *et al.*, 2017; Kjelsden *et al.*, 2019). Both Næss-Schmidt *et al.* (2017) and Kjeldsen *et al*. (2019) identified an increase in EMG_{TA} amplitude after exercise where the tibialis anterior muscle returned to baseline levels 10-min post-exercise. Similarly, a reduction in EMG_{VL} (~12%) was observed after the knee fatiguing protocols in the current study, where previous literature has reported an increase in EMG_{VL} following knee fatiguing exercise (Karabulut and Perez, 2013; Fatela *et al.*, 2016). A possible explanation for the reduction in muscle activation might be the presence of either central (e.g., reduced descending neural drive from the motor cortex) or peripheral (reduced acetylcholine uptake at the motor end plate on the sarcolemma) neuromuscular fatigue. Based on the evidence available on peripheral muscle fatigue, the fatigue sensitivity type III/IV muscle afferents (sensitive to metabolite production) may have inhibited the neural feedback to the central nervous system preventing the muscle to recover (Allman and Rice, 2002; Carrol *et al.*, 2017). Consequently, the central processes responsible for maintaining neural discharge for voluntary muscle activity may have been affected causing a reduction in force and EMG amplitude inducing a loss of muscle fibre activation (Boyas and Guèvel, 2011). Although, the present study did not use methods to assess central and peripheral muscle fatigue (e.g., peripheral nerve stimulation) and can only speculate their influence on the reduction in force and EMG amplitude. A consideration for the future would be to examine central and peripheral muscle fatigue with or without BFR. Furthermore, these findings suggest that the fatigue experienced at the ankle and knee musculature did not increase muscle activation (e.g., fast twitch muscle fibres) as previously described in the BFR literature. Notably, there was no difference in fatigue with or without BFR, and the fatigue experienced was a result of the exercise performed and not BFR. Despite the current study not continuously monitoring EMG amplitude post-exercise and up to the 10-minutes recovery period, what has been established is the fatiguing protocols did reduce EMG amplitude and MVC_{ISOK} post-exercise. Future consideration should be directed towards observation on different muscle contracture and mode of exercise with and without BFR.

4.4.2. Postural Sway

Contrary to the initial hypothesis and the existing literature (Pline *et al.*, 2006; Dickin and Doan, 2008; Bisson *et al.*, 2012; Warnica *et al.*, 2014), muscle fatigue had minimal effects on postural sway during easy (DLS) and challenging (SLS) balance tasks. However, consistent with previous literature (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Lin *et al.*, 2009), there was a significant increase in COP_{AP} following ankle and knee fatigue during DLS. The increase in COP_{AP} sway following muscle fatigue during DLS is not surprising as sagittal plane movement during bipedal stance is controlled primarily by the tibialis anterior and gastrocnemius muscles which were targeted in the current study (Donath *et al.*, 2016). However, the musculature responsible for maintaining frontal plane movement (e.g., peroneal brevis, longus or tibialis posterior), was not fatigued nor was muscle activation measured during balance tasks and can therefore can only speculate as to why the increase in COP_{AP} was a result of tibialis anterior and gastrocnemius muscle fatigue.

The increase in COP_{AP} could be attributed to a change in neuromuscular function through peripheral fatigue (e.g., motor neuron discharge rate, an increase in metabolite production, stretch reflex; Yaggie and McGregor, 2002; Vuillreme *et al.*, 2008; Paillard, 2012). It has been established that the ankle musculature is critical for minimising body sway during quiet standing (Yaggie and McGregor, 2002; Gribble and Hertel, 2004). In this context, during muscle fatigue the response of the muscle is to increase the firing rate (Paillard,

2012) but as fewer motor units become available, there is a notable change in motor neuron discharge rates which can affect intrafusal and extrafusal muscle fibres (Pline *et al*., 2006) resulting in a change to the stretch reflex of the musculature (Warnica *et al*., 2014). Such a change in the stretch reflex may have resulted in an increase in COP_{AP}, which may have been accelerated by the fatigue protocols. Previous studies comparing muscle fatiguing trials have reported a 'quicker' time (or fewer repetitions) to muscle failure due to an increase in metabolite production (e.g., potassium and sodium ion) (Broxterman *et al*., 2015; Kjeldson *et al*., 2019). Therefore, the exercise protocol performed may be capable of slowing the rate lactate diffusion from the fatiguing muscle creating a more pronounced acidic environment, affecting the speed of muscle contraction and torque produced (Loenneke *et al*., 2012b).

Despite changes in muscle function (e.g., reduction in MVC and EMG), the 50% threshold set for the fatigue protocols may have contributed to the mostly null effects of muscle fatigue on postural sway outcomes. The assumption here is the tourniquet pressure (~1.2 x rSBP) set for BFR was too low and may not have been enough to cause the level of peripheral fatigue anticipated to disturb postural control. It is thought that low tourniquet inflation pressure is not as effective at reducing the level of muscle oxygen saturation compared to high tourniquet inflation pressure (≥1.4 x rSBP) when performing low intensity BFR exercise (Dankel *et al*., 2017; Ilet *et al*., 2019). In addition, there was no difference between BFR and NBFR, which may also support the possibility that the pressure applied was too low. Therefore, the 50% fatigue threshold combined with the low cuff pressure of the tourniquet may not have been enough to cause muscle fatigue to the target musculature. However, increasing the cuff pressure increases the risk of increased discomfort (Yasuda *et al*., 2014; Scott *et al*., 2018), perceived exertion (Wernbom *et al*., 2009) and exercise intolerance. In light of this, it was important to prescribe a dosage where participants were able to withstand BFR pressure from the tourniquet at a 'low level' first before attempting in older adults. Future studies should consider a higher dosage of tourniquet inflation pressure, as the pressure set for the current study may have been too low to elicit a perturbation to balance performance.

Another tenable explanation for the minimal effects of muscle fatigue and BFR on postural sway outcomes may be that the young healthy cohort were well equipped to compensate for the fatiguing effects on fine motor control. It has been reported that young adults can recover quickly following muscle fatiguing exercise using isokinetic protocols (Celes *et al*., 2010; Boyas *et al*., 2013). In addition, the choice of contraction (concentric - concentric) and mode of exercise (isokinetic) performed during the fatiguing trials may not have been demanding enough to reduce torque production to disturb postural sway. Thus, opting for a more challenging eccentric contraction may provide a greater stimulus to reduce torque production compared to concentric alone. However, this was an exploratory study where the effects of postural control after performing BFR is relatively unknown. It was important to consider a safe mode and contraction type similar to the activities for daily living (Lindström *et al*., 2006), but also to select a fatiguing protocol commonly used to disturb postural control in a controlled environment (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Dickin and Doan, 2008; Vuillerme and Boisgontier, 2010; Bisson *et al*., 2012; Boyas *et al*., 2019). The lack of evidence of effect should be interpreted cautiously, as the low sample size resulted in the study being underpowered. Therefore, future research should use a larger sample size to either confirm or refute these findings. Furthermore, future research should also focus on identifying a muscle contraction that challenges postural control under BFR conditions and to observe the recovery in postural control following this performance.

4.4.3. Gait Stability

Spatiotemporal gait measures obtained during walking are often used to assess the efficiency of walking and to evaluate fall risk (Fuchioka *et al*., 2015). The results from this study identified a significant increase in stride time post-exercise, returning to baseline levels 10-min after the cessation of exercise. As stride time is regarded as a rhythmic stepping mechanism (Beauchet *et al.,* 2005) with a longer stride time associated with a decline in motor control required to maintain dynamic movement, this observation is interpreted as a disturbance to dynamic postural control. An increase in stride time tends to increase with age (Kang and Dingwell, 2008) as more attentional demands are required in older adults to maintain postural control during locomotion in comparison to young adults (Herssens *et al.,* 2018). However, the largely null effects of muscle fatigue and/or presence of BFR on gait outcomes (double limb support; stride length) may be a result of the rapid recovery of muscle fatigue following the exercise protocol or an insufficient BFR pressure, respectively. In the present study, the null effects of fatigue/ BFR could be further explained by the fact that gait trials were always performed after posturographic trials (~3mins post-exercise). Thus, the possibility that fatigue effects on gait performance were missed due to experimental design cannot be ruled out. The absence of evidence for an effect should be interpreted cautiously, as the low sample size led to the study being underpowered. Future research should utilize a larger sample size to confirm or refute these findings. In addition, future studies examining gait stability *immediately* after muscle fatiguing exercises with BFR or without BFR are required to ascertain whether dynamic balance is altered after fatiguing contractions.

4.4.4. Limitations

There were several limitations that require acknowledgment. Firstly, EMG activity was not measured during quiet standing and gait trials. This would have provided an understanding of muscle co-activation patterns and would have allowed a better understanding of the (predominantly) null effects of the fatigue protocols on the outcome measures. Second, only musculature responsible for maintaining sagittal plane movements were fatigued. Future research on fatiguing the invertors and evertor muscles of the frontal plane, which are paramount for hip and trunk stability and controlling postural sway in the COP_{ML} direction during unipedal stance position, would provide an insight on the immediate effects of BFR on other musculature important for postural control. Third, the fatigue protocol was performed using a controlled laboratory protocol and did not reflect 'normal' everyday movements (e.g., sitting and rising from a chair – mimicking a squat movement pattern), which limits the ecological validity of our quiet standing and gait measurements. Finally, and perhaps most restricting, the sample size was limited and included only young adults, which precludes us from generalising our findings to different age groups. However, the inclusion of only young adults in this exploratory study was intentional to understand the acute effects of BFR exercise on postural control and gait stability. Such observations provide a foundation to extend these findings to older adults (Chapter 5).

4.4.5. Summary

This is the first study to investigate the effects of ankle and knee muscle fatigue *with* and *without BFR* on static and dynamic postural control. Contrary to previous research, few changes in postural sway metrics following ankle and knee muscle fatigue were reported. Novel to the present study, BFR conferred no impact (either beneficial or detrimental) on fatigue induced changes in postural sway and gait stability. Based on these observations, coupled with research that shows that BFR combined with exercise is as effective than exercise alone for several important strength outcomes, the use of BFR might be safe from a postural control and fall risk perspective. Therefore, BFR could be considered as a suitable training intervention in young adults, which does not compromise postural control or gait despite the reduction in muscle torque performance. These findings may have relevance for rehabilitation specialists/ practitioners looking for an alternative form of exercise to promote short term resistance training programme to reduce muscle performance (e.g., increase strength) but does not compromise postural control. The absence of statistically significant findings in the present study should not be interpreted as evidence for the absence of an effect. Instead, this may reflect insufficient statistical power, increasing the possibility of a Type II error. Future research should replicate the current study in older adults. Such information is vital as older adults are more susceptible to an increased fall risk.

Chapter 5:

The Effects of Lower Limb Muscle Fatigue with and without Blood Flow Restriction on Postural Sway and Gait Characteristics in Healthy Older Adults

Highlights:

- This study compared the effects of acute ankle and knee muscle fatigue with blood flow restriction (BFR) and with no blood flow restriction (NBFR) on postural sway and gait stability in older adults.
- Ankle and knee muscle fatigue reduced muscle torque and muscle activity.
- Postural sway amplitude (COP_{AP}) improved post-exercise after ankle and knee NBFR fatiguing exercise for the single leg stance position only, with minimal changes to postural sway amplitude and velocity post-exercise when performing double leg stance position.
- There was no change in gait stability measures under BFR and NBFR conditions.

5.1. Introduction

Postural control is maintained through processing both sensory and motor information received from the central and peripheral nervous systems to limit the risk of falls during quiet standing (Assländer and Perteka, 2014; Boyas *et al*., 2019), with the vestibular, visual, and proprioceptive systems interacting constantly to ensure posture control is maintained before performing the next motor skill (e.g., quiet standing to walk) (Silder *et al*., 2008; Boyas, *et al*., 2019). Ageing and the associated loss of muscle strength or acute reduction in force generating capacity (i.e. muscle fatigue), can increase the risk of falls (Lanza *et al*., 2004; Moore *et al*., 2005; Christie *et al*., 2011). A comprehensive meta-analysis by Avin and Law (2011) revealed age-related differences in acute muscle fatigue with greater depreciation in muscle strength and torque with ageing. Moreover, there is compelling evidence to suggest that general and localised muscle fatigue is greater in older adults, which can disturb postural control through various mechanisms (Paillard, 2012). This is partly due to a reduction in myotendinous efficiency, conversion to a higher composition of slow twitch muscle fibres (Evans and Lexell, 1995), an increase in fatigue resistance (Lanza *et al*., 2004), and the adoption of compensatory strategies to prevent the loss of postural control, but regardless of the mechanisms, the increase in muscle fatigue with ageing has important implications for the risk of falls and the need to develop strategies to combat this fatigue.

Understanding the relationship between muscle fatigue and postural control in older adults is of significant practical value, since postural control and muscle fatigue are known fall risk factors (Fabre *et al*., 2010). In normal ageing, older adults present with a reduction in strength however there is an increase in latency, i.e. where muscle recruitment is delayed so delaying the production of torque at specific joints (e.g., ankle or hip), which can compromise postural control if muscle fatigue or an unexpected perturbation occurs (Kanekar and Aurin, 2014: Donath *et al*., 2016). Therefore, the organisation of the central nervous system and the information processed from the peripheral nervous system (e.g., proprioception) is paramount for controlling posture during quiet standing and initiating dynamic movement (e.g., gait) in older adults (Tucker, *et al*., 2009; Paillard, 2012).

Whilst exercise can improve long-term muscle strength in older adults (Manini and Clark, 2010), intense or prolonged exercise can acutely affect proprioception through alterations to excitation-coupling processes or reduction in motor unit discharge (Bisson *et al*., 2012) and can lead to an increase postural sway via greater movement of the centre of pressure (COP). Several studies have reported an increase in COP amplitude and velocity after ankle or knee muscle fatiguing exercise in older adults (Gribble and Hertel, 2004; Yaggie and McGregor, 2004; Papa *et al*., 2015). In comparison to young adults that adopt an ankle strategy (i.e. the ankle musculature is largely responsible for maintaining static balance), older individuals adopt a 'hip' strategy to maintain balance and consequently exhibit an increase in mediolateral COP amplitude after muscle fatiguing trials (Meltzer *et al*., 2004; Hausdorff, 2007; Hilliard *et al*., 2008 Nam *et al*., 2013). Therefore, it is imperative to explore ways in which older adults can exercise safely, without incurring acute negative side effects from resistance exercise that may acutely increase fall risk.

One potential approach is to improve muscle strength whilst minimising acute fatigue is low intensity exercise with BFR. However, research scrutinising the effects of muscle fatigue in combination with other training modalities, such as low-intensity resistance exercise with blood flow restriction, is scarce. In Chapter 4, it was reported that postural control (e.g., quiet standing and gait stability) was largely unaffected after ankle and knee muscle fatigue with and without BFR in young adults. Given that ageing is associated with a progressive decline in postural control (Sturnieks *et al*., 2018), which is driven, in part, by a loss of leg proprioception (Anson *et al.,* 2017; Henry and Baudry, 2019) it is logical to explore whether the effects of muscle fatigue with BFR may be more pronounced in older adults. Therefore, the aims of this study were to investigate the acute effects of muscle fatigue to either the ankle (dorsiflexors and plantarflexors muscles) or knee (extensors and flexors muscles) with BFR or with NBFR on postural sway amplitude and velocity when performing quiet standing and dynamic balance tasks in older adults. This work will build on the knowledge gleaned from the previous work (Chapter 4) in younger adults.

Research Hypothesis (H_1): Muscle fatigue performed using repeated maximal voluntary isokinetic contraction protocol will elicit a marked reduction in balance (increase in postural sway), gait performance (longer time during the double limb support stance, increase in stride time and shorter stride length) and muscle performance (reduction in torque and increase in EMG amplitude) at the ankle and knee following the fatiguing protocol, with these effects more pronounced in the BFR condition.

5.2. Methods

5.2.1. Participants

Eight recreationally active male (n = 4) and female (n = 4) older adults (age: 64 ± 2.45 years, height: 168.45 ± 12.31cm, weight: 51.25 ± 13.13kg) volunteered to participate in the study. The inclusion and exclusion criteria used for this study can be viewed in Chapter 3; section 3.3.1 – 3.3.2. A sample size calculation using *a priori* power (G*Power version 3.1.9.2, Universitat Kiel, Dusseldorf, Germany; Faul *et al*., 2007) for repeated measures identified a minimum of 28 participants would be required to detect a medium standardised effect (Cohen's f = .25 with α 0.05 and a 1-β error of 0.80) for one group and two within subject measures (Appendix 9.3.1). This study is exploratory and aims to establish a proof of concept. A follow up *post hoc* power calculation was performed based on the 8 participants recruited in the current study (medium standardised effect of Cohen's f = .25 with α 0.05 and a 1-β error of 0.26 [Appendix 9.3.1]).

5.2.2. Experimental Procedures

On the initial visit, all participants completed the necessary informed consent paperwork (Chapter 3; section 3.3) followed by the collection of anthropometric data collected using the procedures identified in Chapter 3; sections 3.9.1, 3.9.3 and 3.17. Thereafter, a double bipolar surface EMG electrode was placed over the muscle belly of the tibialis anterior (EMG_{TA}), medial head of the gastrocnemius (EMG_{GM}) for the ankle fatiguing protocol or the vastus medialis (EMG_{VM}) and vastus lateralis (EMG_{VL}) for the knee fatiguing protocol on the participant's right leg (Chapter 3; section 3.7). A single EMG electrode was placed over two bony prominences for the knee (patella) and ankle (head of the fibula). Quiet standing balance using a double leg stance (DLS; eye closed) and single leg stance (SLS; eyes open - right leg) was measured on a force platform to assess postural sway amplitude (PS), followed by gait assessments (walking gait [timed gait and ground reaction forces]). The outcome measures were assessed before and immediately after fatigue protocols and following a 10-minute passive recovery. For the BFR conditions, two nylon thigh tourniquets were applied to the upper portion of the right and left thigh whilst participants (Chapter 3; section 3.15 – 3.16). The tourniquets remained deflated until the participant was seated in the chair of the dynamometer and ready to perform the fatiguing trial.

The fatigue protocol consisted of repeated unilateral (right leg) maximal voluntary isokinetic contractions (MVC_{ISOK}; concentric/concentric) performed between 5 – 15° for ankle dorsiflexion and plantarflexion and at 90 – 120° for knee extension and flexion at a velocity of 60°/s (Gribble and Hertel, 2004). This fatigue protocol remained the same for the older adult given this population has poorer postural control and may be more sensitive to the fatiguing exercise. The detailed procedure for the ankle and knee exercise protocols can be found in Chapter 3; section 3.6. An initial submaximal warm-up was performed on the dynamometer to familiarise the participant with the protocol and to establish an initial maximal voluntary isometric contraction (MVIC) (3 repetitions x 3-s holds – 1 min rest between). The average of the three MVIC scores was taken, and 50% of the mean MVIC was calculated (Dickins and Doan, 2008) and used to establish baseline maximal torque production before the fatiguing trial and to normalise the EMG amplitude. For the BFR trials, the tourniquets were inflated to 1.2 x resting systolic blood pressure (SBP) (Chapter 3; section 3.15 - 3.16) before starting the fatigue trial. For each fatigue trial, all participants completed ankle (dorsiflexion/plantarflexion) or knee (flexion/extension) contractions until the MVC_{ISOK} had decreased to 50% of the baseline value. This is a standard protocol used to induce fatigue in postural control studies (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Harkins *et al*., 2005; Dickin and Doan, 2008).

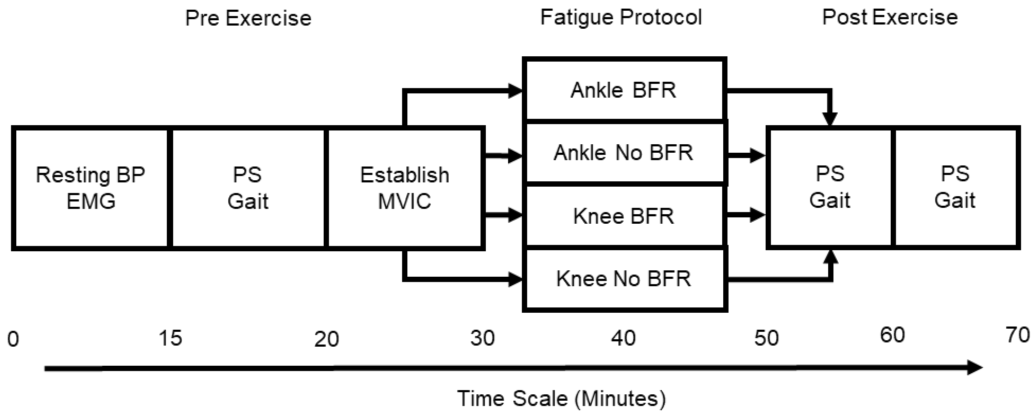

Figure 5.1: Schematic of experimental procedures. Key: BP = Blood pressure, BFR = Blood flow restriction, NBFR = No blood flow restriction, PS = Postural sway, EMG = Electromyography, MVIC = Maximal voluntary isometric contractions.

5.2.3. Blood Pressure

All participants had their resting SBP taken before each trial (0 - 10 minutes; Figure 4.1). The specific details outlining how blood pressure was taken for each participant can be viewed in Chapter 3; Section 3.9.1 – 3.9.3. The average of three resting SBP measures was taken for each participant before completing each BFR trial and used to set a target pressure (millimetres of mercury - mmHg) for BFR during either ankle or knee exercise (Table 5.1).

Table 5.1: Resting systolic blood pressure taken for each condition.

Condition	Avg. Resting SBP (mmHg)	Avg. Target SBP (mmHg)
Ankle BFR	127 ± 13; CV 10.61%	152
Ankle NBFR	129 ± 15; CV 11.76%	-
Knee BFR	130 ± 15; CV 11.83%	156
Knee NBFR	129 ± 12; CV 10.00%	-

Key: BFR = Blood flow restriction, CV = Coefficient of variance, mmHg = millimetres of mercury, NBFR = No Blood flow restriction, SBP = systolic blood pressure. Data expression as mean ± SD, CV%.

5.2.4. Surface Electromyography

All participant's skin was shaved with a disposable razor, abraded with a scouring pad, and cleaned immediately after with an alcohol wipe. Two pre-gelled surface electrodes were applied to the vastus medialis, vastus lateralis (i.e., knee extensor trial) and the tibialis anterior and medial gastrocnemius (i.e., ankle dorsi/plantar flexor trial) musculature with an interelectrode distance of 2.5cm (Chapter 3; section 3.7.1) (Donath *et al.,* 2016). A single small electrode (EL501) was placed on the head of the fibula (ankle trial) and the surface of the patella (knee trial) as a reference electrode (Chapter 3; section 3.7.2 – Figure 3.11). The EMG system used was wireless and allowed for an easier transition when performing the fatigue protocol on the dynamometer and fitting the tourniquet around the proximal portion of the thigh. All EMG data were recorded throughout the muscle fatiguing protocol and stopped once the participants reached the 50% threshold. The specific procedure, including details of the application and calibration for the assessment of EMG, can be reviewed in Chapter 3; section 3.7.

5.2.5. Postural Sway

All participants performed DLS and SLS trials in a randomised order. Each trial lasted 30 s and was repeated three times with an average of the three trials used in the subsequent analysis (a total of 6 trials pre-exercise) (Pinsault and Vuillerme, 2009; Pilot study 1). Immediately following and 10-min post-exercise, all participants completed a single DLS and SLS trial. The specific details outlining the calibration and sampling procedures for assessing postural sway can be reviewed in Chapter 3; section 3.4.

5.2.6. Gait Assessment

To assess gait each participant was instructed to walk at a comfortable pace on a motorised treadmill. The specific procedures to set up for the treadmill can be reviewed in Chapter 3; section 3.8.3.

5.2.7. Blood Flow Restriction

All participants completing the BFR fatigue protocol had a set of nylon pneumatic thigh tourniquets placed around the proximal portion of each thigh; 5cm below the femoroacetabular joint line. The specific application and protocol used to reach the target resting SBP for the blood pressure cuffs can be viewed in Chapter 3; Section 3.15 – 3.16.

5.2.8. Isokinetic Dynamometry

All participants were seated on a Biodex Rev 2 isokinetic dynamometer to perform MVIC and MVC_{Isok} (concentric/concentric) muscular contractions for the (1) ankle dorsiflexors and plantarflexors or (2) knee extensors and flexors. The specific details outlining the set-up of the dynamometer and procedures used to calibrate and adjust for gravity corrections can be viewed in Chapter 3; section 3.6.

5.2.9. Statistical Analysis

All data were analysed using SPSS 26.0 (SPSS Inc, Chicago, ILL, USA) and presented as mean ± standard deviation. The data were initially screened for outliers and checked for normality (Shapiro-Wilk Test) to confirm whether the data were normally distributed and for homogeneity of variance (Levenes test $P \geq 0.05$). A series of separate two-way repeated measures ANOVAs (time [pre, post and post 10] × condition [ankle BFR, ankle NBFR, knee BFR and knee NBFR]) were conducted on each variable. Mauchly's test of sphericity was used to determine whether sphericity was violated and when confirmed, where the epsilon (ε) was >0.75 then a Huynh-Feldt correction was used and if the ε was below <0.75

then a Greenhouse-Geisser correction was employed (Muller and Barton, 1989; Abdi, 2010). The average of three trials for $COP_{AL,}$ COP_{ML}, COP_{VL} for each stance position (DLS and SLS_{R}) was determined before further analysis. For muscle activation the data were exported from Acqknowledge software to a custom-built Microsoft Excel spreadsheet to calculate the percentage from pre- and post-exercise EMG_{GM}, EMG_{TA}, EMG_{VL} and EMG_{VM} (millivolts; μV) amplitude before statistical analysis. If a significant interaction effect was detected, then follow-up simple main effects analyses using Bonferroni correction (pairwise comparisons) was performed, where no interaction effect was detected the data sets were collapsed with main effects analyses conducted. Effect sizes were calculated from the ANOVA were displayed as partial eta-squared (n_p^2) and were defined as small (<0.06), medium (0.06-0.13), and large (≥0.14), whilst Cohen's *d* was used for *post hoc* testing and were defined as negligible (<0.2), small (0.20-0.49), moderate (0.50-0.79) or large (≥0.80) (Cohen, 1988; Lakens, 2013). The alpha value was *a priori* set at $P < 0.05$ for all tests.

5.3. Results

5.3.1. Centre of Pressure

5.3.1.1. Double Leg Stance

For COP_{AP} (Table 5.2), the two-way repeated measures ANOVA revealed no significant interaction effect ($F_{(6, 42)}$ = 1.440, P = 0.232, n_p^2 = 0.224), or main effects of time ($F_{(2, 14)}$ = 0.231, P = 0.798, n_p^2 = 0.044) or condition ($F_{(3, 21)}$ = 1.677, P = 0.215, n_p^2 = 0.251). Similarly for COP_{ML}, no significant interaction effect ($F_{(6, 42)}$ 2.462, P = 0.099, n_p^2 = 0.291), or main effects of time ($F_{(2, 14)}$ = 0.260, P = 0.974, n_p^2 = 0.004) or condition ($F_{(3, 21)}$ = 1.054, P = 0.368, n_p^2 = 0.149) were detected. Again for COP_{VL}, no significant interaction effect ($F_{(6, 42)}$ = 1.043, P = 0.401, n_p^2 = 0.148), or main effects of time ($F_{(2, 14)}$ = 0.837, P = 0.429, n_p^2 = 0.122), or condition ($F_{(3, 21)}$ = 3.962, P = 0.082, n_p^2 = 0.898) were detected.

Table 5.2: Mean ± SD outcome measures for double leg stance position in older adults.

Condition	Pre	Post	10-min post
Ankle BFR			
COP_{AP} (cm)	2.11 ± 0.13	2.20 ± 0.36	1.99 ± 0.27
COP_{ML} (cm)	2.00 ± 0.12	2.22 ± 0.29	1.99 ± 0.25
COP_{VL} (cm/s)	2.47 ± 0.13	2.30 ± 0.36	2.18 ± 0.35
Ankle NBFR			
COP_{AP} (cm)	2.77 ± 0.50	2.69 ± 0.39	3.21 ± 0.13
COP_{ML} (cm)	2.32 ± 0.18	2.27 ± 0.07	2.82 ± 0.09
COP_{VL} (cm/s)	2.08 ± 0.39	2.03 ± 0.44	2.27 ± 0.13
Knee BFR			
COP_{AP} (cm)	2.20 ± 0.20	2.41 ± 0.27	2.00 ± 0.21
COP_{ML} (cm)	1.95 ± 0.27	2.17 ± 0.24	1.51 ± 0.02
COP_{VL} (cm/s)	2.10 ± 0.11	1.86 ± 0.04	1.35 ± 0.06
Knee NBFR			
COP_{AP} (cm)	3.19 ± 0.46	2.69 ± 0.58	2.83 ± 0.29
COP_{ML} (cm)	2.56 ± 0.46	2.21 ± 0.04	2.63 ± 0.14
COP_{VL} (cm/s)	2.34 ± 0.08	1.95 ± 0.29	2.25 ± 0.11

Key: BFR = Blood flow restriction, NBFR = No blood flow restriction, COP_{AP} = Centre of Pressure - Anteroposterior, COP_{ML} = Centre of Pressure - Mediolateral, COP_{VL} = Centre of Pressure – Velocity

5.3.1.2. Single Leg Stance

For COP_{AP} (Table 5.3), a significant interaction was detected ($F_{(6, 42)}$ = 2.443, P = 0.022, n_p^2 = 0.289). The simple main effects revealed a significant decrease in COP_{AP} from pre- to post-exercise after ankle NBFR (5.37 ± 1.28cm to 4.31 ± 0.63cm, ∆19.73%, P = 0.026, d = 1.05) and knee NBFR (5.59 ± 1.00cm to 4.48 ± 1.07cm, ∆19.86%, P = 0.019, d = 1.07), with no change for BFR conditions. No significant difference was detected at post- and 10-min post-exercise between condition, although COP_{AP} was greater at pre- exercise for ankle NBFR compared to ankle BFR (P = 0.002, d = 1.42). For COP_{ML}, no significant interaction effect ($F_{(6, 42)}$ = 2.295, P = 0.115, n_p^2 = 0.277), or main effects of time ($F_{(2, 14)}$ = 0.29, P = 0.661, n_p^2 = 0.047) or condition ($F_{(3, 21)}$ = 4.635, P = 0.440, n_p^2 = 0.436) were detected. For COP_{VL}, no significant interaction effect was detected ($F_{(6, 42)}$ = 1.497, *P* = 0.207, n_p^2 = 0.200). However, a significant main effects of time ($F_{(2, 14)}$ = 5.020, *P* = 0.026, n_p^2 = 0.456) and condition ($F_{(3, 21)}$ = 5.215, *P* = 0.009, n_p^2 = 0.465) were detected. For time, the *post hoc* pairwise comparisons revealed a significant reduction in COP_{VL} from pre- to 10-min post-exercise (collapsed data: pre = 4.96 ± 2.00cm/s, 10-min post = 4.43 ± 1.85cm/s, *P* = 0.027, *d* = 0.28). For condition, the *post hoc* pairwise comparisons tests revealed that COP_{VL} was greater for ankle NBFR compared to ankle BFR (collapsed data: ankle NBFR = 5.03 ± 1.74cm/s, ankle BFR = 3.84cm/s ± 1.78, *P* = 0.008, *d* = 0.68), and was greater for knee NBFR compared to knee BFR (collapsed data: knee NBFR = 5.50 ± 1.84cm/s, knee BFR = 3.88 ± 1.95cm/s, *P* = 0.005, *d* = 0.85).

Table 5.3: Mean ± SD outcome measures for single leg stance position in older adults.

Condition	Pre	Post	10-min post*2
Ankle BFR*3			
COP_{AP} (cm)	2.46 ± 0.35*1	3.06 ± 0.82	2.41 ± 0.30
COP_{ML} (cm)	2.36 ± 0.10	2.03 ± 0.27	2.21 ± 0.10
COP_{VL} (cm/s)	4.38 ± 0.54	3.55 ± 0.66	3.58 ± 0.31
Ankle NBFR			
COP_{AP} (cm)	5.37 ± 1.28	4.31 ± 0.63*	4.61 ± 0.45
COP_{ML} (cm)	2.64 ± 0.11	3.09 ± 0.51	3.09 ± 0.04
COP_{VL} (cm/s)	5.63 ± 0.25	4.65 ± 0.02	4.82 ± 0.48
Knee BFR*3			
COP_{AP} (cm)	2.82 ± 0.54	2.89 ± 0.11	2.96 ± 0.09
COP_{ML} (cm)	2.05 ± 0.23	2.56 ± 0.09	2.34 ± 0.16
COP_{VL} (cm/s)	3.82 ± 0.18	3.85 ± 0.08	3.96 ± 0.18
Knee NBFR			
COP_{AP} (cm)	5.59 ± 1.00	4.48 ± 1.07*	5.24 ± 0.71
COP_{ML} (cm)	3.48 ± 0.20	3.32 ± 0.59	3.29 ± 0.04
COP_{VL} (cm/s)	6.02 ± 0.30	5.14 ± 0.90	5.35 ± 0.60

Key: BFR = Blood flow restriction, NBFR = No blood flow restriction, COP_{AP} = Centre of Pressure - Anteroposterior, COP_{ML} = Centre of Pressure - Mediolateral, COP_{VL} = Centre of Pressure – Velocity
*Significant within differences compared to pre- exercise ($P \leq 0.05$)
*1Significant difference pre- exercise between ankle BFR and ankle NBFR ($P \leq 0.05$)
*2Significant difference from pre- exercise ($P \leq 0.05$)
*3Significant difference between condition (ankle BFR and NBFR + knee BFR and NBFR) ($P \leq 0.05$)

5.3.2. Maximal Voluntary Isokinetic Contraction

With regards to MVC_{ISOK} performance for ankle plantarflexion, no significant interaction was detected ($F_{(1, 14)}$ = 0.039, *P* = 0.847, n_p^2 = 0.003) (Figure 5.2). The main effects analyses revealed a significant effect of time ($F_{(1, 14)}$ = 180.511, *P* = 0.001, n_p^2 = 0.928) but not for condition ($F_{(1, 14)}$ = 1.773, *P* = 0.204, n_p^2 = 0.112). The *post hoc* pairwise comparisons revealed a significant decrease in isokinetic ankle plantarflexion torque production pre- to post-exercise (collapsed data: pre = 17.72 ± 5.61Nm, post = 9.27 ± 3.08Nm, *P* < 0.001, *d* = 1.87). For knee extension, no significant interaction was detected ($F_{(1, 14)}$ = 0.060, *P* = 0.810, n_p^2 = 0.005). Main effects analyses revealed a significant effect of time ($F_{(1, 14)}$ = 87.917, *P* = 0.001, n_p^2 = 0.871) but not for condition ($F_{(1, 14)}$ = 0.166, *P* = 0.690, n_p^2 = 0.012). The *post hoc* pairwise comparisons revealed a significant decrease in isokinetic knee extensor torque production pre- to post-exercise (collapsed data: pre = 99.82 ± 23.33Nm, post: 74.72 ± 19.96Nm, *P* = 0.001, *d* = 1.15) (Figure 5.2).

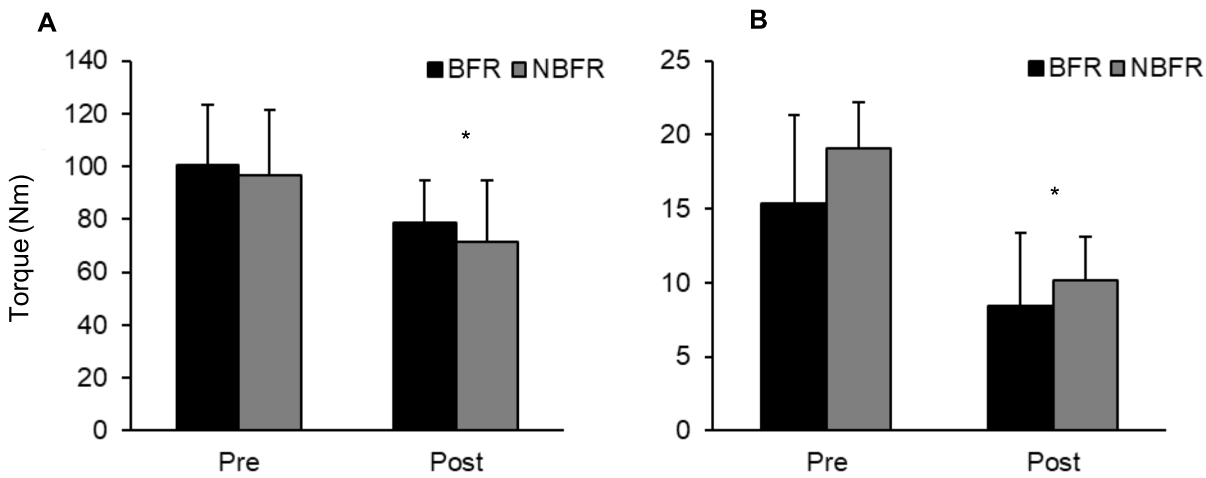

Figure 5.2: Mean ± SD for (A) knee extension and (B) ankle plantarflexion pre- and post-maximal voluntary isokinetic contraction (MVC_{ISOK}) exercise with blood flow restriction (BFR) and with no BFR (NBFR). *Significant difference from pre-exercise (*P* ≤0.05).

5.3.3. Ankle Surface Electromyography

For EMG_{TA} (Table 5.4), no significant interaction effect was detected ($F_{(1, 14)} = 0.730$, $P = 0.407$, $n_p^2 = 0.050$). the main effects analyses revealed a significant effect for time ($F_{(1, 14)} = 0.025$, $P = 0.001$, $n_p^2 = 0.643$) but not for condition ($F_{(1, 14)} = 0.745$, $P = 0.402$, $n_p^2 = 0.051$). The *post hoc* pairwise comparisons revealed a significant decrease in EMG_{TA} activation pre- to post-exercise (collapsed data: pre = 92.27% ± 6.82, post: 77.37% ± 13.29, $P = 0.009$, $d = 0.59$). For EMG_{GM}, there was no interaction effect ($F_{(1, 14)} = 0.105$, $P = 0.751$, $n_p^2 = 0.001$), or main effects of time ($F_{(1, 14)} = 0.011$, $P = 0.919$, $n_p^2 = 0.001$) or condition ($F_{(1, 14)} = 4.230$, $P = 0.059$, $n_p^2 = 0.232$) detected.

Table 5.4: Mean ± SD in Electromyography (EMG) amplitude pre- and post-ankle fatiguing exercise with blood flow restriction (BFR) and with no BFR (NBFR).

	Ankle Pre BFR	Ankle Post BFR	Ankle Pre NBFR	Ankle Post NBFR
EMG_{GM}	82.87 ± 16.07	81.15 ± 18.04	93.03 ± 7.64	94.05 ± 4.46
EMG_{TA}*	91.75 ± 3.93	74.19 ± 12.85	93.50 ± 9.12	79.09 ± 14.21

Key: GM = Gastrocnemius, TA = Tibialis Anterior. *Significant difference from pre-exercise ($P \leq 0.05$).

5.3.4. Knee Surface Electromyography

For EMG_{VM} (Table 5.5), no significant interaction effect was detected ($F_{(1, 14)} = 0.006$, $P = 0.941$, $n_p^2 = 0.243$). The main effects analyses revealed a significant effect of time ($F_{(1, 14)} = 5.030$, $P = 0.043$, $n_p^2 = 0.279$) but not for condition ($F_{(1, 14)} = 1.504$, $P = 0.241$, $n_p^2 = 6.107$). The *post hoc* pairwise comparisons revealed a significant decrease in EMG_{VM} pre- to post-exercise (collapsed data: pre = 92.42 ± 14.41%, post = 79.94 ± 22.41%, $P = 0.001$, $d = 0.66$). For the EMG_{VL}, no significant interaction effect ($F_{(1, 14)} = 1.924$, $P = 0.189$, $n_p^2 = 0.031$), or main effects of time ($F_{(1, 14)} = 2.853$, $P = 0.115$, $n_p^2 = 0.180$) or condition ($F_{(1, 14)} = 2.374$, $P = 0.144$, $n_p^2 = 0.137$) were detected.

Table 5.5: Mean ± SD in EMG amplitude pre- and post-knee fatiguing exercise with blood flow restriction (BFR) and with no BFR (NBFR)

	Knee Pre BFR	Knee Post BFR	Knee Pre NBFR	Knee Post NBFR
EMG_{VM}*	99.64 ± 10.54	87.72 ± 16.21	85.20 ± 13.04	72.16 ± 25.49
EMG_{VL}	83.18 ± 27.82	70.51 ± 27.23	89.71 ± 35.40	86.59 ± 12.33

Key: VM = Vastus Medialis, VL = Vastus Lateralis. *Significant difference from pre-exercise (*P* ≤0.05).

5.3.5. Gait Stability

For stride time, there was no significant interaction effect ($F_{(6, 42)}$ = 1.555, *P* = 0.217, n_p^2 = 0.341). The main effects analyses revealed no significant effect of time ($F_{(2, 14)}$ = 1.777, *P* = 0.248, n_p^2 = 0.372) or condition ($F_{(3, 21)}$ = 0.609, *P* = 0.501, n_p^2 = 0.067). For stride length, there was no interaction effect ($F_{(6, 42)}$ = 0.958, *P* = 0.474, n_p^2 = 0.193), or main effects of time ($F_{(2, 14)}$ = 0.650, *P* = 0.548, n_p^2 = 0.140) or condition ($F_{(3, 21)}$ = 2.189, *P* = 0.142, n_p^2 = 0.354). For double limb support, there was no interaction effect ($F_{(6, 42)}$ = 0.407, *P* = 0.867, n_p^2 = 0.092), or main effects of time ($F_{(2, 14)}$ = 1.525, *P* = 0.275, n_p^2 = 0.276) or condition ($F_{(3, 21)}$ = 1.950, *P* = 0.176, n_p^2 = 0.328) (Table 5.6).

Table 5.6: Mean ± SD outcome measure for gait stability in older adults

Condition	Pre	Post	10-min post
Ankle BFR			
Stride Time (s)	1.12 ± 0.10	1.14 ± 0.15	1.15 ± 0.12
Stride Length (cm)	100.78 ± 14.55	107.15 ± 13.49	109.74 ± 14.68
Double Support (s)	0.29 ± 0.27	0.30 ± 0.30	0.33 ± 0.29
Ankle NBFR			
Stride Time (s)	1.09 ± 0.09	1.14 ± 0.07	1.14 ± 0.12
Stride Length (cm)	99.75 ± 12.82	99.91 ± 14.24	102.67 ± 10.82
Double Support (s)	0.26 ± 0.15	0.37 ± 0.12	0.26 ± 0.22
Knee BFR			
Stride Time (s)	1.13 ± 0.13	1.10 ± 0.05	1.08 ± 0.05
Stride Length (cm)	93.80 ± 25.20	92.2 ± 14.70	99.4 ± 16.70
Double Support (s)	0.21 ± 0.20	0.15 ± 0.13	0.12 ± 0.15
Knee NBFR			
Stride Time (s)	1.01 ± 0.15	1.09 ± 0.19	1.06 ± 0.04
Stride Length (cm)	102.05 ± 16.32	103.44 ± 17.23	104.81 ± 12.73
Double Support (s)	0.28 ± 0.12	0.46 ± 0.22	0.35 ± 0.26

Key: BFR = Blood flow restriction, NBFR = No blood flow restriction

5.4. Discussion

The aim of this study was to investigate the acute effects of muscle fatigue with and with no BFR on the recovery of postural sway and gait characteristics in a cohort of older adults. The main findings from this study were (1) during DLS, postural sway was not affected by muscle fatigue with BFR or NBFR. However, during SLS, a reduction in COP_{AP} was observed pre- to post-exercise for ankle and knee NBFR conditions only, with a reduction in COP_{VL} pre- to 10-mins post exercise, (2) all measures of gait stability were unaffected by muscle fatigue, irrespective of BFR or NBFR conditions, (3) the muscle fatiguing protocol (independent of either BFR or NBFR conditions) successfully reduced muscle torque production indicative of fatigue, and (4) the reduction in torque production was associated with a decline in EMG_{TA} (ankle fatiguing protocol) and EMG_{VM} (knee fatiguing protocol). The results indicate that there is a lack of significant evidence of effect of fatigue and/or blood flow restriction on postural control. However, this does not rule out the possibility of a true effect, as the low power of our analyses may have restricted our ability to identify a small but meaningful difference. Therefore, a study with sufficient power is needed to investigate this issue further.

5.4.1. Characterisation of Muscle Fatigue

Minimising the loss of muscle strength at the ankle after fatiguing exercise is important for the maintenance of balance and prevention of excessive postural sway (Winter, 1995). In the current study, there was a significant reduction in isometric torque and EMG_{TA} following the ankle fatigue protocol, with no change observed in EMG_{GM}. In addition, the results also identified a reduction in EMG_{VM} for the knee, with no change in EMG_{VL} following the knee fatiguing protocol. As no interaction effect was detected, BFR was no different than with no BFR, with these results consistent with the observations in young adults (Chapter 4), where EMG_{TA} and EMG_{VL} were most affected at the ankle and knee post-exercise. The findings for MVC_{Isok} were also similar, where a reduction in torque production occurred at the ankle and knee post-exercise. Furthermore, the magnitude of the reduction in both MVC_{Isok} and EMG were similar in young and older adults and may suggest that both populations experienced a similar level of fatigue.

As ageing is associated with a steady decline in the ability of the neuromuscular system to affect specific processes (e.g., decline in motor units, motor axon conduction velocities (Christie and Kamen, 2009), the older muscle would seemingly be at a disadvantage and be less resistant to fatigue (Christie and Kamen, 2009). However, older adults are more fatigue resistant in comparison to young adults likely because of a greater proportion of slow twitch muscle fibres (Evans and Lexell, 1995), which may partly explain why the older adults were able to recover quickly during the transition from the muscle fatiguing protocol to assess quiet standing balance. Several studies have shown older adults to be more fatigue resistant during submaximal fatiguing protocols with BFR (Wernbom and Aagaard, 2019) and without BFR (Lanza, *et al*., 2004; Christie and Kamen, 2009). This greater fatigue resistance may point towards the type of contraction (concentric/ concentric) and mode of exercise (isokinetic) performed in the current study being a factor for the negligible changes in postural control were observed. In addition, the older adults recruited for the study were healthy and of a 'lower' age range (64 ± 2.45 years), thus it could be possible that the older adults were able to tolerate the fatiguing protocol and recover quickly. A consideration for the future would be choosing a more demanding mode of contraction (e.g., eccentric contraction) to cause more muscle damage and greater disturbance to postural sway.

5.4.2. Postural Sway

In contrast to the initial hypothesis, muscle fatigue had little effect on most postural sway measures during quiet standing. However, for SLS during both ankle and knee NBFR conditions, a significant decrease in COP_{AP} occurred immediately after fatiguing exercise compared to BFR where negligible changes to COP_{AP} were detected. These findings were contrary to our initial predictions. For example, performing muscle fatiguing exercise in older adults has been shown to increase COP_{AP} and COP_{VL} (Yaggie and McGregor, 2002; Gribble and Hertel, 2004; Vuillerme and Boisgontier, 2010; Bisson *et al*., 2012; Boyas *et al*., 2019) and is associated with a longer time to return to baseline measures of COP_{AP} and COP_{VL} (Lin *et al*., 2009; Kim *et al*., 2022). The decrease in COP_{AP} following ankle and knee NBFR fatigue may suggest a change in proprioceptive feedback occurred rather than a force mechanism, as the SLS requires forces approximately 10% of an individual's MVC (Bisson *et al*., 2012). According to the sensory re-weighing theory, tasks performed with a reduction in either visual, vestibular, or somatosensory input increases the demand of the neuromuscular system to dynamically and selectively adjust the contribution of the sensory input to maintain upright posture (Assländer and Peterka, 2014). Moreover, the SLS task was performed with EO compared to EC for the DLS task, which may suggest an increase in proprioception gained from the muscles of the lower limb and access to vision may have contributed to maintaining SLS. Equally, the muscles targeted were the main ankle dorsiflexor (tibialis anterior) and plantarflexor (gastrocnemius) responsible for controlling sagittal plane movement (Tokuno *et al*., 2007; Patterson and Ferguson, 2011). This often leads to individuals adopting an ankle strategy movement to stabilise COP_{AP} displacement (Gribble and Hertel, 2004) or through stiffening the musculature at the ankle or hip (Sakanaka *et al*., 2021). However, this cannot be verified in the current study, as other COP outcomes (e.g., mean frequency of COP) were not calculated, nor was muscle activity measured during quiet standing to determine whether stiffening occurred at the ankle or hip. Although, based on the initial observations from this study, there is some thought that the older adults may have adopted a hip strategy or an increase in stiffening at the ankle occurred to prevent excessive movement after muscle fatigue.

Similar to the findings observed in Chapter 4 in young adults and contrary to the initial hypothesis, fatigue combined with BFR did not disturb postural sway. Previous review articles suggest that BFR trials lead to quicker muscle fatigue due to an increase in metabolite production (Wembrom *et al*., 2019; Baker *et al*., 2020). The assumption here is the addition of a cuff would fatigue the target muscle quickly compared to NBFR condition. However, our findings suggest that fatigue combined with BFR did not reduce postural control despite the reduction in torque production across both conditions. As previously discussed in Chapter 4, the low BFR pressure may have led to less metabolite accumulation in the exercising muscles. In relation to the point, reactive hyperemia may occur where the reperfusion of blood to the exercising muscle aided recovery of the muscles following the fatiguing protocols (Lim *et al*., 2022). Therefore, the lower BFR pressure combined with a low-moderate intensity may not have placed a high enough stress on the target muscle, where oxygen may be still able to be delivered to the muscle and may have contributed to a faster recovery (Fatela *et al.*, 2008; Karabulut *et al*., 2013). A future consideration would be to monitor oxygen saturation level by performing plethysmography on the target muscle during exercise. Performing this assessment would help to determine the level of oxygen entering the restricted limb and may provide a better insight on the level of BFR vs. the level of fatigue (e.g., 50% MVC threshold).

5.4.3. Gait Stability

The results from the spatiotemporal measures were contradictory to the initial hypothesis where muscle fatigue (independent of condition) did not influence any measure of gait stability. Immediately after each muscle fatiguing trial, all participants performed quiet standing tasks which involved the participant moving from the dynamometer and performing the DLS and SLS balance tasks. The time between the muscle fatiguing trial to performing the assessment of gait on the treadmill may have been a contributing factor. As with Chapter 4, the possibility that the rest period may have been enough time to restore proprioception at the ankle, knee, and hip through the recruitment of additional motor units or the contribution of addition muscle(s) through co-activation (Donath *et al*., 2016) while performing quiet standing tasks cannot be excluded. However, the minimal effects observed during the reassessment of postural sway post-exercise leans towards the intensity or mode of exercise performed during the interventions was not enough to disturb quiet standing nor affect dynamic balance. Although, based on the results observed in young and now older adults, it appears that both populations responded in a similar fashion where gait stability was unaffected by BFR and NBFR conditions. This study is the first to investigate specific spatiotemporal measures of gait stability in young and older adults. These findings indicate that BFR was no worse than NBFR after fatiguing ankle and knee resistance exercise. However, the absence of a statistically significant finding in gait stability should not be seen as evidence of no effect. Rather, insufficient statistical power, therefore a fully powered designed study may be needed to draw a more reliable conclusion. Nonetheless, from these initial findings, clinicians could proceed with confidence knowing a bout of BFR exercise in young but especially older adults would not lead to an adverse effect to stability following lower limb exercise.

5.4.4. Limitations

Limitations that require acknowledgment include, firstly, EMG activity during quiet standing and gait trials was not measured, which would have provided a better understanding of the changes in muscle co-activation patterns following fatigue during these tasks. Secondly, only muscles responsible for sagittal plane movement were fatigued, where fatiguing other musculature (invertor and evertor muscles – frontal plane movement) that are also important for postural control may have resulted in greater disturbances in balance and gait. Traditionally, fatiguing the muscles of the sagittal plane are used in research, but it would be interesting to test frontal plane movements as these are important contributors to maintain postural sway in older adults. Thirdly, only postural sway and gait stability were measured, with other factors associated with falls (e.g., sit to stand test) not included in the test battery, which could limit the ecological validity of the study. In addition, this study was performed in healthy older adults, and therefore the findings cannot be generalised to more frail groups who may be a greater risk of falls. Finally, an important limitation of this study is the small sample size, which hinders the ability to make a definitive conclusion about the effectiveness of this approach. The reduced statistical power should be interpreted as insufficient evidence for an effect, rather than a conclusive indication of no effect. Nevertheless, in this exploratory study, it was essential to assess the safety of BFR in older adults and to understand the immediate effects of BFR exercise on postural control and gait stability during fatiguing exercise.

5.4.5. Summary

The findings from this study indicate that despite substantial fatigue being induced as evidenced by the reduction in torque, there were minimal changes in postural sway with no alteration observed for measures of gait stability after ankle or knee muscle fatigue protocol in either BFR or NBFR conditions. Although, there were significant changes to postural sway for NBFR condition, the reduction in postural sway in the presence of fatigue could be interpreted as a stiffening response that is mostly considered a maladaptive response but did not happen during BFR conditions. This finding alone could be helpful when prescribing low-intensity resistance exercise with BFR in the knowledge that the potential to increase the risk of fall post-exercise would be minimal in older adults. However, this should be interpreted with caution, as similar to the Chapter 4, the lack of evidence of effect in this study is due to the insufficient statistical power in this study. Therefore, a fully powered study is required. Overall, this work contributes to a better understanding of the acute effects of BFR exercise on postural sway compared to normal fatiguing conditions in both young (Chapter 4) and older adults.

Chapter 6:

The Effects of 6-Weeks Low Intensity Resistance Training with Simultaneous Blood Flow Restriction versus High Intensity Resistance Training on Characteristics of Muscular Strength, Proprioception, Postural Sway, and Gait Stability in Older Adults

Highlights:

- Low intensity resistance training with blood flow restriction (LIBFR) is a unique, and potentially effective, alternative to high intensity (HI) resistance exercise.
- LIBFR and HI training were effective at increasing characteristics of strength and proprioception.
- LIBFR and HI training did not elicit any changes in postural sway and gait stability.

6.1. Introduction

A decline in both physical function and muscle mass (sarcopenia) can directly affect activities of daily living (ADL) in older adults (Cruz-Jentoft *et al*., 2019) and is consistently associated with an increase in falls, fractures, hospitalisation, and poorer quality of life (Perry *et al*., 2007; Christensen *et al*., 2009; Manini and Pahor, 2009; Bull *et al*., 2020). Ageing is associated with a decline in postural control and muscle strength, which presents two important modifiable risk factors to target from a fall prevention perspective (Granacher *et al*., 2011; Sherrington *et al*., 2011). To slow down the effects of ageing, HI resistance training is widely recommended to preserve muscle quality and to attenuate some of the physiological markers linked with ageing (Metter *et al*., 1997; Liu and Latham, 2009; Barbat-Artigas *et al*., 2014; Fragala *et al*., 2014 Plaza-Florido *et al*., 2020), which is closely linked to enhancing physical function (Lexell, 2000; Lee and Park, 2013).Consequently, the National Institute of Health and Care Excellence (NICE) exercise guidelines (NICE, 2008) include twice weekly resistance training for older adults to combat functional decline.

Regular resistance exercise can delay the physiological decline of normal ageing when performed at moderate-to-high intensity (Seguin and Nelson, 2003; Howe *et al*., 2011), with improvements in muscle strength and hypertrophy being most prominent when training with high external resistance (workload) (Moreland *et al*., 2004; Granacher *et al*., 2011; Schoenfeld *et al*., 2015). However, performing resistance exercise at a high intensity (ranging from 60 – 90% of an individual's 1RM) to improve muscular strength and hypertrophy (Plaza-Florido et al., 2020), is often too challenging for older adults (Chao et al., 2000; Richardson et al., 2018), especially in those with sarcopenia (Manini and Pahor, 2009; Grønfeldt, et al., 2020) and dynapenia (Manini and Clark, 2012). Moreover, if the resistance is too high this may lead to low tolerance, aversive responses, and consequently poor adherence (Rivera-Torres et al., 2019). Therefore, the overall tolerability and effectiveness of resistance training with high loads in older adults is questionable.

One method of resistance training that has grown in popularity over recent years is to combine low-intensity resistance training with blood flow restriction (LIBFR) (Wernbom and Aagaard, 2019). This alternative method of training produces comparable adaptations associated with HI resistance training (e.g., muscle hypertrophy, motor unit recruitment, increase in fast twitch muscle fibre response) (Yasuda *et al.,* 2005; Laurentino *et al*., 2008; Karabulut *et al.,* 2013; Wernbom and Aagaard, 2019) but requires a lower effort of exertion and physiological cost (Abe *et al*., 2006). However, little is known about the impact LIBFR training has on measures of physical function (e.g., timed up and go [TUG], sit-to-stand [STS]) and postural control in older adults. To the author's knowledge, only two studies have compared LIBFR to moderate-to-high intensity after 6 and 12 weeks of resistance training in older adults on quiet standing (double and single leg balance; Bigdeli *et al*., 2020; Linero *et al*., 2021) and physical function (TUG; Bigdeli *et al*., 2020). The findings from Bigdeli *et al*. (2020) and Linero *et al*. (2021) reported improvements in quiet standing after LIBFR training but did not use techniques to assess the centre of pressure (COP) through reliable quantitative methods (e.g., force platform). Their approach used a rudimentary assessment of double (Bigdeli *et al*., 2020) and single (Linero *et al*., 2021) leg balance, which can provide information about balance impairments but not on the underlying problem or mechanisms that improved the participants performance. Therefore, using a quantitative assessment to assess quiet standing could provide a thorough and better understanding on postural performance using a force platform (e.g., postural sway) to assess global COP variables (e.g., centre of pressure – anterior posterior = COP_{AP}) (Paillard and Noe, 2015). Despite the initial inquiry to investigate the effects of LIBFR on postural control in older adults, the current interpretations drawn from the literature are limited by the lack of focus on proprioception (e.g., joint position sense [JPS]), gait stability, and balance performance after performing LIBFR training. No studies have investigated the potential mechanisms driving improvement in strength, proprioception, or balance. Therefore, it may be helpful to understand how the neuromuscular system responds to LIBFR training in older adults, which may provide insight into the potential benefits of this training technique for a population that may not tolerate HI resistance training, making it difficult for them to achieve meaningful results. Thus, the current study aimed to compare the effects of LIBFR and HI resistance training on quiet standing and dynamic (gait stability) balance performance in older adults. This study will also assess the changes in the neuromuscular mechanisms (muscle strength, muscle coordination patterns, JPS, and force sense (FS) associated with any improvements in postural control and gait characteristics.

Research Hypothesis (H_1): Significant improvements in strength, quiet standing and dynamic balance will occur after 6-weeks of HI and LIBFR exercise. Furthermore, measures of neuromuscular control associated with force and joint position sense will improve over the 6-weeks for both LIBFR and HI.

6.2. Methods

6.2.1. Participants

Sixteen male (*n* = 6) and female (*n* = 10) recreationally active older adults (Table 6.1) volunteered to take part in a 6-week (12 sessions) randomised control, parallel-designed study. The inclusion and exclusion criteria used to screen all participants can be reviewed in Chapter 3; section 3.3.1 – 3.3.2. Originally, eighteen participants volunteered for the study. One participant had to withdraw 3 weeks into the study due to a short illness which led to missing training sessions. A sample size calculation was performed using *priori* power (G*power: version 3.1.9.2, Universitat Kiel, Dusseldorf, Germany; Faul *et al*., 2007). The calculation for a mixed model ANOVA revealed that 28 participants were required to detect a moderate effect size (Cohen's

f = 0.25) with α 0.05 and a 1-β error of 0.80 suggested (Appendix 9.3.2). This training study is exploratory and aims to establish a proof of concept. A follow up *post hoc* power calculation was performed based on 16 participants recruited for the current study suggesting the study was underpowered (medium standardised effect of Cohen's f = 0.25 with α 0.05 and a 1-β error of 0.53 [Appendix 9.3.1]).

Table 6.1: Participant Characteristics

	BFR (*n* = 8)	NBFR (*n* = 8)
Age (yrs)	62.38 ± 2.26	62.75 ± 1.80
Height (cm)	170.49 ± 8.48	169.83 ± 12.91
Body mass (kg)	76.39 ± 9.75	69.57 ± 17.27

Key: cm =centimetres, kg = kilograms, yrs = years, Data expressed as means ± SD.

6.2.2. Data Collection

All participants were required to attend a pre (0 – weeks), mid (3 weeks) and post (6 weeks) data collection session (Figure 6.1). All participants had their height (cm), weight (kg) (Chapter 3; section 3.17) and resting blood pressure (mmHg) taken (Chapter 3; sections 3.9.1 and 3.9.3). All participants were instructed and asked to perform a series of balance, gait, strength, and proprioception tests, in addition to being familiarised with the questionnaires (Appendix A6 and A7) completed for each training session.

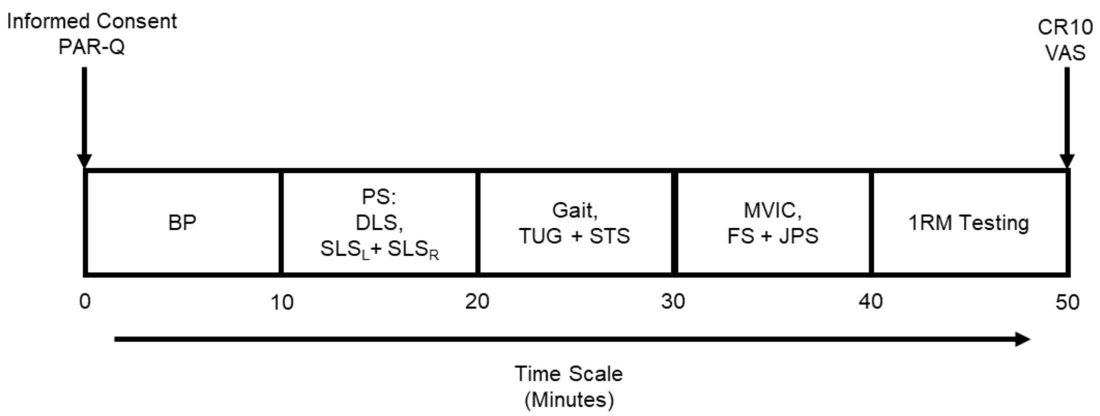

Figure 6.1: A schematic identifying the data collection procedure complete by all participants for pre, mid and post time points. Key: BP = Blood pressure, CR10 = Discomfort scale – Category ratio-10, DLS = double leg stance, EC = Eyes closed, FS = Force sense, JPS = Joint position sense, MVIC = Maximum voluntary isometric contraction., PAR-Q = Physical Activity Readiness Questionnaire, PS = Postural sway, SLS_L = Single leg stance – left leg, SLS_R = single leg stance – right leg, TUG = Timed up and Go, STS = Sit to stand, VAS = Visual analogue scale.

6.2.3. Blood Pressure

Systolic (SBP) and diastolic (DBP) blood pressure (mmHg) were taken pre- and post-exercise for the BFR group using the left arm to collect resting, with the procedures taken for the collection of blood pressure detailed in Chapter 3; section 3.9.1.

6.2.4. Postural Sway

Ten minutes later, all participants performed two stance positions in a randomised order: 1) Double leg stance (DLS) with eyes closed performed on a fixed firm surface, and 2) Single leg stance position with eyes open on the left (SLS_L) and right (SLS_R) limb (Lin *et al.,* 2008; Ruhe *et al*., 2010). Each position was held for 30 seconds and repeated three times. Data were extracted from the force platform with the following postural sway measures used for analysis: COP_{AP}; COP_{ML} and COP_{VL} (Moghadam *et al.,* 2011). The calibration and sampling procedures for the assessment of postural sway can be viewed in Chapter 3, section 3.4.1.

6.2.5. Gait Stability

Ten minutes later, all participants walked at a self-selected speed for 2 minutes on a treadmill that reflected 'a normal pace' the participant would usually adopt in their normal 'day to day' life. During the final minute, the participants' mean stride time (seconds; s), stride length (cm) and double support phase (s) were recorded. The specific details outlining the calibration and sampling procedures for the assessment of gait stability can be reviewed in Chapter 3; section 3.8.3.

6.2.6. Timed Up and Go

Five minutes later, all participants performed the timed up-and-go test (TUG) using a standard chair (seat height ~43 cm). The specific details describing the setup and protocol can be viewed in Chapter 3; Section 3.8.2. The ICC was reported as 0.84 for the current study demonstrating an excellent level of reliability.

6.2.7. Sit to Stand Test

Two minutes later, all participants performed the sit-to-stand (STS) test using the same chair performed in Chapter 6; section 6.2.1. The sit used to assess STS was used to assess time (STS^T) and power (STS^P). The specific protocol and set up used to assess STS^T and STS^P can be viewed in Chapter 3; Section 3.8.1, along with the equation used to calculate STS^P. The ICC was reported as 0.90 for STS^T and 0.98 for STS^P demonstrating an excellent level of reliability.

6.2.8. Isokinetic Dynamometry

After 5 minutes of rest, the participants performed three maximum voluntary isometric contractions (MVIC) for knee extension (Chapter 3; section 3.6.8), followed by assessing FS for knee extension (Chapter 3; Section 3.6.1) and JPS for knee extension and flexion (Chapter 3; Section 3.6.2). For FS, 10% (FS10) and 20% (FS20) of the participants MVIC were calculated and used. The ICC was reported as moderate - good for FS10 (ICC: 0.49) and excellent for FS20 (ICC = 0.86). For JPS, 45° and 70° ranges of motion were used to assess knee extension and flexion. An ICC of 0.56 for knee extension at 45° (JPS Ext 45°), 0.24 for knee extension at 70° (JPS Ext 70°), 0.55 for knee flexion at 45° (JPS Flex 45°) and 0.24 for knee flexion at 70° (JPS Flex 70°), showing a range of poor-to-moderate reliability. Details on the set-up of the dynamometer and procedures used to calibrate and adjust for gravity corrections can be reviewed in section (Chapter 3; section 3.6.1 – 3.6.3).

6.2.9. Assessment of Muscle Strength

Ten minutes later, the participants performed a ten-repetition maximum (10RM) protocol to estimate one-repetition maximum (1RM) (Reynolds *et al*., 2006), which would be used to determine the weight when performing the barbell box squat (depth set to ~43cm) and calf raise exercises. Before completing each exercise, all participants performed a warm-up set (10 repetitions) with a standard training bar (~5kgs; Perform Better, Warwickshire, UK). After a two-minute rest period, the participants chose a weight they felt was challenging enough but manageable to perform either the squat or calf exercise through their full range of motion before attempting the 10RM trial (Knutzen *et al*., 1999). The weight for each exercise was increased until momentary failure at 10RM occurred. If the participant could lift more weight than the prescribed weight, then an additional 3-minute rest period was provided, with the weight increased by 10-15% (Cook *et al*., 2017; Richardson *et al*., 2018). The final weight lifted was recorded and used to calculate each participant's individual 1RM using a prediction equation (weight lifted [kg] / 1.0278 – (0.0278 x number of repetitions: Brzycki, 1999). The prediction equation used for estimating 1RM has been reported as a valid method, in line with other estimated 1RM calculations used in older adults (Knutzen *et al*., 1999). The estimated 1-RM from a 10RM test was reassessed at week 3 to adjust the participant's weight for the following 3 weeks (Cook *et al*., 2017).

6.2.10. Subjective Measures

6.2.10.1. Discomfort Scale

During the 10RM session and during every training session, all participants were asked to rate their perceived muscular discomfort from a Borg rating of perceived discomfort scale (CR10) (Borg, 1998) to gauge the participant's discomfort after each set (Loenneke *et al*., 2016) for the squat and calf raise exercise. The CR10 works on a scale from 0-10 and has several anchors to help the participant rate their level of perceived exertion (Hollander *et al*., 2003; Loenneke *et al*., 2016). All participants were asked to read a set of instructions on the CR10 and confirmed that they fully understood the instructions before each session. The anchors provided and instructions used for the CR10 can be viewed in Appendix A6.

6.2.10.2. Perceptual Outcomes

After the data collection session and for each training session, all participants were asked to rate their enjoyment, fatigue, perception of effort (volume-load) and perceived effectiveness using a visual analogue scale (Kuys *et al*., 2011; Richardson *et al*., 2018). Each statement was provided on an A4 sheet (portrait) piece of paper with a 100mm horizontal line spanning from the left (e.g., perception of volume/load = No workload) to the right (e.g., perception of volume/workload = high workload) hand side below each statement (Appendix A7). All participants were asked to mark a vertical line based on their feelings at that moment in time.

6.2.11. Thigh Tourniquet

Finally, all participants in the BFR group were asked to familiarise themselves with the thigh tourniquets using the set-up described in Chapter 3; sections 3.15 – 3.16. The two thigh tourniquets were applied after the participant warm-up protocol on the treadmill before completing the resistance training programme. The thigh tourniquets were inflated to 1.2 x of the participants resting SBP (mmHg) in a staggered fashion using the steps provided in Chapter 3; section 3.16.

6.2.12. Training Protocol

Prior to taking part in the training study, all participants were randomised (www.randomizer.org) and allocated to either the LIBFR or HI training group, where the workload of each group was matched (total weight lifted = repetitions x sets x relative load). The workload for each group was based on previous LIBFR training protocols consisting of the same number of repetitions x set (Loenneke *et al.*, 2012c; Sousa *et al.*, 2017). The participants attended each session at the University of Northampton Sports Strength Laboratory for each training session (~50 minutes per session; Figure 6.2). All participants selected for the LIBFR group had their resting blood pressure taken (Chapter 3; section 3.9.1). Two minutes later, the LIBFR and HI groups performed a warm-up protocol consisting of a self-selected paced walk on a motorised treadmill (Chapter 3; section 3.8.3) before completing the resistance exercise programme. For the LIBFR group, a thigh tourniquet to the upper portion of each leg was worn and set to a specific pressure based on their resting SBP (Chapter 3; section 3.15 – 3.16) (Figure 6.2).

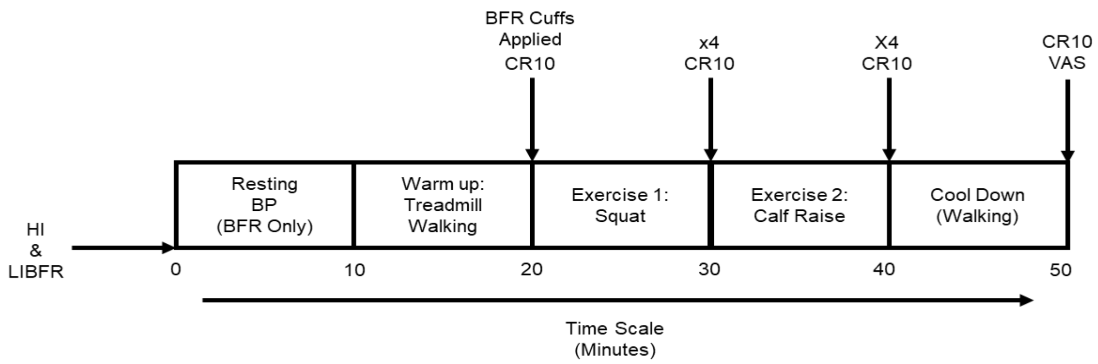

Figure 6.2: Training protocol for HI and LIBFR groups.

Each resistance training session consisted of performing a bilateral squat and calf raise exercise (Figure 6.3) using a standard 10 or 20kg Olympic barbell (Perform Better, Warwickshire, UK). The depth (range of motion) of the squat was reduced using a foam plyometric box (Perform Better, Warwickshire, UK) placed approximately ~15cm behind the participant's feet. The height of the plyometric box replicated the chair height used for the STS test (height: ~43cm). All participants squatted to the depth of the plyometric box and ascended once the buttocks touched the box. The calf raise exercises were performed in an upright position with feet shoulder-width apart and feet on the floor. The participants were instructed to fully plantarflex at the ankle and flex the proximal and distal phalangeal joints (command: to the top of your toes was given as a cue) to ensure maximum plantarflexion was achieved with each repetition. A momentary isometric hold (~1 second) was instructed at the top of the movement with a controlled descent before performing the next repetition.

Figure 6.3: Squat (A) and Calf raise (B) exercises completed in each training session.

All participants in the LIBFR group completed 1 set of 30 repetitions followed by 3 sets x 15 repetitions at 35% 1RM. This training sequence has been reported in previous studies for both squat (Fahs *et al*., 2012; Scott *et al*., 2017) and calf (Patterson and Ferguson, 2011; Kjeldsen *et al*., 2019) exercises. The HI group completed one set of 15 repetitions and three sets of 8 repetitions at 70% 1RM. For both groups, there was a 90-second rest period between sets and a 240-second rest before completing the next exercise (Scott *et al*., 2015). The tourniquet pressure for the LIBFR group was maintained during each exercise and was deflated immediately at the end of the exercise (completion of 4 sets) (Letieri *et al*., 2018). The tourniquets were inflated back to the target blood pressure before completing the final exercise repeating the steps in Chapter 3; section 3.17 (Figure 6.4).

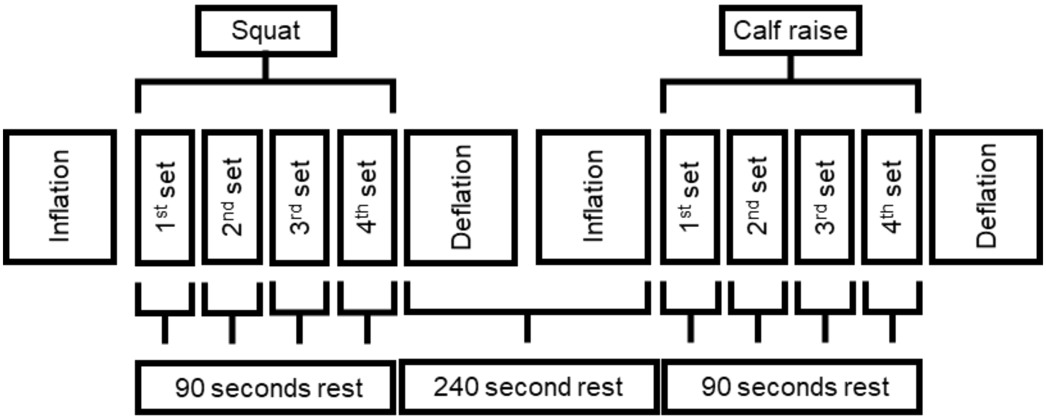

Figure 6.4: The resistance training protocol used for both HI and LIBFR.

6.2.13. Statistical Analysis

All data were analysed using SPSS 26.0 (SPSS Inc, Chicago, ILL, USA) and presented as mean ± standard deviation. The data were initially screened for outliers and checked for normal distribution (Shapiro-Wilk Test) and homogeneity of variance (Levenes Test). The average of three trials for $COP_{AL,}$ COP_{ML}, COP_{VL} for each stance position (DLS, SLS_L and SLS_R) was determined before further analysis. In addition, the mean of five trials for FS and the average of three trials for JPS, MVIC, STS^P, STS^T and TUG were used in the analysis. A mixed-model two-way ANOVA (Group [LIBFR vs. HI] x Time [Pre-training vs. Mid-training vs. Post-training]) was used to calculate differences between all dependent variables. Mauchly's test of sphericity was used to determine whether sphericity was violated and when confirmed, where epsilon (ε) was >0.75 then a Huynh-Feldt correction was used and if the ε was below <0.75 then a Greenhouse-Geisser correction was employed (Muller and Barton, 1989; Abdi, 2010). If a significant interaction effect was detected, then follow-up simple main effects analyses using Bonferroni correction (pairwise comparisons) were as performed, where no interaction effect was detected the data sets were collapsed with main effects analyses conducted. Effect sizes were calculated from the ANOVA and were displayed as partial eta-squared (n_p^2) and were defined as small (<0.06), medium (0.06-0.13), and large (≥0.14), whilst the magnitude of change (Δ) and Cohen's *d* was used for *post hoc* testing and were defined as negligible (<0.2), small (0.20-0.49), moderate (0.50-0.79) or large (≥0.80) (Cohen, 1988; Lakens, 2013). The alpha value was *a priori* set at P <0.05 for all tests.

6.3. Results

6.3.1. Centre of Pressure

6.3.1.1. Double Leg Stance

For COP_{AP}, no significant interaction effect ($F_{(2, 28)}$ = 1.264, P = 0.290, n_p^2 = 0.085), or main effects of time ($F_{(2, 28)}$ = 2.454 , P = 0.104, n_p^2 = 0.152) or group ($F_{(1, 14)}$ = 0.155, P = 0.700, n_p^2 = 0.011) were detected. For COP_{ML}, no significant interaction effect was detected ($F_{(2, 28)}$ = 1.163 , P = 0.311, n_p^2 = 0.077), but a main effect of time ($F_{(2, 28)}$ = 6.328, P = 0.016, n_p^2 = 0.311) but not of group ($F_{(1, 14)}$ = 0.075, P = 0.839, n_p^2 = 0.003) were detected. The *post hoc* pairwise comparisons revealed a significant increase in COP_{ML} from the pre- to mid-testing points (collapsed data: pre = 2.33 ± 1.01cm, mid = 2.97 ± 0.86cm, P = 0.041, d = 0.68). For COP_{VL}, there was no significant interaction effect ($F_{(2, 28)}$ = 1.551, P = 0.230, n_p^2 = 0.100), or main effects of time ($F_{(2, 28)}$ = 0.545, P = 0.586, n_p^2 = 0.037) or group ($F_{(1, 14)}$ = 0.215, P = 0.650, n_p^2 = 0.015) (Table 6.2).

6.3.1.2. Single Leg Stance – Left Leg

For COP_{AP}, no significant interaction effect ($F_{(2, 28)}$ = 2.235, P = 0.187, n_p^2 = 0.113), or main effects of time ($F_{(2, 28)}$ = 2.311, P = 0.118, n_p^2 = 0.142) or group ($F_{(1, 14)}$ = 0.175, P = 0.821, n_p^2 = 0.004) were detected. For COP_{ML}, no significant interaction effect ($F_{(2, 28)}$ = 1.223, P = 0.376, n_p^2 = 0.065), or main effects of time ($F_{(2, 28)}$ = 1.609, P = 0.222, n_p^2 = 0.103) or group ($F_{(1, 14)}$ =0.004, P = 0.949, n_p^2 = 0.001) were detected. For COP_{VL}, no significant interaction effect ($F_{(2, 28)}$ = 0.075, P = 0.882, n_p^2 = 0.005), or main effects of time ($F_{(2, 28)}$ = 1.193, P = 0.166, n_p^2 = 0.120) or group ($F_{(1, 14)}$ = 0.391, P = 0.542, n_p^2 = 0.027) were detected (Table 6.2).

6.3.1.3. Single Leg Stance – Right Leg

For COP_{AP}, no significant interaction effect ($F_{(2, 28)}$ = 1.057, P = 0.357, n_p^2 = 0.070), or main effects of time ($F_{(2, 28)}$ = 0.330, P = 0.704, n_p^2 = 0.023) or group ($F_{(1, 14)}$ = 0.326, P = 0.577, n_p^2 = 0.023) were detected. For COP_{ML}, no significant interaction effect ($F_{(2, 28)}$ = 0.183, P = 0.778, n_p^2 = 0.013), or main effects time ($F_{(2, 28)}$ = 2.143, P = 0.150, n_p^2 = 0.133) or group ($F_{(1, 14)}$ = 0.024, P = 0.880, n_p^2 = 0.002) were detected. For COP_{VL}, no significant interaction effect ($F_{(2, 28)}$ = 0.206, P = 0.707, n_p^2 = 0.015), or main effects of time ($F_{(2, 28)}$ = 1.010, P = 0.347, n_p^2 = 0.067) or group ($F_{(1, 14)}$ = 0.185, P = 0.985, n_p^2 = 0.001) were detected (Table 6.2).

Table 6.2: Mean ± SD outcomes measures for quiet standing (postural sway) and dynamic (mean gait stability) balance after 6-weeks of LIBFR and HI resistance training.

		LIBFR			HI		
		Pre	**Mid**	**Post**	**Pre**	**Mid**	**Post**
Postural Sway							
	COP_{AP} (cm)	3.36 ± 1.63	3.49 ± 1.02	3.62 ± 0.74	2.89 ± 1.12	4.10 ± 1.20	4.00 ± 1.02
DLS	COP_{ML} (cm)*	2.54 ± 1.12	2.90 ± 1.17	2.32 ± 0.80	2.12 ± 0.92	3.05 ± 0.42	2.37 ± 0.77
	COP_{VL} (cm/s)	1.80 ± 0.42	1.82 ± 0.42	1.99 ± 0.38	1.89 ± 0.56	2.10 ± 0.50	1.89 ± 0.48
	COP_{AP} (cm)	4.94 ± 1.52	3.90 ± 1.05	3.98 ± 1.24	4.19 ± 1.51	3.61 ± 1.23	4.70 ± 1.69
SLS_L	COP_{ML} (cm)	4.32 ± 0.72	4.22 ± 1.57	4.35 ± 1.77	4.55 ± 1.63	3.71 ± 1.53	4.79 ± 2.20
	COP_{VL} (cm/s)	3.24 ± 1.24	3.11 ± 0.92	2.98 ± 0.89	2.96 ± 0.73	2.83 ± 0.59	2.77 ± 0.65
	COP_{AP} (cm)	4.31 ± 1.54	3.98 ± 0.71	3.78 ± 1.00	3.94 ± 2.25	4.80 ± 2.15	4.35 ± 1.28
SLS_R	COP_{ML} (cm)	3.50 ± 0.41	3.40 ± 1.10	3.30 ± 0.60	3.28 ± 1.33	3.64 ± 1.81	4.76 ± 1.62
	COP_{VL} (cm/s)	2.88 ± 0.92	2.83 ± 0.84	2.93 ± 0.91	2.90 ± 1.14	2.71 ± 0.65	3.00 ± 0.84
Gait Stability							
Stride Time	(s)	1.17 ± 0.11	1.15 ± 0.11	1.10 ± 0.10	1.08 ± 0.06	1.10 ± 0.11	1.07 ± 0.06
Stride Length	(cm)	96.48 ± 16.15	104.65 ± 10.03	106.61 ± 19.74	101.53 ± 11.86	102.46 ± 12.31	103.22 ± 8.23
Double support	(s)	0.31 ± 0.20	0.37 ± 0.26	0.26 ± 0.14	0.23 ± 0.11	0.25 ± 0.14	0.23 ± 0.14

Key: DLS = Double Leg Stance, SLS_L = Single Leg Stance – Left Leg, SLS_R = Single Leg Stance – Right Leg, COP_{AP} = Centre of Pressure - Anteroposterior, COP_{ML} = Centre of Pressure - Mediolateral, COP_{VL} = Centre of Pressure - Velocity, LIBFR = Low-intensity Blood Flow Restriction, HI = High Intensity Resistance Training *significant difference pre – mid time testing time points.

6.3.2. Functional Performance

6.3.2.1. TUG

For TUG, there was no significant interaction effect ($F_{(2, 28)}$ = 0.700, *P* = 0.505, n_p^2 = 0.048), but a main effect of time ($F_{(2, 28)}$ = 15.522, *P* = <0.001, n_p^2 = 0.988) but not of group ($F_{(1, 14)}$ = 0.071, *P* = 0.793, n_p^2 = 0.005) were detected. The *post hoc* pairwise comparisons revealed that TUG performance was significantly slower at pre- compared to mid- (collapsed data: pre: 7.52 ± 0.98s, mid: 6.83 ± 0.72s, *P* = 0.003, *d* = 0.80) and pre- compared to post-training (collapsed data: pre 7.52 ± 0.98s, post: 6.58 ± 0.78s, *P* = 0.002, *d* = 1.06) time points (Figure 6.5).

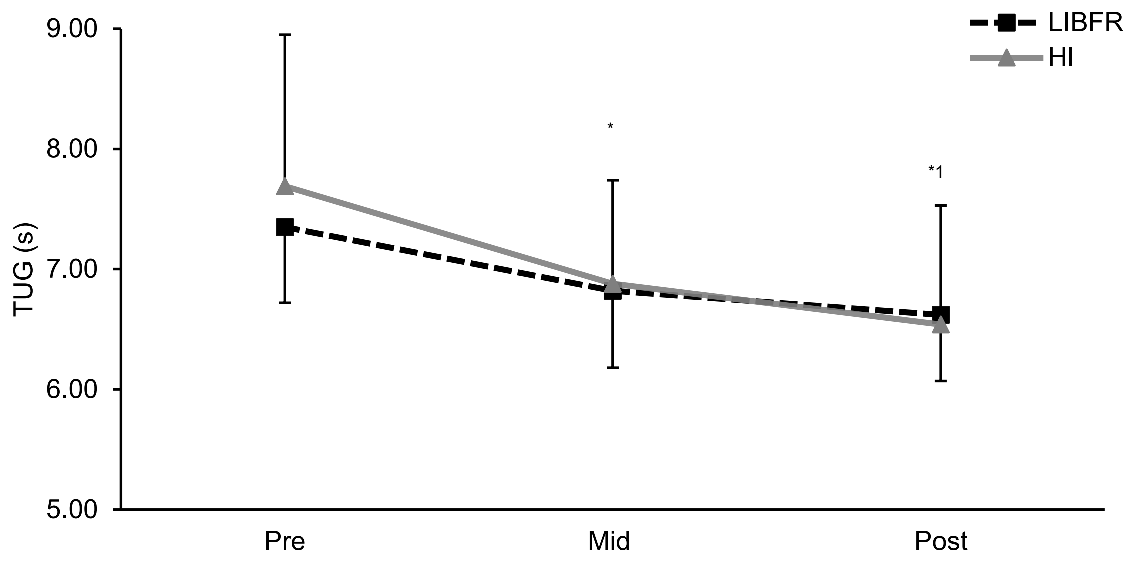

Figure 6.5: Mean ± SD in timed up and go (TUG) following 6 weeks of LIBFR and HI training. *Significant difference from pre- to mid-training testing point. *1Significant difference from pre- to post-training testing points (*P* ≤0.05).

6.3.2.2. STS^T

For STS^T, a significant interaction effect was detected ($F_{(2, 28)}$ = 3.531, *P* = 0.047, n_p^2 = 0.201). Simple main effects revealed a significantly faster STS time from pre- to mid- (pre = 9.92 ± 1.30s, mid = 8.49 ± 1.41s, ∆14.42% *P* = <0.001, *d* = 0.96) and pre- to post- training (pre = 9.92 ± 1.30s, post = 7.61 ± 1.19s,

Δ23.29%, P = 0.017, d =1.85) for the HI group, while no change over time was detected for LIBFR on STS^T performance (Figure 6.6). No significant between group difference was detected at mid- and post-training time points. However, STS^T was greater in the HI compared to the LIBFR group at the pre-training time point (P = 0.043, d = 0.72).

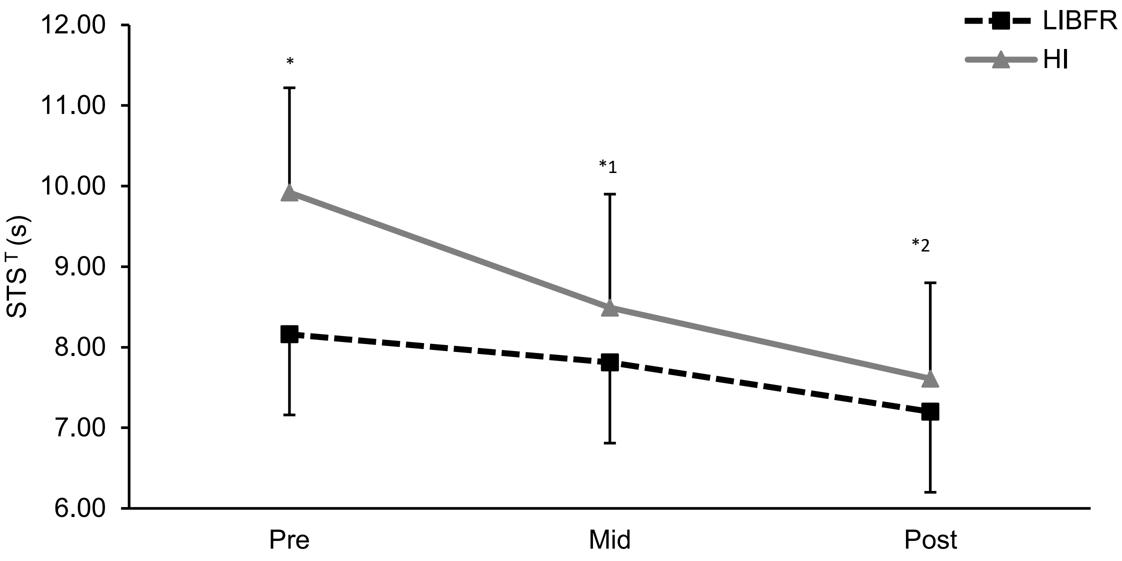

Figure 6.6: Mean ± SD in sit to stand - time (STS^T) following 6 weeks of LIBFR and HI exercise. *Significant between group difference at pre- testing point (HI and LIBFR). *1Significant difference from pre- to mid-training testing points for HI. *2Significant difference from pre- to post-training testing points (P ≤0.05) for HI.

6.3.2.3. STS^P

For STS^P, no significant interaction effect ($F_{(2, 28)}$ =1.792, P = 0.190, n_p^2 = 0.112), but a main effect of time ($F_{(2, 28)}$ =10.590, P < 0.001, n_p^2 = 0.431) but not of group ($F_{(1, 14)}$ = 0.572, P = 0.462, n_p^2 = 0.039) was detected. The *post hoc* pairwise comparisons revealed a significant increase in STS^P from pre- to post-training (collapsed data: pre = 209.14 ± 127.10W, post = 261.40 ± 171.20W, P = 0.006, d = 0.34) and mid- to post-training (collapsed data: mid = 228.60 ± 133.62W, post = 261.40 ± 171.20W, P = 0.022, d = 0.21) timepoints (Figure 6.7).

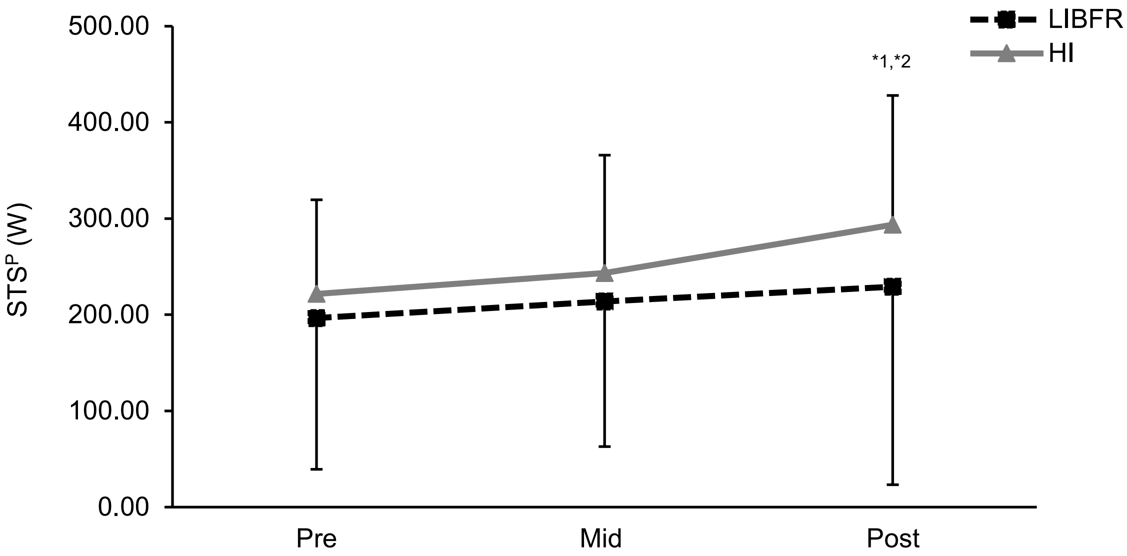

Figure 6.7: Sit to stand – Power (STSP) response after 6-weeks of LIBFR and HI exercise. *1Significant difference from pre- to post-training testing points (P ≤0.05). *2Significant difference from mid- to post-training testing points (P ≤0.05).

6.3.2.4. Gait Stability

For stride time, no significant interaction effect ($F_{(2, 28)}$ = 0.958, P = 0.390, n_p^2 = 0.064), or main effects of time ($F_{(2, 28)}$ = 3.184, P = 0.062, n_p^2 = 0.185) or group ($F_{(1, 14)}$ = 1.291, P = 0.275, n_p^2 = 0.084) were detected. For stride length, no significant interaction effect ($F_{(2, 28)}$ = 2.551, P = 0.156, n_p^2 = 0.124), or main effects of time ($F_{(2, 28)}$ = 3.659, P = 0.390, n_p^2 = 0.188) or group ($F_{(1, 14)}$ = 1.069, P = 0.977, n_p^2 = 0.059) were detected. For double limb support, no significant interaction effect ($F_{(2, 28)}$ = 0.275, P = 0.759, n_p^2 = 0.019), or main effects of time ($F_{(2, 28)}$ = 0.583, P = 0.565, n_p^2 = 0.040) or group ($F_{(1, 14)}$ = 1.711, P = 0.212, n_p^2 = 0.109) were detected (Table 6.2).

6.3.3. Strength

6.3.3.1. Squat Exercise

For the squat exercise (Figure 6.8), no significant interaction effect ($F_{(2, 28)}$ = 0.196, *P* = 0.823, n_p^2 = 0.014), but a main effect of time ($F_{(2, 28)}$ = 40.447, *P* = 0.001, n_p^2 = 0.743) but not of group ($F_{(1, 14)}$ = 0.886, *P* = 0.362, n_p^2 = 0.060) were detected. The *post hoc* pairwise comparisons revealed a significant increase in strength from pre- to post-training (collapsed data: pre = 39.31 ± 13.74kg, post = 51.25 ± 15.75kg, *P* = <0.001, *d* = 0.80).

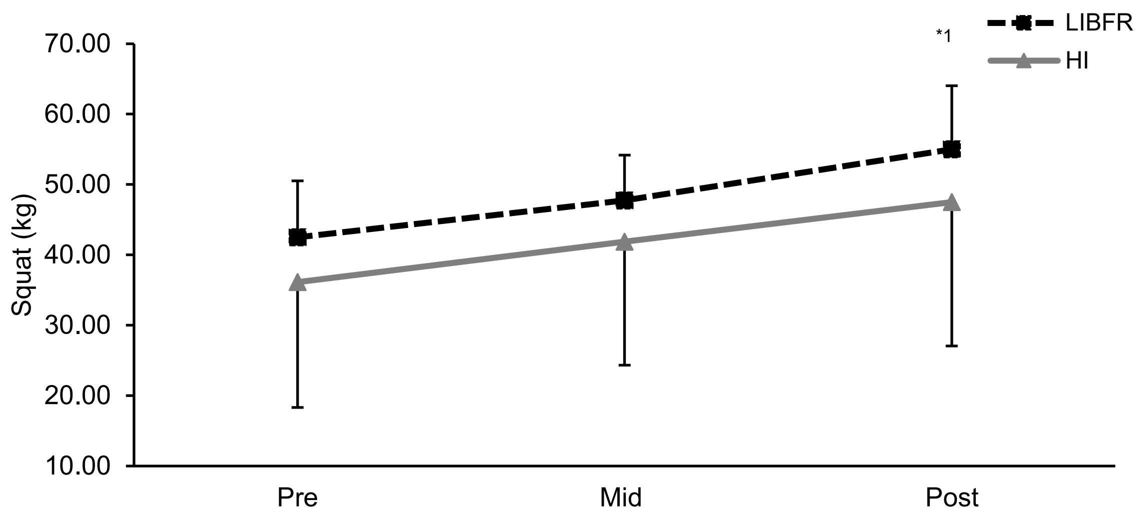

Figure 6.8: Mean ± SD in squat strength following 6 weeks of LIBFR and HI exercise. *1Significant difference from pre- to post-training testing points (*P* ≤0.05).

6.3.3.2. Calf Exercise

For calf exercise, no significant interaction effect ($F_{(2, 28)}$ = 0.520, *P* = 0.600, n_p^2 = 0.036), but a main effect of time ($F_{(2, 28)}$ = 30.095, *P* = <0.001, n_p^2 = 0.683) but not of group ($F_{(1, 14)}$ = 0.372, *P* = 0.552, n_p^2 = 0.026) were detected (Figure 6.9). The *post hoc* pairwise comparisons revealed a significant increase in strength from pre- to mid-training (collapsed data: pre = 38.57 ± 14.48kg, mid = 50.38 ± 15.82kg, *P* = <0.001, *d* = 0.78), mid- to post-training (collapsed data: mid = 50.38 ± 15.82kg, post = 57.00 ± 19.58kg,

$P < 0.001$, $d = 0.37$) and pre- to post-training time points (collapsed data: pre = 38.57 ± 14.48kg, post = 57.00 ± 19.58kg, P = <0.001, d = 1.07).

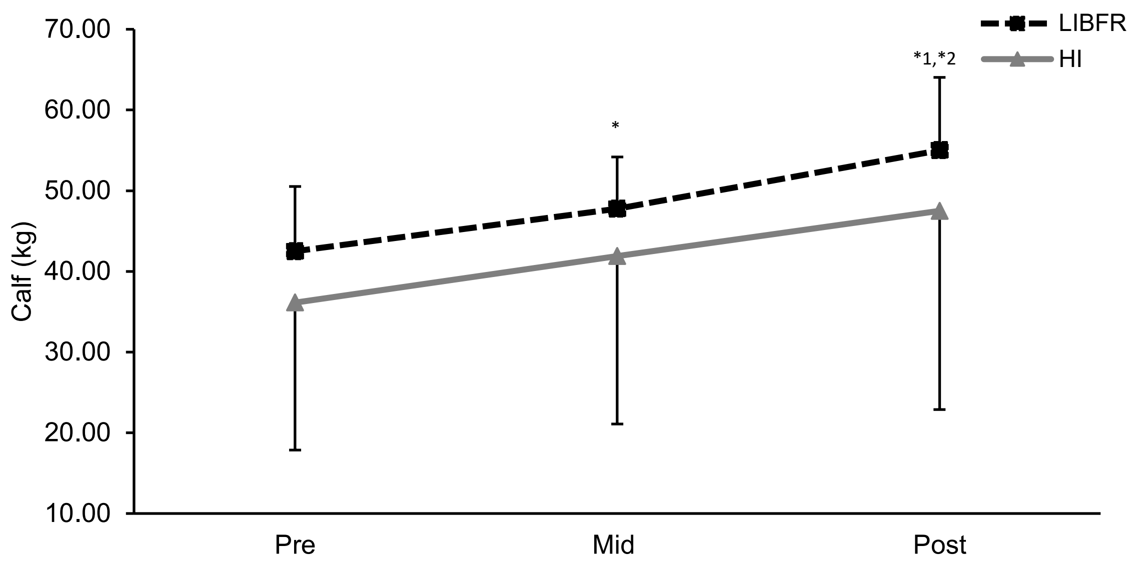

Figure 6.9: Mean ± SD in calf strength following 6 weeks of LIBFR and HI exercise. *Significant difference from pre- to mid- testing points ($P \leq 0.05$). *1Significant difference from pre- to post-training testing points ($P \leq 0.05$). *2Significant difference from mid- to post-training testing points ($P \leq 0.05$).

6.3.4. Neuromuscular and Proprioception

6.3.4.1. Maximal Voluntary Isometric Contraction – Knee Extension

For knee extension, no significant interaction effect ($F_{(2, 28)} = 0.303$, $P = 0.663$, $n_p^2 = 0.021$), but a main effect of time ($F_{(2, 28)} = 9.037$, $P = 0.004$, $n_p^2 = 0.392$) but not of group ($F_{(1, 14)} = 0.053$, $P = 0.821$, $n_p^2 = 0.004$) were detected. The *post hoc* pairwise comparisons revealed a significant increase in strength from mid- to post-training (collapsed data: mid = 138.30 ± 40.31Nm, post = 149.65 ± 41.60Nm, $P = 0.007$, $d = 0.28$) (Figure 6.10).

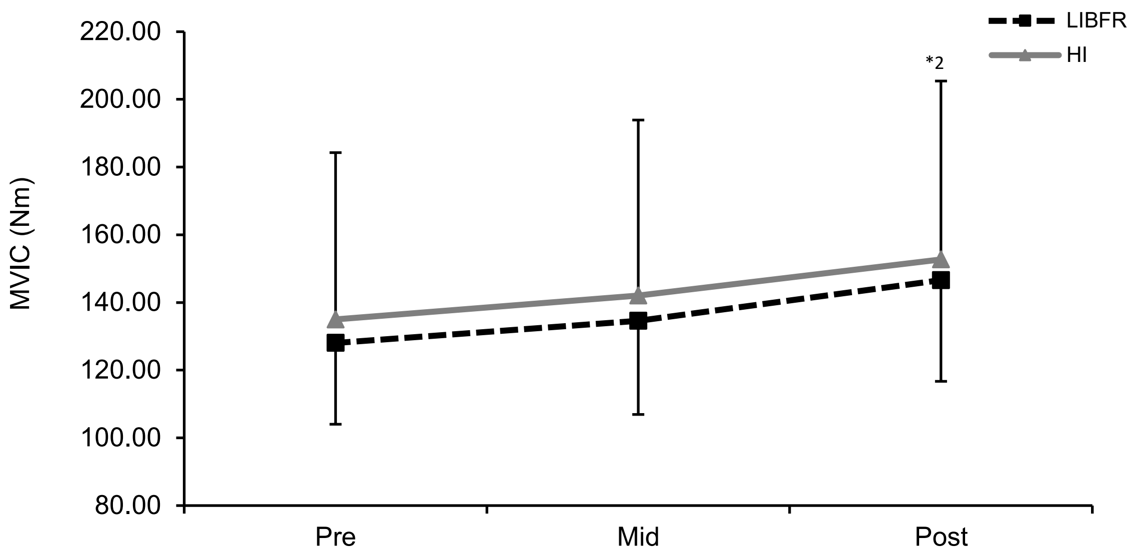

Figure 6.10: Mean ± SD in Maximal Voluntary Isometric Contraction (MVIC) following 6 weeks of LIBFR and HI exercise. *2Significant difference from mid- to post-training testing points ($P \leq 0.05$).

6.3.4.2. Force Sense

For force sense at 10% MVIC (FS10), a significant interaction effect was detected ($F_{(2, 28)} = 5.113$, $P = 0.014$, $n_p^2 = 0.268$). The simple main effects analyses revealed that the HI group were worse at force matching (increase in error rate) from pre- to mid-training time points (pre = 2.36 ± 1.73Nm, mid = 4.83 ± 2.59Nm, ∆104.66%, $P = 0.027$, $d = 1.12$) (Figure 6.11). No significant changes were detected in the LIBFR group. No significant between group difference was detected at the mid- and post- training time points. However, FS10 was greater in the LIBFR compared HI group at the pre- training time point ($P = 0.039$, $d = 1.13$).

For force sense at 20% MVIC (FS20), no significant interaction effect ($F_{(2, 28)} = 0.920$, $P = 0.410$, $n_p^2 = 0.062$), but a main effect of time ($F_{(2, 28)} = 4.109$, $P = 0.044$, $n_p^2 = 0.227$) but not of group ($F_{(1, 14)} = 0.872$, $P = 0.795$, $n_p^2 = 0.005$) were detected. The *post hoc* pairwise comparisons revealed an improvement in force matching from pre- to post-training testing points (collapsed data: pre = 4.13 ± 2.75Nm, post = 3.22 ± 1.53Nm, $P = 0.025$, $d = 0.41$) (Figure 6.12).

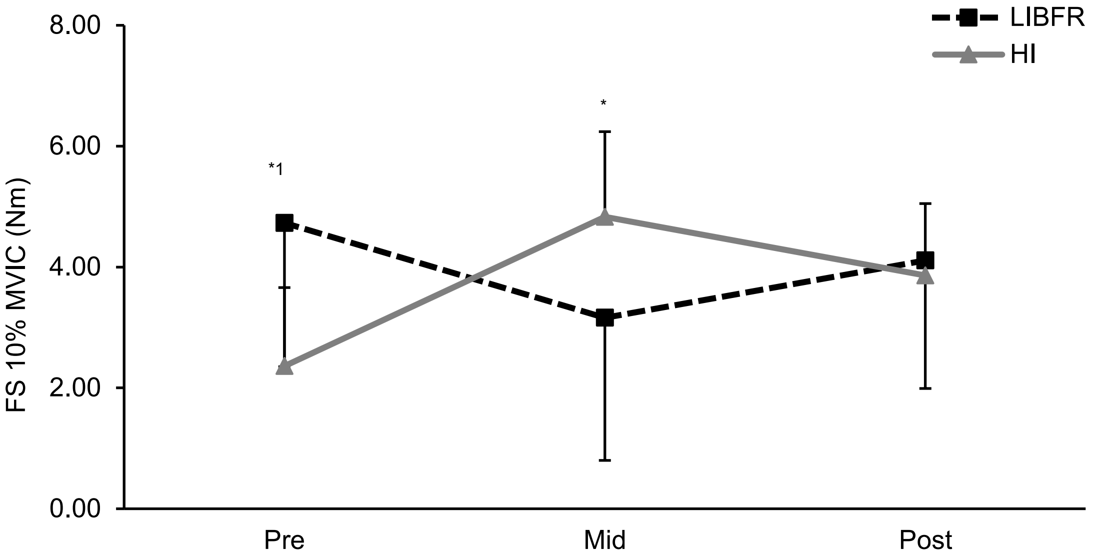

Figure 6.11: Mean ± SD in force sense (10% MVIC) following 6 weeks of LIBFR and HI resistance training. *Significant interaction effect pre- to mid-training testing points for the HI ($P \leq 0.05$). *1Significant between group difference at pre-training ($P \leq 0.05$).

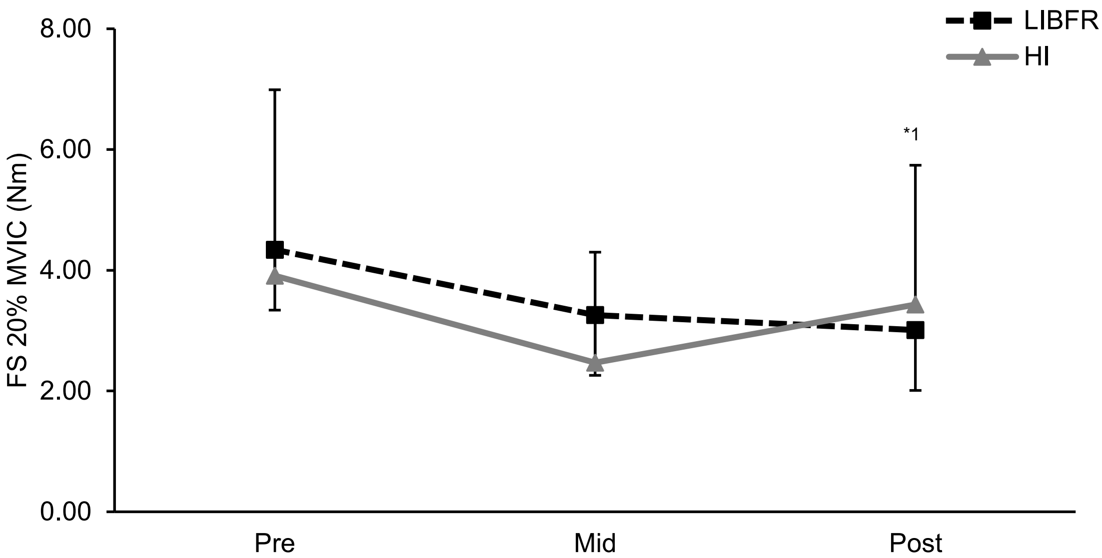

Figure 6.12: Mean ± SD in force sense (20% MVIC) following 6 weeks of LIBFR and HI resistance training. *1Significant difference from pre- to post- training testing points ($P \leq 0.05$).

6.3.4.3. Joint Position Sense: Extension

For JPS during knee extension performed to 70° range of motion (JPS Ext 70°) (Figure 6.13a), no significant interaction effect ($F_{(2, 28)}$ = 0.524, *P* = 0.587, n_p^2 = 0.036), or main effects of time ($F_{(2, 28)}$ = 0.662, *P* = 0.516, n_p^2 = 0.045) or group ($F_{(1, 14)}$ = 0.389, *P* = 0.543, n_p^2 = 0.027) were detected. For JPS during knee extension performed at 45° (JPS Ext 45°) (Figure 6.13b), no significant interaction effect ($F_{(2, 28)}$ = 1.145, *P* = 0.325, n_p^2 = 0.076), but a main effect of time ($F_{(2, 28)}$ = 7.854, *P* = 0.002, n_p^2 = 0.359) but not of group ($F_{(1, 14)}$ = 0.543, *P* = 0.473, n_p^2 = 0.037) were detected. The *post hoc* pairwise comparisons revealed a significant improvement in JPS Ext 45° from pre- to mid-training (collapsed data: pre = 3.77 ± 1.92°, mid = 2.21 ± 1.07°, *P* = 0.030, *d* = 1.00) and pre- to post-training time points (collapsed data: pre = 3.77 ± 1.92°, post = 2.67 ± 0.90°, *P* = 0.046, *d* = 0.73).

6.3.4.4. Joint Position Sense: Flexion

For JPS during knee flexion performed to 70° range of motion (JPS Flex 70°) (Figure 6.13c), no significant interaction effect ($F_{(2, 28)}$ = 0.214, *P* = 0.808. n_p^2 = 0.015), but main effects of time ($F_{(2, 28)}$ = 8.566, *P* = 0.001, n_p^2 = 0.380) and group ($F_{(1, 14)}$ = 10.329, *P* = 0.006, n_p^2 = 0.425) were detected. The *post hoc* pairwise comparisons revealed a significant improvement in JPS Flex 70° from pre- to mid-training (collapsed data: pre = 5.00 ± 1.97°, mid = 3.10 ± 0.94°, *P* = 0.020, *d* = 1.23). The *post hoc* pairwise comparisons identified the LIBFR group were better at the repositioning task at 70° knee flexion than the HI group (collapsed data: LIBFR = 3.26 ± 0.93°, HI = 4.07 ± 1.50°, *P* = 0.021, *d* = 0.65). For knee flexion performed to 45° range of motion (JPS Flex 45°) (Figure 6.13d), no significant interaction effect ($F_{(2, 28)}$ =1.090, *P* = 0.345, n_p^2 = 0.072), but main effects of time ($F_{(2, 28)}$ = 4.130, *P* = 0.032, n_p^2 = 0.228) and group ($F_{(1, 14)}$ = 6.509, *P* = 0.023, n_p^2 = 0.317) were detected. The *post hoc* pairwise comparisons revealed a significant improvement in JPS Flex 45° pre- to post-training (collapsed data: pre = 4.29 ± 2.07°, post = 2.79 ± 1.69°, *P* = 0.048, *d* = 0.79). The post hoc pairwise comparisons for group identified the LIBFR group were better at the repositioning task at 45° knee flexion than the HI group (collapsed data: LIBFR = 2.94 ± 0.93°, HI = 4.40 ± 2.34°, *P* = 0.023, *d* = 0.82).

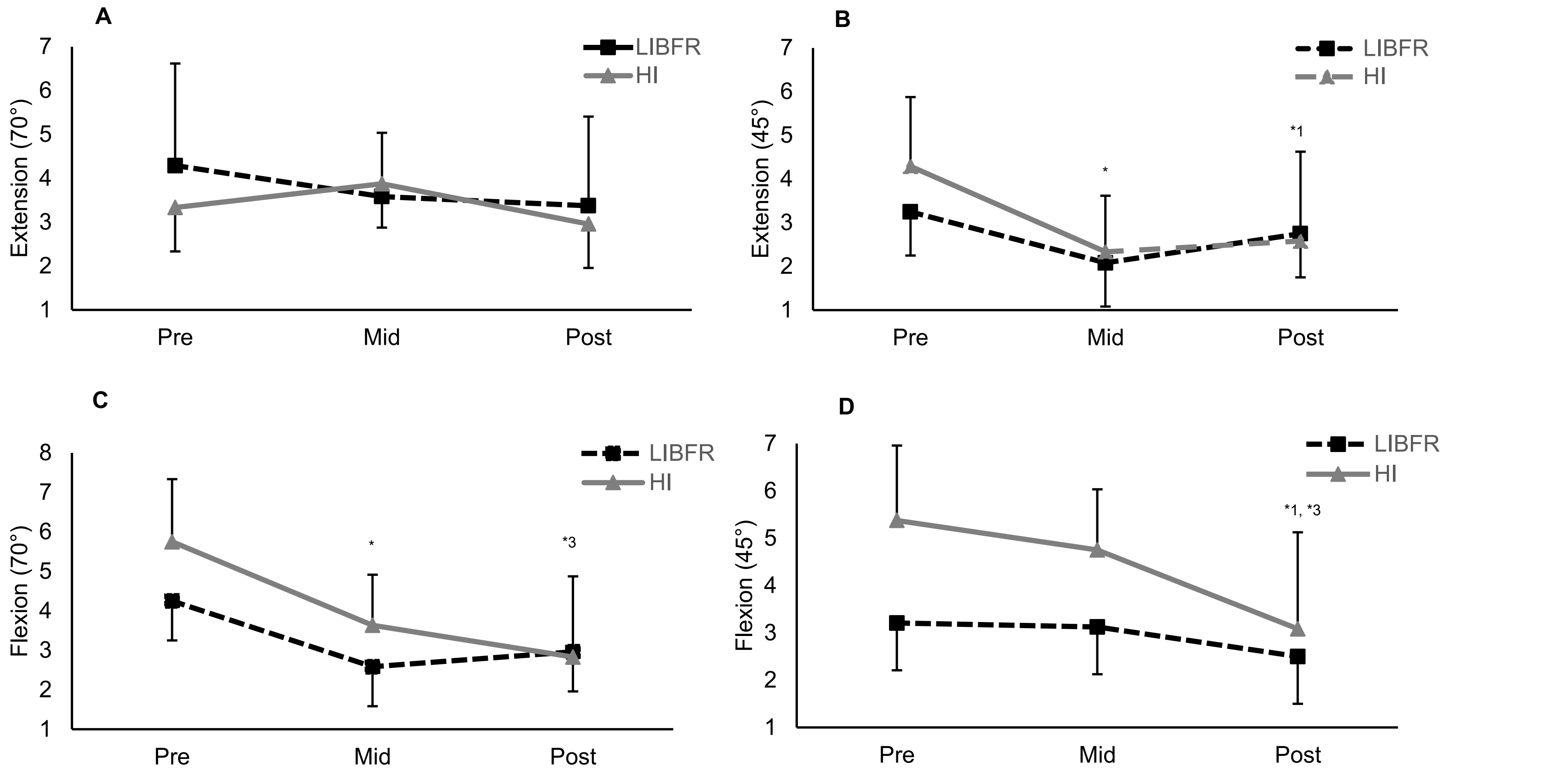

Figure 6.13 Joint Position Sense (JPS) responses after 6 weeks of LIBFR and HI exercise for extension (A, B) and flexion (C, D). *Significant difference from pre- to mid- training testing points (*P* ≤0.05). *1Significant difference from pre- to post- training testing points (*P* ≤0.05). *2Significant difference from mid- to post- training testing points (*P* ≤0.05). *3Significant difference between group (LIBFR vs. HI) (*P* ≤0.05).

6.3.5. Discomfort Scale

For $CR10_{Squat}$ (Table 6.3), no significant interaction effect ($F_{(1, 14)}$ = 0.747, *P* = 0.402, n_p^2 = 0.051), with a main effect of time ($F_{(1, 14)}$ = 4.667, *P* = 0.049, n_p^2 = 0.250) but not of group ($F_{(1, 14)}$ =3.518, *P* = 0.082, n_p^2 = 0.201) were detected. The *post hoc* pairwise comparisons revealed an improvement in discomfort from pre- to post-training (collapsed data: pre = 4.42 ± 1.44, post = 4.27 ± 0.93, *P* = 0.040, *d* = 0.12). For $CR10_{Calf}$, no significant interaction effect ($F_{(1, 14)}$ = 0.381, *P* = 0.547, n_p^2 = 0.026) or main effect of time ($F_{(1, 14)}$ = 0.095, *P* = 0.762, n_p^2 = 0.007), but a significant effect of group ($F_{(1, 14)}$ = 12.187, *P* = 0.004, n_p^2 = 0.465) were detected. The *post hoc* pairwise comparisons revealed a significant group difference between LIBFR and HI, with the LIBFR displaying a higher level of discomfort compared to HI (collapsed data: LIBFR = 3.34 ± 0.56, HI = 2.97 ± 0.92, *P* = 0.004, *d* = 0.49).

6.3.6. Perceptual Measures

For enjoyment (Table 6.3), no significant interaction effect ($F_{(1, 14)}$ = 2.226, *P* = 0.158, n_p^2 = 0.137), or main effects of time ($F_{(1, 14)}$ = 0.999, *P* = 0.335, n_p^2 = 0.067) or group ($F_{(1, 14)}$ = 0.006, *P* = 0.941, n_p^2 = 0.190) were detected. For perceived effectiveness, no significant interaction ($F_{(1, 14)}$ = 0.012, *P* = 0.913, n_p^2 = 0.001), or main effects of time ($F_{(1, 14)}$ = 0.050, *P* = 0.826, n_p^2 = 0.004) or group ($F_{(1, 14)}$ = 0.998, *P* = 0.335, n_p^2 = 0.067) were detected. For perception of load, no significant interaction ($F_{(1, 14)}$ = 3.119, *P* = 0.099, n_p^2 = 0.182), or main effects of time ($F_{(1, 14)}$ = 0.660, *P* = 0.430, n_p^2 = 0.045) or group ($F_{(1, 14)}$ = 0.916, *P* = 0.335, n_p^2 = 0.061) were detected. Similarly for fatigue, no significant interaction ($F_{(1, 14)}$ = 1.275, *P* = 0.278, n_p^2 = 0.083), or main effects of time ($F_{(1, 14)}$ = 0.006, *P* = 0.940, n_p^2 = 0.000) or group ($F_{(1, 14)}$ = 0.748, *P* = 0.402, n_p^2 = 0.051) were detected.

Table 6.3: Mean ± SD of subjective measures after 6-weeks of LIBFR and HI resistance training.

	LIBFR		HI	
	Pre	**Post**	**Pre**	**Post**
CR10				
Exercise 1: Squat	4.63 ± 1.65*1	3.84 ± 0.79	4.22 ± 1.28	4.69 ± 0.91
Exercise 2: Calf	3.44 ± 0.66*2	3.25 ± 0.46	3.00 ± 1.07	2.94 ± 0.82
VAS				
Item 1: Enjoyment	83.24 ± 9.22	83.38 ± 7.06	78.88 ± 23.53	87.79 ± 11.21
Item 2: Perceived Effectiveness	81.91 ± 11.16	81.62 ± 9.76	87.94 ± 12.01	87.06 ± 16.53
Item 3: Perception of Load	71.03 ± 10.85	67.06 ± 15.52	70.88 ± 22.21	81.62 ± 18.23
Item 4: Fatigue	47.06 ± 22.60	41.03 ± 23.01	50.59 ± 22.57	57.50 ± 33.62

Key: CR10 = Rate of Discomfort - CR10, VAS = Visual Analogue Score.
*1Significant difference from pre- to post- training ($P \leq 0.05$).
*2Significant group difference ($P \leq 0.05$).

6.4. Discussion

This study explored the effects of LIBFR and HI resistance training on quiet standing balance and gait stability in older adults, and attempted to identify the neuromuscular (muscle strength, muscle coordination patterns, JPS, and FS) mechanisms driving any improvements in postural control and gait characteristics. The main findings from this study were as follows: (1) both LIBFR and HI groups elicited similar increases in muscular strength, maximal voluntary isometric contraction, (2); with improvements in proprioception for JPS and FS, and functional performance for STS^P, STS^T and TUG (3); while there was no change in postural sway or gait stability apart from an increase in postural sway amplitude (COP_{ML}) from pre to mid-training; (4) there was a decrease in CR10 score (pre- [1st session] to post-training [12th session]) for squat and higher CR10 scores for the calf in the LIBFR compared to HI group; (5) there was no significant difference observed for VAS for both LIBFR and HI groups post-exercise. Overall, the findings indicate that both LIBFR and HI exercise were effective at increasing characteristics of strength and proprioception but did not improve determinants of postural sway and gait stability in older adults.

6.4.1. Centre of Pressure

There were minimal changes in postural sway following the 6-week training programme for the LIBFR and HI groups. The only change identified was at the pre- to mid-training testing points where an increase in postural sway amplitude (COP_{ML}) occurred for the DLS position. The increase in postural sway may be due to the presence of muscle fatigue after the first 3 weeks resistance training and possible development of overreaching (Halson, 2014; Hill *et al*., 2016; Fisher *et al*., 2017). Following exercise, residual neuromuscular fatigue can reduce the rate of force production and can affect subsequent sessions (Hill *et al*., 2016). However, in the present study the participants responded well to resistance exercise, which was reflected by the increase in strength (squat and calf exercise) and MVIC. Therefore, it could be possible that the disturbance in postural sway could be an early indication of overreaching despite the achievements gained in strength.

The results here indicate that resistance training alone may not be enough to improve postural sway. Muscle weakness has been associated with an increase in fall risk in older adults and there is a thought behind combining strength training with balance exercise to reduce fall risk (Orr *et al*., 2006; Pizzigalli *et al.,* 2011). In accordance with previous studies (Topp *et al*., 1996; Bellew *et al*., 2003) and reviews (Orr *et al*., 2008; Granacher *et al*., 2011; Lelard and Ahmaidi, 2015), there was no change in postural sway following resistance exercise. Despite the resistance training programme targeting the gastrocnemius, quadricep and hamstrings muscles responsible for postural control, the duration of the resistance programme (e.g., weeks) and number of sessions performed per week could have been a factor (Granacher *et al*., 2011). Indeed, as postural control is reliant on sensory and motor feedback components (Paillard, 2012) it is plausible that once an increase in strength has been achieved in the muscles responsible for postural control then further improvements may not be possible (Topp *et al*., 1996; Orr *et al*., 2008) unless the system is challenged (e.g., incorporating balance training). However, it is important to note that the lack of evidence of effect following blood flow restriction on postural control does not, however, preclude the possibility of a true effect. The low power of the analyses may have limited our ability to detect small but meaningful differences. In addition, a future recommendation could be including other forms of training to progressively challenge balance alongside LIBFR training on postural control.

6.4.2. Functional Measures

This study identified a significant improvement for the HI group for STS^T (pre- to post-training and mid- to post-training) but not for the LIBFR group. In addition, there were improvements in STS^P performance from pre- to post- and mid- to post-training time points. Previous resistance exercise programmes focusing on HI strength training in older adults have demonstrated improvements in STS^T (Schlicht *et al*., 2001; Nicholson *et al*., 2015). It was speculated that the improvement in the HI but not LI group may be explained by the higher external resistance (~70 1RM) performed by the HI group compared to the LIBFR group. The higher external resistance combined with the 'box squat' exercise replicating the STS movement may have provided a greater stimulus improving the rate of force development produced by the musculature of the thigh. A greater stress is applied through the thigh musculature at a slower and more controlled eccentric phase of the movement may have helped to develop isometric strength and increasing rate of force development under higher loads in this specific movement pattern. In addition, the descending phases of the squat movement allows the individual to position themselves in a greater hip flexion angle with the shins at a near vertical position. With a greater hip flexion angle achieved at the bottom of the box squat, the muscles responsible for hip extension are under more load and in turn generate greater hip power compared to the conventional squat (Flanagan *et al*., 2003; McBride *et al*., 2010). Consequently, the combination of a set depth and tempo (3-1-1) instructed while performing the box squat in the present study may have enhanced the development of isometric strength and rate of force development as previously discussed.

The findings from STS^T and STS^P data would suggest the strength gained from the squat exercise may have contributed to the improvement, but more importantly, development of power and faster times to complete the STS tasks. The improvement in STS^P is important, as greater power enables older individuals to react quickly from a perturbation by being able to maintain postural control through use of a reactive step strategy (Han and Yang, 2015). Although this study did not explore the effects of this type of training on reactive balance, the reduced time to complete STS^T and increase in STS^P may have improved the antigravity muscle of the lower extremity (e.g., ankle plantarflexors and knee extensors) that are vital for regaining and maintaining postural control (Paillard, 2012). Future studies should therefore investigate how LIBFR and HI resistance training can influence reactive balance control during sudden, unpredictable perturbations.

The findings from the TUG performance suggests that there was an improvement across two time points (pre- to mid-training and pre- to post-training testing points). Previous resistance exercise training programmes have demonstrated an increase in TUG performance after several weeks of strength training in older adults (Sousa and Sampaio, 2005; Kalapotharakos *et al*., 2010), while some authors have shown negligible improvements in TUG (Nicholson *et al*., 2015). The current study supports the former and could mean the improvements gained are related to the strength improvements from the squat and calf exercises, in addition to improvement in step initiation through greater neuromuscular control. The TUG assessment is known for being a strong predictor of not only mobility but also lower limb strength and core stability, which is closely related to dynamic balance and gait speed (Nicholson *et al*., 2015; Clarkson *et al*., 2017). In general, the TUG performance is worse in older adults who are physically inactive (Clarkson *et al*., 2017). However, the participants recruited for this study were active 'healthy' older adults with good pre-training scores. The improvements made over the 6-week period are important to note as the participants who already had a sufficient amount of mobility were still able to improve their TUG scores. Therefore, targeting the two exercises (calf and squat exercise) combined with a low frequency of training was successful in improving TUG performance. A future consideration would be to examine elderly adults to see whether the frequency of training and exercise would provide similar improvements in TUG observed in older adults.

There was no change in either double support, stride length and stride time at the mid- and post-training time points for either LIBFR or HI groups. It is well established that gait stability is a good predictor of fall risk (Hausdorff, 2007) and there is a relationship between postural sway and gait stability (Callisaya *et al*., 2010a). For example, the double support and stride length are both related to balance whilst walking (Hausdorff *et al*., 2001) and are associated to poor proprioceptive feedback if double support and stride length are below the threshold required for older adults (Brach *et al*., 2008; Callisaya *et al*., 2010b). Previous studies have reported improvements in double support and stride length after strengthening (Nouchi *et al*., 2014; Wang *et al*., 2015) and with lower load/ and intensity (Nicholson *et al*., 2015). In comparison to the current study, the programmes conducted by Nouchi *et al*. (2014) and Wang *et al*. (2015) were performed more frequently and were combined with balance tasks. This methodological difference may explain the null findings in the present study as each group performed only two sessions per week and did not include specific balance related tasks. Since older adults are more susceptible to falls but equally are more resistance to fatigue (Lanza *et al*., 2004), performing the walking tasks were either not challenging or should have been performed over a longer duration and may have provided a greater insight on gait stability. In addition, the older adults recruited for the study were “young” older adults and already had a good level of gait stability before starting the resistance programme. Another significant factor is that the study's low statistical power contributed to the observed null effects, potentially limiting the ability to identify small but meaningful differences.

6.4.3. Muscle Function

Preserving muscular strength in older adults is imperative as a reduction in strength is linked to a decrease in muscle function and is a well-known fall risk factor in this population (Vechin *et al*., 2015). The findings from this study provide an insight into the benefits of resistance training, highlighting its potential to enhance lower limb strength (calf and squat) using a short-term (6 week) training intervention with older adults. These results are comparable to other short-term (6-8 weeks; Yokokawa *et al.,* 2008; Patterson and Ferguson, 2011) and long-term (9 – 12 weeks; Karabulut *et al.,* 2010: Yasuda *et al.,* 2014; Yasuda *et al*., 2016; Vechin *et al.,* 2015; Bigdeli *et al.,* 2020; Linero *et al*., 2021) interventions when incorporating both functional (squat) and isolated (calf) movements in older adults. A possible explanation for the improvement in strength observed in the current study could be due to the participants recruited had minimal to no resistance training experience, which may have contributed to the marked increases in strength over a relatively short period of training (Gabriel, Kamen and Frost, 2006). Furthermore, studies who have accounted for untrained or novice older adults in their studies have identified an increase in leg extensor strength after short or long-term resistance training. For example, Harris *et al*. (2004) studied older adults (72.2 ± 5.1 years) performing moderate (~60% 1RM) to high (80% 1RM) intensity resistance exercise (2 x days per week) in novice older adults yielding an increase in knee extensor strength when reassessing the individuals 1RM. Similarly, Bigdeli *et al*. (2020) reported improvements in knee extensor strength after older adults (67.7 ± 5.8 years) performed regular (3 x days per week) low intensity (~25 – 35%) resistance exercise. The increase in strength demonstrated in the current study could be due to the volume (sets x repetitions) prescribed coupled with a sensible progressive training load. Adaptations to resistance training is dependent on such training variables to ensure the safe prescription (e.g., reduce the possibility of overtraining and improve adherence) and volume exercise, which can lead to improvement to the neuromuscular system (Grabriel, Kamen and Frost, 2006; Cannon and Marion, 2010). Moreover, it is well known that the neural adaptations after short-term resistance exercise can increase muscle co-activation, force sensation and motor unit firing rate (Knight and Kamen, 2001; Gabriel, Kamen and Frost, 2006), which may be a possible mechanism for the initial improvements observed in the current study.

6.4.4. MVIC

The 6-week resistance training programme elicited a significant increase in MVIC of the knee extensors with a significant change from mid- to post-training. Previous studies have reported an 11.2% (Cook *et al*., 2017), 13.7% (Yasuda *et al*., 2016), and 20.4% (Yokohama *et al*., 2008) increase in MVIC on the knee extensors after 8-12 weeks (2 sessions per week) of low intensity resistance training. Similar to muscle strength, there was a significant increase made in the mid- to post-training testing points for MVIC. The improvements in MVIC could be explained through the early neuromuscular adaptation, which may have led to an increase in motor unit firing rate (Knight and Kamen, 2001). Henneman's (1985) size principle of muscle unit recruitment implies that a submaximal contraction sustained for a period leads to the recruitment of higher order motoneurons (e.g., fast twitch muscle fibres). The main movement performed in this study was a box squat, where the eccentric (downward) portion of the lift was controlled before a brief amortization phase at the bottom. The challenging and slower eccentric phase of the movement combined with the 'touch and go' amortization phase may be the stimulus contributing to a greater level of peripheral fatigue and potentially stimulating fast twitch muscles. Van Roie *et al*., (2013) supports the notion, that an increase in neural drive occurs when older adults are working towards near maximal effort suggesting an increase in motor unit recruitment. However, this cannot be confirmed in the present study as sEMG was not conducted during MVIC performance and should be a future consideration. Nonetheless, the short-term effects of the resistance training programme improved MVIC in healthy older adults.

6.4.5. Force Sense

Proprioception is the ability to perceive the position of a joint and to detect force which is closely related to balance (Givioni *et al*., 2007; Windhorst, 2007). However, it remains unclear as to whether strength training alone has any benefit to improving proprioception (Smith *et al*., 2012). In the current study a significant improvement (pre- to post-training testing points) in FS20 was observed with no significant difference at any other timepoint. In addition, a significant interaction effect for FS10 demonstrated that the LIBFR group improved their ability to match force with less error within 3 weeks of LIBFR exercise compared to HI where the error rate was greater within 3 weeks. The LIBFR group were less accurate at FS10 at the end of the intervention compared to HI group who were unable to match pre-intervention scores. According to the literature, there has been no previous account on the effects of LIBFR on FS and whether this modality is capable in reducing the error rate. However, previous studies have reported no improvement in force sense after strength training with no BFR (Smith *et al*., 2012; Niespodziński *et al*., 2018). One possible reason for the reduction (improvement) in error rate in the LIBFR group for FS10, could be explained by the momentary pause (isometric contraction) at the bottom portion of the squat movement pattern. This part of the movement places a greater isometric muscular contraction of the quadricep and gluteal muscles before initiating the ascent phase. Givoni *et al*. (2007) has previously described that performing repeated isometric contractions places a greater demand on the sensorimotor system to match the force produced. Therefore, the improvements observed might be related to more efficient afferent feedback generated from the muscle to the central nervous system to sense tension and respond accordingly. In addition, part of the improvements made in FS10 may be related through an improvement in mechanoreceptors (e.g., Golgi tendon organs) (Missenard *et al*., 2009; Torres *et al*., 2010). The test position to assess FS was set at 70° knee extension as it reflects daily activities and has been said to stimulate Ruffini nerve endings and Golgi tendon organs (Torres *et al*., 2010). In particular, the Golgi tendon organs are responsible for sensing ‘muscle force’ and to protect the musculo-tendon complex from over stretching. It might be possible that during LIBFR, the higher repetitions performed during the squat and increase in hypoxia from LIBFR may have altered the function of the Golgi tendon organs to perceive force. This may have led to the LIBFR group being able to improve their accuracy during the reassessment FS when no BFR was applied to the thigh. Although speculative, we can deduce that the improvements made for the LIBFR group when force matching may have been through a greater recruitment of mechanoreceptors responsible for kinaesthesia.

6.4.6. Joint Position Sense

The absolute error rate for JPS while actively repositioning the limb back to the original start position is primarily controlled through the muscle spindles (Paschalis *et al*., 2007; Ribeiro *et al*., 2007b) and may contribute to improved postural control. The results obtained from the current study identified no improvement in JPS (Ext 70°) for either group. Although, an improvement for JPS (Ext 45°) was identified only in pre- to mid-training and pre- to post-training. The box squat exercise used for each training session in the LIBFR and HI groups was chosen because of its relevance to ADL and ability to promote quadricep/ hamstring strength gains through removal of the stretch shortening cycle observed in the conventional squat. The depth set for the box squat was approximately at a knee angle of 45° and may indirectly point towards some of the reasons an improvement in JPS were more profound in 45° compared to 70° range of motion. Riemann and Lephart, (2002) suggests mid joint ranges of motion (45°) maybe dependent on stimulation of muscle spindles compared to 70° which is a common knee range of motion angle for weight bearing exercise (gait pattern) recruiting the Ruffini nerve endings, Golgi tendon organs and Pacinian corpuscle. Therefore, the repeated box squat exercise at approximately 45° may have improved the participants ability to repeat the 45° angle at the mid- and post-training testing points more accurately because of the repeated specificity training effect. Indeed, the changes in strength observed over the 6-week training regime may have changed muscle activation and morphology of the muscle spindles, which supports the thought around the improvement and efficient intramuscular coordination.

6.4.7. Discomfort Scale and Perceptual Outcomes

The CR10 was used to assess the level of discomfort in both calf and squat exercises completed during each training session. A significant difference in time was observed (pre- to post-training) for the $CR10_{SQUAT}$, with no change to $CR10_{CALF.}$ However, the LIBFR group displayed a higher level of discomfort compared to the HI group for $CR10_{CALF}$. In accordance with the present study, CR10 tends to be higher with LIBFR exercise due to the application of the tourniquet to the thigh, which is not dependent on age, as both young (Wernbom *et al*., 2006; Karabulut *et al*., 2011; Loenneke *et al.,* 2011) and older adults (Clarkson *et al*., 2017; Harper *et* *al*., 2019) experience higher levels of CR10 with LIBFR. Although, some authors have reported no change in sessional CR10 in older adults (Yasuda *et al*., 2014; Scott *et al*., 2018). The current study agrees with the latter suggesting older adults can similarly tolerate HI and LIBFR which could be due to the LI of the exercises. Another potential reason for no difference in discomfort could be due to the cuff pressure not being high enough while training and the participants not feeling any discomfort whilst training (Brandner and Warmington, 2017). Regardless, the present findings suggest that performing either LIBFR or HI exercise was tolerable and produced similar changes in CR10 results for both squat and calf raise exercise.

The results from the four items assessed for VAS was unable to detect any significant interaction effect. The LIBFR group 'perception of load' and 'fatigue' reported at the end of each session was lower in comparison the HI group. Likewise, 'enjoyment' and 'perceived effectiveness' were equally as high for LIBFR and HI suggesting both types of exercises were as effective and enjoyable as each other.

6.4.8. Limitations

Although there were favourable strength and proprioception outcomes for both LIBFR and HI groups, there are some limitations to the study. Older adults were recruited who did not have previous resistance exercise experience but did already have a good level of mobility and physical function that may mitigate the potential to improve further (i.e. ceiling effect). Therefore, recruiting clinical participants in the future with impaired physical function or muscle strength should be considered. Additionally, there was a lack of a true (passive) control group to use as a comparison to the experimental groups that could impact internal validity and risk of bias. Furthermore, the training period was short and the potential for adaptation to training may have taken longer to achieve in the older group due to the good level of function prior to taking part in the training programme. Therefore, increasing the frequency of the sessions per week or total duration of weeks in the programme should be considered in the future. In addition, the detraining effects of each group were not observed after completing the training plan, which is an important area to consider as older adults can often undergo periods of non-physical activity where deterioration of physical function can accelerate. One important limitation of this study is the relatively small sample size. As previously mentioned in Chapters 4 and 5, the current findings have lower statistical power, meaning they should be interpreted as a lack of evidence for an effect rather than a definitive conclusion that no effect exists. The small sample size, along with the reduced sensitivity and high variability in the data, may have limited the ability to detect a true effect that could still be present but remain undetected under the current conditions.

6.4.9. Summary

Six weeks of LIBFR training elicited improvements in strength, neuromuscular, proprioception and functional measures in older adults. The low load used in combination with BFR could suggest that strength and neuromuscular adaptations can be achieved without using HI exercise. This study is the first to investigate the potential effects LIBFR may have on the determinants of quiet standing and gait stability as these measures are subject to change as humans age and is linked to fall risk (Meltzer *et al*., 2004; Hausdorff, 2007). Therefore, it was important to observe whether adaptations to either quiet standing or gait stability could be achieved through LIBFR.

The findings from this study might be of benefit to clinicians or rehabilitation specialists looking for an exercise alternative to achieve the strength benefits associated with HI exercise. There were improvements in strength and proprioception which is an important finding from the study as it demonstrates that there might be a potential improve movements associated to ADLs. The added benefit from this study alludes to the previous point made on ADLs, as exercises mimicking everyday functional movements (e.g., rising from a chair and sitting back down; STS) can be incorporated successfully into resistance exercise programmes. Another finding from this study was the psychological measures showing lower CR10 and VAS scores indicating that the type of training performed was less strenuous and more enjoyable. This is important to note as adherence can be a main factor influencing an individual’s decision to continue exercising if the exercise is difficult or strenuous. Therefore, the novelty and physiological effects resistance training provides could be an effective way to promote and encourage older adults to perform resistance training.

Chapter 7: General Discussion

The aim of this thesis was to explore the acute and chronic effects low intensity resistance training with blood flow restriction (LIBFR) has on quiet standing and gait stability in healthy young and older adults. A series of experimental studies were completed to achieve the overall aim of the study which were set by clear objectives (Chapter 1):

- Establish a healthy able-bodied model by examining the acute effects of lower limb muscle fatigue with blood flow restriction (BFR) and with no blood flow restriction (NBFR) exercise on quiet standing balance and gait stability in young adults.
- Explore the acute effects of lower limb muscle fatigue with BFR and NBFR exercise on quiet standing and gait stability in older healthy adults.
- To examine the effects of a chronic exposure to LIBFR and high intensity (HI) resistance exercise with no BFR in older adults on muscular strength, proprioception, postural sway, and gait stability.

This Chapter will discuss the main outcomes from the experimental studies conducted addressing the objectives mentioned in the previous.

7.1. Main Findings and Novel Contribution to the Literature

In this thesis, each experimental pilot study used a classic method to induce muscle fatigue to either the ankle or knee with BFR and NBFR (Chapter 4 and 5), followed by a 6-week resistance exercise programme comparing traditional HI to LIBFR (Chapter 6). The originality of this thesis is evident through the novel approach for each experimental pilot study where the concurrent application of a thigh tourniquet for LIBFR with fatiguing exercise (pilot study 3 and 4) and chronic exercise training (pilot study 5) was used to evaluate the changes in postural sway and gait stability. This work offers a fresh approach to resistance training in young and older adults by incorporating this modality into each pilot study. By integrating BFR into each pilot study, the aim was to broaden the understanding of BFR exercise and its potential influence on postural sway and gait in both young and older adults, exploring both immediate and long-term effects.

This study provides new insights into the effects of lower limb muscle fatigue on postural control using traditional methods to elicit fatigue are well reported in the literature (Yaggies and McGregor, 2002; Gribble and Hertel, 2004; Dickin and Doan, 2008; Orr *et al*., 2008; Bisson *et al*., 2012; Paillard, 2012). However, there is a shortage of information regarding the effects of lower limb muscle fatigue with BFR on postural sway and gait characteristics in young adults. Chapter 4 makes an original contribution to the literature, and to our knowledge, few studies have investigated these effects before using BFR. The muscle fatiguing trials to the ankle and knee were performed using the isokinetic dynamometry (mode: Isokinetic; muscle contraction - concentric/concentric) with BFR and NBFR. The test conditions were chosen because of the importance the ankle and knee are for the maintaining postural control.

In Chapter 4, young adults demonstrated a significant reduction in muscle torque production, independent of condition (i.e. BFR vs. NBFR) and muscle fatigue location (i.e. ankle vs. knee). This was a surprise as young adults have a high ratio of fast twitch muscle fibres (Evans and Lexell, 1995; Christie and Kamen, 2009), and capable of resisting fatigue over a longer duration (e.g., repetitions to failure). In addition to the reduction in muscle torque during the fatiguing protocol which was accompanied with a reduction in EMG_{TA} (ankle fatiguing protocol) and EMG_{VL} (knee fatiguing protocol). However, it was no surprise that a reduction in muscle torque may have accounted for the increase in postural sway amplitude (COP_{AP}) when performing the DLS stance under ankle and knee conditions. Following quiet standing tasks, the assessment of gait stability was performed. There was an increase in stride time immediately after ankle and knee fatiguing exercise, which returned to baseline measures 10 minutes post exercise with no change in stride length and double support. This result was contrary with the initial hypothesis as there was an expectation that following fatiguing exercise there would be a disturbance to gait stability. In this context, knowing that gait stability was unaffected after either ankle or knee fatiguing protocols was an encouraging finding before testing progressed to older adults (Callisaya, 2010a, 2010b).

The findings from Chapter 4 addresses a previously unexplored aspect of BFR exercise and was essential for our understanding of the modality on postural sway and gait stability, which served as an important marker from a methodological point to ensure the protocol was safe to use in older adults (Chapter 5).

Aging is characterised by several changes to the neuromuscular system (e.g., reduction in strength muscle and mass) which is important to maintain in older adults due to a close association with the demise in postural control (Riemann and Lephart, 2002a). The success of an older adult to perform repeated or prolonged tasks without developing muscle fatigue is crucial for the maintenance of postural control (Paillard, 2012). However, to date, no studies have explored how fatiguing exercise performed using the ankle or knee whilst undergoing BFR in older adults and addresses a clear gap in the literature. Therefore, in Chapter 5, the purpose of this study was to observe muscle fatigue and the recovery of postural sway and gait characteristics in older adults following the same procedures in young adults in Chapter 4.

In Chapter 5, older adults exhibited a significant reduction in muscle torque production (independent of BFR vs. NBFR condition and ankle vs. knee exercise) and was accompanied by a reduction in EMG_{TA} (ankle fatigue protocol) and EMG_{VM} (knee fatigue protocol). These findings are in line with the results observed in Chapter 4 and served as a proof of concept that both protocols were successful in reducing torque production and EMG amplitude, despite older adults being more fatigue resistant in comparison to young adults (Christie and Kamen, 2009). In addition, the type of contraction (concentric/concentric) performed on the isokinetic dynamometer is less likely to cause as much muscle fatigue compared to an eccentric muscle contraction (Nosaka and Newton, 2002). This may be of clinical importance when prescribing LIBFR in older adults where focussing on the concentric portion of an exercise is important (e.g., rising from a chair). For postural sway, there was a decrease in COP_{AP} (ankle and knee NBFR - SLS) and decrease in velocity (COP_{VL} - SLS) following ankle and knee muscle fatiguing trials, respectively. These findings were not consistent with Chapter 4, where there was an increase in COP_{AP} after the ankle and knee muscle fatiguing trials when performing the DLS balance task. Moreover, the results in Chapter 5 highlight how older adults adapted to the exercise and were able to improve their balance in more difficult stance positions (i.e. SLS) (Springer *et al.*, 2007; Bisson *et al.*, 2011). The reduction in muscle torque production and EMG_{TA} may have resulted in the decrease in COP_{AP} and could have been accelerated by an increase in motor response immediately post exercise when reperforming the SLS balance tasks. This may have occurred due to other age-related factors (e.g., muscle stiffness and co-contraction) which may have been responsible for the decrease in COP_{AP}.

There was no change observed in any measures of gait stability following ankle and knee fatigue (independent of BFR vs. NBFR conditions), which may have been a result of recovery post exercise.

Based on the findings reported in young (Chapter 4) and older (Chapter 5) adults, it was apparent that BFR and NBFR was effective at reducing muscle torque production, EMG amplitude and causing a slight disruption to postural sway in certain stance positions. The ankle and knee muscles targeted for the fatiguing protocol were used due to previous literature identifying a greater disturbance in postural sway following muscle fatiguing exercise. Moreover, the muscles targeted in the fatiguing protocol are not only important for the maintenance of postural sway but are also important for activities of daily living (ADL). While many studies find that muscle fatigue impairs balance performance, there are several possible reasons why some studies might find null effects of fatigue on postural control:

1. The body can adapt to fatigue by using different muscles or motor strategies to maintain balance, which can mask the effects of fatigue on postural control. Some balance tasks may not be challenging enough to reveal impairments caused by fatigue.
2. Some balance tasks may not be challenging enough to reveal impairments caused by fatigue.
3. The method used to induce fatigue (e.g., type of exercise, duration, and intensity) can result in varying levels of fatigue and can affect balance differently.
4. Fatigue localised to specific muscle groups might not affect balance as much as generalized fatigue.
5. The assessments used might not have been sensitive enough to detect subtle changes in balance performance.
6. Participants might improve their balance performance through practice or repeated testing, which can counteract the effects of fatigue.

Examining the longitudinal effects of LIBFR compared to HI resistance training on postural sway and gait stability was an important next step to answer the "*so what*" question. Whilst acute fatigue studies limited our ability to examine certain outcome measures due to short term time course events, a chronic study allows the measurement of a broader range of outcomes with greater relevance and application to real world settings (e.g., ADLs), as well as more mechanistic outcomes (e.g., force sense, joint position sense) providing a deeper understanding of the interventions impact. This is due to the extended time frame of a chronic study permits the assessments to be undertaken during resting conditions providing an opportunity to comprehensibly assess a range of assessment measures.

Preserving muscle strength is essential for older adults to maintain as it is closely linked to ADL but also enables the individual to continue living independently through preventing the onset of age-related conditions such as sarcopenia. The original aspect of Chapter 6 was to evaluate the effects of LIBFR and HI resistance exercise on quiet standing, gait stability and neuromuscular mechanisms in older adults. This study identified improvement in strength, maximal voluntary isometric contraction, joint position, and force sense. The changes observed in strength and maximal voluntary contraction is an example of how performing resistance-based exercise (independent of LIBFR or HI) can develop strength in a relatively short time frame in older adults. There is a suggestion that neurological adaptions occurred from the improvements identified in joint position and force sense tasks (Paschalis *et al*., 2007; Ribeiro *et al*., 2007b). However, minimal improvements in postural sway and gait characteristics were identified despite observing an increase (pre - mid testing point) in postural sway amplitude (COP_{ML}). The likely reason for the increase in COP_{ML} is overreaching in which transient neuromuscular fatigue emanating from previous training sessions may have influenced balance performance. For the functional performance measures, an improvement in STS^{T} for the HI group was observed with no change in for the LIBFR group. In addition, there were improvement in STS^{P} from pre- to post- and mid- to post- training time points. The improvements made in strength and MVC may have contributed to the improvements identified from the STS^{T} and STS^{P} performances. However, the LIBFR group improved FS10 within 3 weeks of training suggesting there may have been a better afferent feedback loop was developed (Torres *et al*., 2010). The discomfort scale performance results revealed an improvement in $CR10_{Squat}$ (pre: 1st session – post: 12th session), while a group difference was observed between LIBFR and HI for $CR10_{Calf}$ where the LIBFR group reported a higher level of discomfort compared to the HI group. The overall findings here provided a valuable insight in LIBFR training suggesting the older adults were able to tolerate either LIBFR or HI exercise, which is essential for adherence to exercise in this population but did not have an adverse effect on balance performance.

In older adults there are still many questions either unanswered or have not been debated as to how LIBFR can be incorporated successfully in this population. Several review studies have eagerly debated and questioned the methodological and efficacy around LIBFR in older adults with a level of scepticism. Based on the findings from Chapter 6, older adults who are 'healthy' and relatively active can complete LIBFR in a safe and controlled environment. Overall, the findings indicate that LIBFR was effective at increasing characteristics of strength and proprioception but did not improve determinants of postural sway and gait characteristics in older adults.

7.2. Limitations

There were several limitations in this thesis which should be acknowledged. The outcome of this section is to provide practical application and future considerations for BFR in older adults and where the next possible steps would be for the use of this modality.

7.2.1. Sample

An area for consideration is the population used throughout the thesis was at the lower end of the 'older adult' age category (60+ yrs) and were all generally fit and healthy individuals. This may have impacted the results gathered from the quiet standing and dynamic postural sway data collected in Chapters 4, 5 and 6, as postural sway was already close to a *physiological minimum*, to begin with. From the acute pilot studies, both young (Chapter 4) and older (Chapter 5) adults, experienced minimal changes to quiet standing and dynamic postural sway immediately after ankle or knee fatigue with BFR. In addition, this may also explain why minimal improvements, or no change occurred in older adults during the training study (Chapter 6). Previous investigations on LIBFR in older adults have mainly recruited participants who are healthy and have not suffered a fall and may explain why negligible changes to postural sway occurred in the current studies.

However, as the studies conducted in this thesis were experimental, it was important to recruit healthy young and older adults mimicking previous studies in BFR because of the unknown effect LIBFR may have on balance. Therefore, a consideration for the future would be including participants who have experienced a fall and implementing LIBFR in this category to see whether an improvement balance along with strength adaptations could be gained.

Although a series of studies were presented in this thesis as pilot experiments, the insights gained from these studies were valuable for informing the design and implementation of larger experimental studies. However, it is important to acknowledge the small sample size of the pilot investigations. The small sample size of the pilot studies potentially limits the statistical power of the study (increasing the likelihood of a type II error), reducing the generalisability to the general population and increased variability (outliers having a disproportionate large impact on a study outcome). Larger, fully powered studies are recommended to validate the observations made in this thesis. The small sample size of the pilot studies in this thesis were significantly influenced by the COVID-19 pandemic and resource constraints, which impacted the studies in several ways. For example, COVID-19 imposed a restriction on face-to-face testing (due to social distancing measures), which presented significant recruitment challenges (i.e., reduced access to recruitment opportunities). Limited financial resources reduced the ability to conduct extensive recruitment. Collectively, these factors contributed to a smaller than anticipated sample size.

7.2.2. Postural Sway

Postural sway was assessed using quiet standing (e.g., DLS, SLS_L and SLS_R) and gait stability in each experimental study which was deemed robust and reliable and has been consistently used across the literature. Minimal changes in postural sway amplitude and velocity could have been due to the entry level of the participants, who have already displayed a reasonable ability to maintain balance. Therefore, more challenging balance tasks (e.g., EC) could have been considered in each study, which would have provided a good view of how COP was controlled through different stance positions. In addition to using more challenging stance positions, there is a cognitive aspect to maintaining postural control, which can be disturbed more when performing dual tasks under fatiguing conditions (i.e., performing non-postural task [reading] while standing). Therefore, performing a non-postural task during quiet standing has been shown to increase postural sway, especially in older adults who have a postural deficit (Prado *et al*., 2007). In this context, combining dual tasks and performing fatigue protocols may have more relevance to fall rates but may have provided a better insight following our fatiguing trials in 'healthy' participants who did not have a previous history of falls.

7.2.3. Blood Pressure

One limitation of pilot studies 3, 4 and 5 was the method used to assess and prescribe BFR pressure to the lower limbs. Lower limb BFR pressure was determined by measuring the upper arm resting SBP using a blood pressure monitor to determine lower limb BFR pressure. The validity of resting SBP derived from the upper arm to determine lower limb BFR pressures to the thigh has been used successfully in previous studies (Downs *et al*., 2014; Cook *et al*., 2017; Kjeldsen *et al*., 2019). However, several review articles have queried the accuracy and safety when prescribing lower limb BFR pressure using the previously described procedure. When prescribing lower limb BFR pressure, it is common to use a tourniquet on the upper thigh and a handheld doppler with the probe on the posterior tibial artery/pulse. The cuff is gradually inflated to a pressure where no blood flow can be heard through the doppler probe and is then termed arterial occlusion pressure (AOP). A percentage based on the individual AOP is then used to set a target BFR pressure to the lower limb. This procedure is widely accepted across the literature and is classed as a more accurate method to prescribe lower limb BFR. Despite this, currently, there is minimal research on lower limb AOP pressure in older adults but exists in young adults. Secondly, there is an issue around heterogeneity with AOP, which is related to the fluctuation of blood pressure during the day-by-day circadian rhythm. Older adults tend to have a greater rate of change to blood pressure which can be hard to control (e.g., hypertension), and structurally there can be issues with pulsatility and loss of elastic properties to arteries leading to pathophysiological changes to the circulatory system (e.g., arterial stiffness), and may explain why AOP can be heterogenous and more so in older adults. For each experimental pilot study in this thesis, all participants completing BFR trials attended each session at the same time of day for each visit to avoid fluctuating blood pressure. After completing pilot and experimental studies using the process of resting SBP was taken from the brachial artery of the participant's right arm, there appeared to be no detrimental effect on either young or older participants whilst completing BFR trials in each study when adopting this method. Previous research assessing resting SBP in the upper and lower limb identified minimal differences between the two measurements with a high correlation (Pascarelli *et al*., 1964; Hocken, 1967; Gerhard-Herman *et al*., 2006). Therefore, given the similarity between upper and lower limb resting SBP, this method was quick and easy to measure and prescribe lower limb resting SBP for BFR training in young and older adults. Future studies should focus on measuring lower limb AOP for prescribing BFR training in older adults, as this would help improve safety and understanding when measuring AOP in the lower limb and help to identify a 'sweet spot' as to what resting SBP could be used in the LIBFR programme for older adults.

7.3. Practical Application

As we age, our musculoskeletal health and function undergo changes that can significantly impact our well-being. Older adults are particularly vulnerable to the effects of declining muscle quality, strength, mass, and postural control. These changes can limit their ability to carry out ADL and increase the risk of falls. A simultaneous loss of muscle strength and mass can increase cardiometabolic diseases in older adults resulting in higher mortality rates. This highlights the importance to reduce the effects of aging through a considered approach towards sensible programming of resistance exercise. As HI resistance exercise is widely regarded as an effective strategy to reverse age-related changes in older adults, it is often contraindicated and not tolerated well in older adults. Therefore, alternative forms of LI exercise have become increasingly important as a substitute for HI resistance exercise in this population. It was apparent that LIBFR could benefit patients at risk of muscle atrophy and serve as a suitable alternative to HI resistance exercise in older adults. Clinicians and rehabilitation specialists may find LIBFR advantageous as an alternative to HI resistance exercise, given its potential for improving strength and neuromuscular measures. Therefore, if LIBFR is performed by patients safely in a controlled environment where the practitioner can monitor the patient's progress, then in the long term, this novel training modality may be an appropriate alternative to HI exercise which has the potential to reduce muscle atrophy associated with an age-related decline in the musculoskeletal system and could help enhance the quality of life in older adults.

Chapter 8: References

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Chapter 9: Appendix

9.1. A1: Ethics Approval

Dear Dominic

Please see the message from the hair of the Research Ethics Committee below:

I confirm that I have reviewed Dominic's resubmission and I am satisfied that the minor revisions have been successfully completed, as outlined in the covering letter. I am therefore happy to approve this as a Chair's action.

You may now proceed with your study and your APG registration will be confirmed at the next Research Degrees Committee meeting.

Best wishes

David

David Watson
Postgraduate Research Manager
The University of Northampton
Park Campus
Boughton Green Road
Northampton
NN2 7AL

THE UNIVERSITY OF NORTHAMPTON
Graduate School

RESEARCH DEGREES BOARD RECOMMENDATION FOR REGISTRATION AS AN ADVANCED POSTGRADUATE STUDENT

The Research Degrees Board, having reviewed the attached information, recommends to the Research Degrees Committee that the following student be registered for a research degree at The University of Northampton:

Student: Dominic Alexander Langdon

Signed ....
Chair, Research Degrees Board

Date ....

Confirmation (signatures required):

☐ The Research Ethics Committee has been consulted and advice given.

Signed, Chair of the Research Ethics Committee ....

☒ The research proposal had been appropriately and sufficiently risk assessed

Signed, Supervisor

☒ The student has completed the mandatory induction.

Signed, Research Training Co-ordinator

☒ The student has completed the mandatory online research ethics module.

Signed, Research Training Co-ordinator

Accompanying documents:

☒ **REG1** - Application for Registration form

☒ **REG2** - Details of Supervisory Team

☒ **REG3** - Resource requirements

☒ **REGTPW** - Postgraduate Training Plan of Work

☒ Project proposal

☒ Project Plan

Feedback from Research Ethics Committee	
Student: Dominic Langdon	Date: 22 September 2016

Action required	Tick
No action required	
Submit amendments for Chair's Action	✓
Submit amendments for consideration by members by email	
Resubmit application to future REC meeting	

Decision relating to the proposal	Tick
Full approval was given	
Advisory comments were given	
Amendments are required before full approval can be given	✓
Approval in principle was given	
Amendments are required before approval in principle can be given	
In its current form, approval could not be given	

Feedback on proposal

The submission was exemplary in its presentation and the researcher was thanked for such a thorough response. The Committee required three minor changes that could be approved quickly by Chair's action:

Changes from Dominic Langdon have been annotated and made in a blue coloured font (Also numbered)

- 1) The recruitment poster should focus on the research study and not on an offer of training. If the research is to determine the efficacy of the training it is not appropriate to advertise the training as effective.

Correction: The comments above have been taken into consideration and the recruitment poster has been adjusted accordingly to reflect the research study itself.

- 2) The image used on the poster was perhaps inappropriate to the study and target group, and may suggest unrealistic expectations of the training.

Correction: Image has been changed and is reflective of the area in which the blood flow restriction cuff would be applied to the lower limb.

- 3) Rather than offering participants the opportunity to withdraw at any time, careful consideration needs to be given to ensuring that the timing of the withdrawal (a) allows participants sufficient time to make a reasoned and reflective decision and (b) does not potentially undermine the integrity of the data-set (and ultimately the study itself) by being an open invitation.

Correction: Changes have been made in accordance to the comments above. All have been included to the ethics form and participation information sheets. Careful consideration has been made and included to all forms.

9.2. A2: Example Participant Information Form

Participant Information Sheet

The effects of combined acute muscle fatigue with and without blood flow restriction on postural control in healthy young adults

My name is Dominic Langdon, and I am a Doctor of Philosophy (PhD) student at the School of Health (SoH), The University of Northampton. As part of my PhD, I am looking for volunteers to take part in a research project. The findings from this research project would contribute towards my qualification of Doctor of Philosophy (PhD). Dr. Mathew Hill and Dr. Anthony Baross at the University of Northampton are supervising this study.

What is the purpose of the study?
The aim of the study is to see whether the use of a specially made blood pressure cuff applied around the thigh and inflated to 1.2 x your resting blood pressure influences your ability to maintain balance.

Why have I been chosen?
You have been asked to take part in the study because you meet the criteria of this research project and are physically healthy to participate.

Do I have to take part?
Taking part is entirely voluntary. If you do decide to take part, you will be given this information sheet to keep and may be asked to give your written consent. You will receive a copy of the signed consent form. If you decide to take part, you are free to withdraw at any time during data collection and up until the first week after data has been collected without giving a reason. Any information or responses you may have already given will be destroyed.

What will my participation involve?
Once you have agreed to take part in the study you will be required to attend 4 sessions at the Waterside laboratory at the University of Northampton. As a participant you will be required for the following intervention (Flow diagram below)

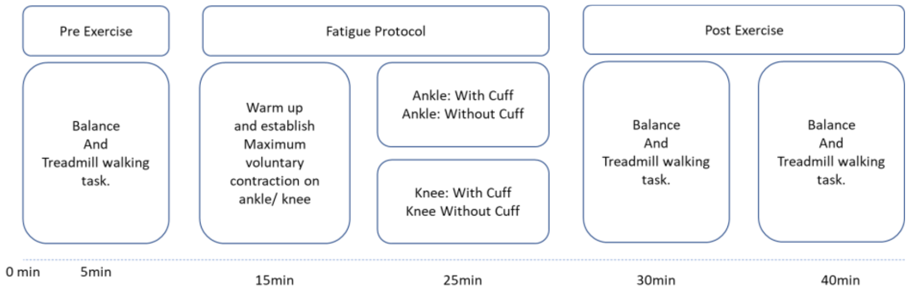
Day 1: This will include a run through of the equipment and the collection of resting systolic blood pressure. A blood pressure cuff will be placed around your right arm. You will then be required to remain supine position with eyes open, remaining quiet for a 10-minute rest period. Towards the end of the 10-minute rest period, resting systolic blood pressure will be taken using a digital blood monitor. Following this, you will be asked to perform two static balance tasks (double leg and single leg) and a 3-minute walk on the treadmill.

Once you have finished both pre-tests, you will then be set up on the dynamometer to perform a set of 3 maximum voluntary contraction to either the ankle or knee. The right leg will be set up for electromyography, where a sticky electrode pad will be placed on two muscles of the thigh and two muscles on the calf. This will be completed before the fatigue protocol takes place.

The blood pressure cuff will be placed bilaterally to the upper portions of the thigh before the completion of the fatigue protocol. The results from this maximum voluntary contraction test, would then be used for the fatigue protocol, where you will exercise until your maximum strength (maximum voluntary contraction results) has reduce by 50% of your original value (I.e. Your 100% maximum voluntary contraction may be 300 Newtons, therefore you would work repeatedly until your strength had reduced by 50% which would equate to 150 Newtons).

Immediately after the completion of the fatiguing protocol, you will perform the same static balance and treadmill walking task as you did at the start. Followed by another bout

Immediately after the completion of the fatiguing protocol, you will perform the same static balance and treadmill walking task as you did at the start. Followed by another bout of static and treadmill walking after 10 minutes. After the completion of these tasks you will perform a gentle cycle on an ergometer for 5 minutes to help with recovery after the fatigue protocols.



- Day 2, 3 and 4: The same protocol as previously mentioned will be completed.

What are the possible benefits of taking part?
The information gained from the study will provided a broader understanding to the neurological responses of the body when blood flow restriction is applied after a fatigue protocol. The information will provide a more thorough understanding as to what percentage of strength causes the most disturbances in balance and how this could be replicated in older adults where balance is important.

What are the possible risks or disadvantages of taking part?
By taking part in this study there will be possible risk of discomfort and altered sensation from wearing the blood pressure cuff for a period time. The sensation experienced will be temporary and may last the duration of the exercise trial. There has been special provision put in place to ensure that the experience described in the previous is minimised. But it is important you are aware of the risk that may be involved.

What if something goes wrong?
If you wish to complain, or have any concerns about any aspect of the way you have been approached or treated during this study, please contact the investigator on the contact details below

Will my taking part in this study be kept confidential?
All personal information and data collected during the study will be anonymised and kept confidential. Hard copies of the information collected from the participant will be stored in a secure filing cabinet under lock and key. Data will be kept on a password protected computer and will be kept through the timescale of investigator's PhD qualification. Access to the data collected will be restricted to the supervisor(s) mentioned in the previous and will not have direct access unless granted by the investigator. The data will be disseminated for either conference/ peer reviewed journals either during or after completion of the PhD.

What will happen to the results of the study?
The full results of the study will be available in my dissertation. If you wish to receive a summary report about the study's findings you can indicate this on your consent form.

Who is organising the study?
This research forms part of my Doctor of Philosophy Studies and has been organised by myself and supervisors.

Who has reviewed the study?
The study has been approved by the research ethics committee (REC) at the University of Northampton and the School of Health (SoH) ethics committee. A detailed formal research proposal was submitted along with the ethics proposal to ensure the protocol suggested is ethically sound and not invasive. This proposal was viewed by the academic postgraduate board

Contact for further information
If you have any questions about the research, you can contact me or my supervisor(s) as follows:

Investigator:	1st Supervisor:	External Supervisor:
Dominic Langdon	Dr. Anthony Baross	Dr. Mathew Hill
dominic.langdon@northampton.ac.uk	Baross.anthony@northampton.ac.uk	ab2225@coventry.ac.uk

9.3. A3: Physical Activity Readiness and Questionnaire Form (PAR-Q)

UNIVERSITY of NORTHAMPTON
Sport and Exercise Science
PRE-TEST MEDICAL QUESTIONNAIRE

Name: …….

Date of Birth: ……………………. Age: …………….. Sex:………………………….

Please answer the following questions by putting a circle around the appropriate response or filling in the blank space.

1. How would you describe your current level of physical activity?
   Sedentary / Moderately active / Active / Highly active

2. How would you describe your current level of fitness?
   Unfit / Moderately fit / Trained / Highly trained

3. How would you consider your present body weight?
   Underweight / Ideal / Slightly overweight / Very overweight

4. Smoking Habits:
   Are you currently a smoker? Yes / No
   Are you a former smoker? Yes / No
   How long is it since you stopped? ……Years ……Months
   Were/are you an occasional smoker? Yes / No
   …………per day
   Were/are you a regular smoker? Yes / No
   …………per day

5. Do you drink alcohol? Yes / No
   If you answered Yes, do you have:
   An occasional drink / A drink every day / More than one drink per day?

6. Have you had to consult your doctor within the last six months? Yes / No
   If you answered Yes, please give details

……
……………………………………………

7. Are you aware of any allergies? If so, please state Yes / No

……
……
..

8. Are you currently taking any form of medication? Yes / No
   If you answered Yes, please give details

……
……

9. As far as you are aware, do you suffer or have you ever suffered from:
a. Diabetes? Yes / No b. Asthma? Yes / No
c. Epilepsy? Yes / No d. Bronchitis? Yes / No
e. Any form of heart complaint? Yes / No f. Raynaud's Disease? Yes / No
g. Marfan's Syndrome? Yes / No h. Aneurysm or Embolism? Yes / No

10. Is there a history of heart disease or cardiovascular disease in your family?
Yes / No

11. Have you ever suffered from deep vein thrombosis or peripheral cardiovascular disease? Yes/ No

12. If answered 'Yes', how long ago did you have the condition?

……

13. Have you suffered or noticed any unusual swelling around any of the following joints in the past TWO WEEKS. Yes/ No

Joint:	Ankle	Knee	Hip
Duration:			

14. Have you seen or consulted a GP about the swelling? Yes/ No

15. Do you currently have any form of muscle or joint injury? Yes / No
16. Have you had to suspend your normal training in the last 2 weeks? Yes / No

17. As far as you are aware, is there anything that might prevent you from successfully completing the tests which have been outlined to you? Yes / No

18. Have you suffered any neuropathic pain in the past two weeks?
(Neuropathic pain can be consider pain results from nerves to either the peripheral and central nervous system)
Yes / No

19. Have you ever been diagnosed with neuropathic? Yes/ No

If answered 'Yes', please indicate the type of pain using the table below.

Type of pain	Head	Neck	Back	Arms	Legs	Other
Burning						
Sharp or Stabbing						
Cramping						
Dull Ache						
Shooting or radiating						

9.4. A4: Example Informed Consent

CONSENT FORM

Project title: The effects of combined acute muscle fatigue with and without blood flow restriction on postural control in healthy young adults.

Researchers Name: Dominic Langdon.

This form should be read in conjunction with the participant information sheet provided.

Please read the following statements and **sign your initials in the box** to show that you have read and understood them and that you agree with them.

		Please initial box
1.	I confirm that I have read and understand the information sheet for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.	
2.	I understand that my involvement is voluntary and that I am free to withdraw at any time during data collection and up until the first week after data has been collected, without giving any reason.	
3.	I understand that the information I disclose will remain confidential.	
4.	My data will not be identifiable by anyone other than the research team and all reasonable steps will be taken to ensure that my personal information is kept confidential.	

Please tick the box if you would like to receive summary results of the study ☐

Participant statement

I agree to take part in the study

Name: ________________________ Date: ________________________ Signature: ________________________

If you would like to obtain a summary report of the results of the study then please give your contact details below:

____

____

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9.5. A5: Pilot Study 1 – Supplementary Data

Table 9.1: Anteroposterior COP between and within session reliability for young adults.

AP	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
	1	56.9		0.44	Fair to Good	1.8	2.28	5.96	
DLS EO FIRM	2	57.2	0.75	0.57	Fair to Good	1.9	2.3	6.08	$F_{(2,18)}$ =.737, *P* =.493
	3	59.2		0.43	Fair to Good	2.0	2.33	6.4	
	1	59.4		0.62	Fair to Good	2.0	2.3	6.32	
DLS EC FIRM	2	60.1	0.59	0.48	Fair to Good	1.8	2.03	5.68	$F_{(2,18)}$ =.693, *P* =.513
	3	49.3		0.79	Excellent	2.1	3.34	7.56	
	1	42.7		0.79	Excellent	1.5	3.05	6.13	
DLS EO FOAM	2	56.9	0.44	0.46	Fair to Good	2.1	2.63	6.88	$F_{(2,18)}$ =.382, *P* =688
	3	50.8		0.28	Poor	1.7	2.58	6.02	
	1	33.3		0.36	Poor	1.8	5.04	8.62	
DLS EC FOAM	2	26.1	0.58	0.28	Poor	1.0	4.02	6.1	$F_{(2,18)}$ =.071, *P* =.932
	3	42.2		0.33	Poor	1.9	3.72	7.43	
	1	29.1		0.31	Poor	1.2	4	6.37	
SLS_L EO FIRM	2	50.3	0.43	0.46	Fair to Good	1.0	2.77	6.4	($F_{(2,18)}$ =1.766, *P* =1.99
	3	33		0.50	Fair to Good	1.5	4.15	7.06	
	1	17.66		0.43	Fair to Good	1.1	7.02	9.28	
SLS_L EC FIRM	2	52.22	0.57	0.74	Good	3.7	5.26	12.59	$F_{(2,18)}$ =.141, *P* =.870
	3	45.62		0.77	Excellent	3.2	5.7	12.08	

	1	44.12		0.57	Fair to Good	2.2	3.66	8.09	
SLS_L EO FOAM	2	66.24	0.64	0.07	Poor	3.2	2.97	9.45	$F_{(2,18)}$ =4.585, P =0.25
	3	57.89		0.20	Poor	3.7	4.42	11.81	
	1	49.1		0.56	Fair to Good	3.2	3.17	7.76	
SLS_R EO FIRM	2	44.01	0.5	0.71	Good	3.9	3.9	8.02	$F_{(2,18)}$ =2.158, P =.144
	3	52.59		0.97	Excellent	2.5	3.5	8.43	
	1	37.31		0.58	Fair to Good	2.8	5.98	11.56	
SLS_R EC FIRM	2	42.91	0.74	0.35	Poor	3.0	5.85	11.81	$F_{(2,18)}$ =7.168, P = .006
	3	43.68		0.62	Fair to Good	3.0	5.71	11.69	
	1	49.11		0.15	Poor	2.6	4.09	9.22	
SLS_R EO FOAM	2	46.68	0.66	0.02	Poor	2.0	3.47	7.49	$F_{(2,18)}$ =3.819, P =.041
	3	35.59		0.29	Poor	1.9	4.85	8.62	
	1	47.57		0.05	Poor	2.5	4.16	9.13	
TAN_L EO FIRM	2	36.2	0.58	0.81	Excellent	1.9	4.84	8.9	$F_{(2,18)}$ =1.278, P =.303
	3	32.24		0.88	Excellent	1.8	5.4	9.06	
	1	27.92		0.29	Poor	1.7	5.95	9.29	
TAN_L EC FIRM	2	33.5	0.66	0.59	Fair to Good	1.9	4.01	7.75	$F_{(2,18)}$ =.128, P = .880
	3	41.11		0.37	Poor	2.3	4.89	9.55	
	1	31.94		0.10	Poor	1.6	4.76	7.95	
TAN_L EO FOAM	2	24	0.33	0.14	Poor	1.2	5.16	7.56	$F_{(2,18)}$ =.058, P = .944
	3	41.51		0.58	Fair to Good	2.1	4.37	8.6	
	1	27.94		0.47	Fair to Good	1.6	5.56	8.69	
TAN_R EO FIRM	2	35.78	0.71	0.89	Excellent	2.0	5.02	8.96	$F_{(2,18)}$ =1.801, P = .194
	3	35.76		0.78	Excellent	2.0	4.8	8.84	
	1	31.65		0.40	Fair to Good	1.8	5.48	9.11	
TAN_R EO FIRM	2	30.24	0.7	0.62	Fair to Good	1.6	4.06	7.27	$F_{(2,18)}$ =.193, P = .827
	3	35.56		0.69	Fair to Good	1.8	4.57	8.12	
TAN_R EO FOAM	1	30.62	0.2	0.29	Poor	1.5	4.63	7.57	$F_{(2,18)}$ =4.509, P = .026

	2	56.3		0.43	Fair to Good	3.1	3.84	9.94
	3	52.26		0.87	Excellent	2.5	3.57	3.84

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

Table 9.2: Mediolateral COP between and within session reliability for young adults.

ML	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
	1	57.6		0.72	Fair to Good	1.2	1.4	3.7	
DLS EO FIRM	2	86	0.96	0.77	Fair to Good	1.69	0.8	4.18	$F_{(2,18)}$ = 1.770, P =.199
	3	53.78		0.17	Poor	0.91	1.2	3.05	
	1	99.2		0.33	Poor	3.2	0.9	7.35	
DLS EC FIRM	2	77	0.24	0.28	Poor	1.4	0.8	3.6	$F_{(2,18)}$=4.422, P =0.27
	3	53.2		0.20	Poor	1.3	1.9	4.56	
	1	62.5		0.66	Fair to Good	2	2.1	6.15	
DLS EO FOAM	2	97.9	0.36	0.65	Fair to Good	2.1	0.6	4.88	$F_{(2,18)}$ =.179, P =.838
	3	58.2		0.21	Poor	1.6	1.9	4.98	
	1	46.5		0.47	Fair to Good	2.2	3.8	8.27	
DLS EC FOAM	2	73.9	0.25	0.85	Excellent	1.5	1.1	4.18	$F_{(2,18)}$ =.404, P =.673
	3	54.6		0.74	Fair to Good	1.8	2.4	5.93	
	1	54		0.62	Fair to Good	3.2	4.3	10.75	
SLS_L EO FIRM	2	79.8	0.49	1.00	Excellent	1.6	1	4.17	$F_{(2,18)}$ =.302, P =.743
	3	64.7		0.83	Excellent	2.7	2.6	7.94	
	1	55.9		0.77	Excellent	4.7	5.9	15.26	
SLS_L EC FIRM	2	74.8	0.25	0.42	Fair to Good	1.5	1.1	4.11	$F_{(2,18)}$ =.486, P =623
	3	47.9		0.18	Poor	2.9	4.8	10.69	

	1	64		0.60	Fair to Good	3.9	3.8	11.59	
SLS_L EO FOAM	2	82.5	0.31	0.69	Fair to Good	1.5	0.8	3.92	$F_{(2,18)}$ =.3.331, P =0.59
	3	59.9		0.77	Excellent	2.7	3.1	8.54	
	1	59.3		0.75	Excellent	4.2	4.8	13.16	
SLS_R EO FIRM	2	76.8	0.29	0.20	Poor	2.1	1.1	5.28	$F_{(2,18)}$ =1.061, P =.367
	3	66.1		0.89	Excellent	2.6	2.4	7.59	
	1	52.3		0.37	Poor	3.4	4.9	11.81	
SLS_R EC FIRM	2	88.1	0.60	0.67	Fair to Good	2.3	1	5.59	$F_{(2,18)}$ = 1.935, P =.173
	3	58.7		0.54	Fair to Good	2.9	3.3	9.06	
	1	55.9		0.81	Excellent	3.9	5	12.84	
SLS_R EO FOAM	2	90	0.20	0.68	Fair to Good	2.2	0.9	5.33	$F_{(2,18)}$ =.334, P =.721
	3	42.2		0.76	Excellent	1.8	3.6	7.16	
	1	54.6		0.60	Fair to Good	2.9	3.8	9.51	
TAN_L EO FIRM	2	57.8	0.24	0.63	Fair to Good	1.8	2.1	5.69	$F_{(2,18)}$ =0.81, P =.921
	3	51.4		0.59	Fair to Good	1.6	2.3	5.46	
	1	78.3		0.45	Fair to Good	4	2.5	10.61	
TAN_L EC FIRM	2	92.7	0.35	0.59	Fair to Good	1.5	0.5	3.5	$F_{(2,18)}$ =4.098, P =.034
	3	41.7		0.80	Excellent	2.1	4.3	8.47	
	1	60.6		0.61	Fair to Good	3	3.3	9.33	
TAN_L EO FOAM	2	61.1	0.31	0.85	Excellent	1.6	1.8	5.07	$F_{(2,18)}$ = 3.104, P =.069
	3	41.6		0.75	Excellent	1.6	3.2	6.31	
	1	62.8		0.24	Poor	2.4	2.5	7.29	
TAN_R EO FIRM	2	45.7	0.43	0.91	Excellent	1.9	3.4	7.23	$F_{(2,18)}$ =1.422, P =267
	3	39.8		0.63	Fair to Good	1.1	2.5	4.81	
	1	46.3		0.45	Fair to Good	2.3	4	8.54	
TAN_R EO FIRM	2	74.3	0.29	0.21	Poor	1.4	1	3.82	$F_{(2,18)}$ =2.975, P =.077
	3	34.6		0.92	Excellent	1.3	3.4	5.92	
TAN_R EO FOAM	1	62.1	0.10	0.41	Fair to Good	2.2	2.3	6.57	$F_{(2,18)}$ =1.293, P =.299

	2	64.5		0.31	Fair to Good	1.8	1.7	5.24	
	3	51.9		0.69	Fair to Good	1.9	2.8	6.71	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

Table 9.3: Velocity COP between and within session reliability for young adults.

Velocity	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
	1	13.11		0.20	Poor	0.2	1.62	1.99	
DLS EO FIRM	2	26.53	0.3	0.88	Excellent	0.3	1.21	1.85	$F_{(2,18)}$ =1.258, P =.308
	3	19.86		0.79	Excellent	0.2	1.17	1.6	
	1	15.13		0.53	Fair to Good	0.3	2.06	2.61	
DLS EC FIRM	2	32.88	0.4	0.21	Poor	2.3	1.36	2.31	$F_{(2,18)}$ =3.381, P =0.57
	3	30.1		0.96	Excellent	2.3	1.4	2.27	
	1	14.65		0.32	Poor	0.3	2.11	2.66	
DLS EO FOAM	2	56.39	0.4	0.76	Excellent	1.0	1.19	3.08	$F_{(2,18)}$ =.493, P =.619
	3	49.53		0.39	Poor	0.8	1.21	2.75	
	1	34.07		0.20	Poor	1.6	4.36	7.54	
DLS EC FOAM	2	30.78	0.6	0.77	Fair to Good	0.9	2.77	4.53	$F_{(2,18)}$ =.132, P =887
	3	30.98		0.91	Excellent	0.8	2.57	4.22	
	1	15.42		0.14	Poor	0.5	3.78	4.82	
SLS_L EO FIRM	2	16.04	0.3	0.89	Excellent	0.5	3.19	4.11	$F_{(2,18)}$ =520, P =.603
	3	17.68		0.76	Excellent	0.5	2.95	3.91	
	1	13.81		0.29	Poor	1.1	8.81	10.96	
SLS_L EC FIRM	2	31.17	0.4	0.34	Poor	2.2	6.7	11.05	$F_{(2,18)}$ =726, P =.498
	3	29.82		0.55	Fair to Good	2.2	7.19	11.59	

	1	25.68		0.74	Fair to Good	1.1	4.34	6.53	
SLS_L EO FOAM	2	27.79	0.6	0.93	Excellent	1.0	3.55	5.54	$F_{(2,18)}$ =1.463, *P* =.258
	3	31.25		0.94	Excellent	1.2	3.81	6.28	
	1	19.03		0.74	Fair to Good	0.7	3.7	5	
SLS_R EO FIRM	2	19.61	0.4	0.81	Excellent	0.6	3.35	4.57	$F_{(2,18)}$ =.477, *P* =.628
	3	14.44		0.36	Poor	0.4	2.88	3.61	
	1	49.03		0.23	Poor	2.0	3.22	7.27	
SLS_R EC FIRM	2	57.04	0.7	0.44	Fair to Good	2.6	3.21	8.43	$F_{(2,18)}$ =4.351, *P* =.029
	3	59.59		0.15	Poor	2.0	2.25	6.22	
	1	49.13		0.09	Poor	2.5	4.02	9.07	
SLS_R EO FOAM	2	45.27	0.9	0.31	Poor	2.0	3.52	7.42	$F_{(2,18)}$ =.565, *P* =.838
	3	44.71		0.07	Poor	1.9	3.58	7.46	
	1	19.31		0.56	Fair to Good	0.6	3.22	4.38	
TAN_L EO FIRM	2	17.32	0.2	0.86	Excellent	0.4	2.79	3.66	$F_{(2,18)}$ =3.235, *P* =.63
	3	19.29		0.88	Excellent	0.5	2.54	3.44	
	1	25.73		0.38	Poor	1.7	6.62	9.97	
TAN_L EC FIRM	2	72.57	0.7	0.16	Poor	2.8	2.09	7.64	$F_{(2,18)}$ =6.319, *P* =.008
	3	42.57		0.55	Fair to Good	2.3	4.6	9.22	
	1	26.69		0.64	Fair to Good	1.1	4	6.13	
TAN_L EO FOAM	2	39.47	0.4	0.60	Fair to Good	1.9	4.26	8.09	$F_{(2,18)}$ =.305, *P* =.741
	3	41.54		0.91	Excellent	6.5	3.31	6.52	
	1	15.68		0.35	Poor	0.5	3.3	4.22	
TAN_R EO FIRM	2	16.39	0.2	0.74	Excellent	0.4	2.59	3.36	$F_{(2,18)}$ =1.505, *P* =.249
	3	47.49		0.13	Poor	1.3	2.18	4.78	
	1	30.62		0.11	Poor	1.7	5.23	8.54	
TAN_R EO FIRM	2	53.8	0.4	0.06	Poor	1.6	2.14	5.27	$F_{(2,18)}$ =.235, *P* =.793
	3	44.36		0.76	Excellent	1.8	3.26	6.75	
TAN_R EO FOAM	1	27.19	0.3	0.51	Fair to Good	1.0	3.48	5.38	$F_{(2,18)}$ =.454, *P* =694

	2	39.13		0.65	Fair to Good	1.9	4.23	7.99
	3	48.84		0.40	Fair to Good	1.9	3.12	7

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

Table 9.4: Anteroposterior COP within and between session reliability data in older adults.

AP	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
	1	63.3		0.33	Poor	2.3	1.9	6.4	
DLS EO FIRM	2	98.05	0.51	0.97	Excellent	3.5	1.2	8.2	$F_{(2,15)}$ =.53, P =949
	3	35.3		0.55	Fair to Good	0.5	1.1	2.1	
	1	41		0.15	Poor	1.6	2.9	6.2	
DLS EC FIRM	2	30.5	0.53	0.93	Excellent	1.0	2.0	4.1	$F_{(2,15)}$ =1.184, P =.333
	3	49		0.80	Excellent	1.5	1.7	4.6	
	1	42.3		0.18	Poor	2.2	3.8	8.1	
DLS EO FOAM	2	48.7	0.11	0.97	Excellent	3.1	2.7	8.9	$F_{(2,15)}$ =1.966, P =.175
	3	44.3		0.55	Fair to Good	1.5	2.5	5.5	
	1	33.5		0.36	Poor	2.1	5.2	9.4	
DLS EO FOAM	2	25.3	0.13	0.97	Excellent	1.7	4.5	7.9	$F_{(2,15)}$ =0.11, P =989
	3	23.9		0.15	Poor	1.3	5.0	7.5	
	1	57.9		0.55	Fair to Good	3.3	2.7	9.3	
SLS_L EO FIRM	2	49.2	0.58	0.83	Excellent	2.6	2.2	7.5	$F_{(2,15)}$ =.576, P =574
	3	33.6		0.12	Poor	1.1	2.8	5.0	
SLS_R EO FIRM	1	64.5	0.34	0.23	Poor	4.0	2.7	10.7	$F_{(2,15)}$ =1.060, P =.370
	2	87.1		0.14	Poor	5.7	0.3	11.7	

	3	42.5		0.62	Fair to Good	1.8	3.1	6.6	
	1	54.8		0.22	Poor	3.8	4.3	12.0	
TAN_L EO FIRM	2	66.9	0.15	0.86	Excellent	4.5	1.7	10.7	$F_{(2,15)}$ =2.898, P =.086
	3	60.1		0.85	Excellent	3.1	2.4	8.7	
	1	49.6		0.22	Poor	2.6	3.5	8.6	
TAN_L EO FOAM	2	40.3	0.46	0.86	Excellent	2.3	2.9	7.4	$F_{(2,15)}$ =1.129, P =.349
	3	67.4		0.77	Excellent	3.0	2.2	8.1	
	1	41.5		0.82	Excellent	2.5	4.6	9.7	
TAN_R EO FIRM	2	64.2	0.72	0.94	Excellent	4.5	1.9	11.0	$F_{(2,15)}$ =.798, P =.469
	3	80.4		0.60	Fair to Good	4.0	1.3	9.4	
	1	43.7		0.58	Fair to Good	2.5	3.5	8.5	
TAN_R EO FOAM	2	52.9	0.28	0.94	Excellent	3.7	2.7	10.2	$F_{(2,15)}$ =1.380, P =.282
	3	41.3		0.66	Fair to Good	1.7	2.7	6.1	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ''poor'', 0.40–0.75 = ''fair to good'', >0.75 = ''excellent'' reliability (Carpenter, 2001, Ruhe *et al*., 2010) (P = <0.05).

Table 9.5: Mediolateral Within and Between session reliability data in Older Adults.

ML	Day	CV (%)	BETWEEN ICC	WITHIN ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
	1	80.7		0.28	Poor	1.6	0.7	3.92	
DLS EO FIRM	2	98.96	0.47	0.06	Poor	1.82	0.1	3.75	$F_{(2,15)}$ =.303, P =.744
	3	52.28		0.34	Poor	0.7	0.7	2.1	
DLS EC FIRM	1	86.6	0.48	0.40	Fair to Good	1.6	0.6	3.86	$F_{(2,15)}$ =.1.990, P =.179
	2	87.5		0.35	Poor	1.6	0.3	3.42	

	3	48.7		0.28	Poor	0.8	0.9	2.46	
	1	63.5		0.07	Poor	1.7	1.4	4.69	
DLS EO FOAM	2	93.5	0.43	0.40	Fair to Good	1.4	0.2	3.04	$F_{(2,15)}$ =.938, *P* =.418
	3	45.6		0.65	Fair to Good	1	1.3	3.35	
	1	66.1		0.66	Fair to Good	2.5	1.9	6.91	
DLS EO FOAM	2	88.2	0.51	0.40	Fair to Good	2.3	0.1	4.61	$F_{(2,15)}$ =.1.056, *P* =.378
	3	59		0.65	Fair to Good	1.9	1.5	5.26	
	1	51.2		0.66	Fair to Good	1.7	1.8	5.32	
SLS_L EO FIRM	2	72.1	0.23	0.60	Fair to Good	2.7	0.7	6.19	$F_{(2,15)}$ =.4.534, *P* =.034
	3	41.3		0.53	Fair to Good	1.3	2	4.57	
	1	53.6		0.30	Poor	1.9	2.2	5.88	
SLS_R EO FIRM	2	55.9	0.61	0.50	Fair to Good	2.1	1.3	5.55	$F_{(2,15)}$ =.526, *P* =.604
	3	59.7		0.90	Excellent	2.2	1.7	6.15	
	1	46.9		0.46	Fair to Good	1.8	2.6	6.18	
TAN_L EO FIRM	2	58.6	0.63	0.59	Fair to Good	1.8	1	4.67	$F_{(2,15)}$ =.398, *P* =.680
	3	41		0.70	Fair to Good	1.3	1.7	4.29	
	1	64.4		0.26	Poor	1.7	1.4	4.73	
TAN_L EO FOAM	2	118.2	0.39	0.02	Poor	3.8	0.9	6.8	$F_{(2,15)}$ =.1.465, *P* =.270
	3	38.1		0.66	Fair to Good	1.3	2.2	4.72	
	1	54.6		0.54	Fair to Good	2.4	7.47	2.7	
TAN_R EO FIRM	2	57.6	0.24	0.04	Poor	4.5	2.7	11.69	$F_{(2,15)}$ =.357, *P* =.707
	3	58		0.03	Poor	2.6	1.6	6.87	
	1	39		0.20	Poor	1.5	2.6	5.67	
TAN_R EO FOAM	2	44.2	0.54	0.04	Poor	2.5	2.6	7.64	$F_{(2,15)}$ =.2.029, *P* =.174
	3	58.2		0.03	Poor	1.7	1.3	4.73	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ‘‘poor’’, 0.40–0.75 = ‘‘fair to good’’, >0.75 = ‘‘excellent’’ reliability (Carpenter, 2001, Ruhe *et al*., 2010) (*P* = <0.05).

Table 9.6: Velocity COP within and between session reliability data in older adults.

Velocity	Day	CV (%)	Between ICC	Within ICC	ICC Description	CI	Lower Bound	Upper Bound	One Way ANOVA
	1	23.5	0.91	0.04	Poor	0.3	1.3	1.9	
DLS EO FIRM	2	29.25		0.76	Excellent	0.4	0.8	1.6	$F_{(2, 15)}$ =.473. *P* =.634
	3	18.49		0.76	Excellent	0.2	0.9	1.3	
	1	40.2	0.29	0.90	Excellent	0.7	1.3	2.7	
DLS EC FIRM	2	32.2		0.70	Fair to Good	0.6	1.0	2.2	$F_{(2, 15)}$ =.910, *P* =.428
	3	30.2		0.76	Excellent	0.4	1.1	2.0	
	1	36	0.48	0.01	Poor	0.8	1.7	3.2	
DLS EO FOAM	2	28.4		0.60	Fair to Good	0.7	1.5	2.8	$F_{(2, 15)}$ =.243, *P* =.788
	3	31.2		0.76	Good	0.5	1.3	2.4	
	1	48.8	0.86	0.05	Poor	2.2	3.0	7.3	
DLS EO FOAM	2	21.7		0.87	Excellent	0.9	3.0	4.8	$F_{(2, 15)}$ =.664, *P* =.533
	3	19.1		0.04	Poor	0.7	3.1	4.5	
	1	50.9		0.04	Poor	2.2	2.3	6.7	
SLS_L EO FIRM	2	37.5	0.89	0.32	Poor	1.5	2.2	5.3	$F_{(2, 15)}$ =.1.425, *P* =.279
	3	40.3		0.42	Fair to Good	1.4	2.2	4.9	
	1	41.5		0.04	Poor	1.5	2.7	5.6	
SLS_R EO FIRM	2	48.8	0.70	0.65	Fair to Good	2.1	1.9	6.1	$F_{(2, 15)}$ =.888, *P* =.437
	3	30.3		0.35	Poor	1.2	2.9	5.3	
	1	57.8		0.07	Poor	1.9	1.9	5.8	
TAN_L EO FIRM	2	47.3	0.48	0.77	Good	1.3	1.2	3.8	$F_{(2, 15)}$ =.045, *P* = .956
	3	33.1		0.74	Good	0.9	1.6	3.3	

	1	45.9		0.08	Poor	1.4	2.1	4.8	
TAN_L EO FOAM	2	60	0.53	0.32	Poor	2.2	1.2	5.7	$F_{(2, 15)}$ =.056, *P* =.945
	3	34.4		0.91	Excellent	0.9	1.9	3.8	
	1	39.2		0.83	Excellent	1.2	2.3	4.6	
TAN_R EO FIRM	2	43.8	0.54	0.15	Poor	1.2	1.3	3.7	$F_{(2, 15)}$ =.486, *P* =.627
	3	34.4		0.93	Excellent	0.9	1.6	3.4	
	1	54.4		0.49	Fair to Good	1.8	1.7	5.4	
TAN_R EO FOAM	2	53.9	0.64	0.60	Fair to Good	2.1	1.5	5.6	$F_{(2, 15)}$ =.198, *P* =.823
	3	39.9		0.88	Excellent	1.0	1.6	3.7	

Key: DLS = Double leg stance, SLS_L = Single leg stance – left leg, SLS_R = Single leg stance – right leg, EO = Eyes Open, EC = Eyes Closed, OA = Older Adults, YA = Young Adults, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, ICC are <0.40 = ''poor'', 0.40–0.75 = ''fair to good'', >0.75 = ''excellent'' reliability (Carpenter, 2001, Ruhe *et al*., 2010) (*P* = <0.05

9.6. A6: Category Ratio-10

Borg CR10 Scale (1982)[12]

0	Nothing at all
0.5	Extremely weak (just noticeable)
1	Very weak
2	Weak (light)
3	Moderate
4	Somewhat strong
5	Strong (heavy)
6	
7	Very strong
8	
9	
10	Extremely strong (almost max)
•	Maximal

The Borg scale is used to help us understand the intensity or severity of your discomfort experienced when performing an exercise. We will ask you to use this scale to rate the intensity of your discomfort before, during, and after your exercise.
Please review the scale to see the various levels from which you can choose.

The top of the scale, "0 or nothing at all," means *no pain or discomfort at all*.

The bottom of the scale, "10 or maximal," means the *most discomfort you have ever felt/ you have ever experienced or could imagine experiencing*.

When we ask you to rate the intensity of your discomfort breathlessness, please place your finger on the number that best describes the intensity you are experiencing You may also place a finger between 2 numbers if that better describes the intensity of your discomfort
Please let us know if you have any questions before we begin.

9.7. A7: Visual Analogue Scale

VAS: Please rate your enjoyment and perception of the exercise based on the statements below

Session No:____

1. Enjoyment

No Enjoyment	________________________	Very Enjoyable

2. Perceived Effectiveness

Not effective	________________________	Strongly Agree, very effective

3. Perceptions of Volume Load

No workload	________________________	High workload

4. Fatigue

Not Fatiguing (no exertion)	________________________	Fatiguing (Extremely hard)

9.8. Power Analysis – Extracted from G*Power

9.8.1. Pilot Study 3 + 4

9.8.1.1. A Priori Power

F tests – ANOVA: Repeated measures, within – interaction

Analysis:	A priori: Compute required sample size		
Input:	Effect size f	=	0.2526456
	α err prob	=	0.05
	Power (1−β err prob)	=	0.80
	Number of groups	=	2
	Number of measurements	=	3
	Corr among rep measures	=	0.5
	Nonsphericity correction ϵ	=	1
Output:	Noncentrality parameter λ	=	10.7234063
	Critical F	=	3.1751410

	Numerator df	=	2.0000000
	Denominator df	=	52.0000000
	Total sample size	=	28
	Actual power	=	0.8201386

9.8.1.2. Post Hoc

F tests – ANOVA: Repeated measures, within – interaction

Analysis:	Post hoc: Compute achieved power		
Input:	Effect size f	=	0.2526456
	α err prob	=	0.05
	Total sample size	=	8
	Number of groups	=	2
	Number of measurements	=	3
	Corr among rep measures	=	0.5
	Nonsphericity correction ϵ	=	1
Output:	Noncentrality parameter λ	=	3.0638304
	Critical F	=	3.8852938
	Numerator df	=	2.0000000
	Denominator df	=	12.0000000
	Power (1–β err prob)	=	0.2630846

9.8.2. Pilot Study 5

9.8.2.1. A Priori Power

F tests – ANOVA: Repeated measures, within–between interaction

Analysis:	A priori: Compute required sample size		
Input:	Effect size f	=	0.2526456
	α err prob	=	0.05
	Power (1–β err prob)	=	0.80
	Number of groups	=	2
	Number of measurements	=	3
	Corr among rep measures	=	0.5
	Nonsphericity correction ϵ	=	1
Output:	Noncentrality parameter λ	=	10.7234063
	Critical F	=	3.1751410
	Numerator df	=	2.0000000
	Denominator df	=	52.0000000
	Total sample size	=	28
	Actual power	=	0.8201386

9.8.2.2. Post Hoc

F tests – ANOVA: Repeated measures, within–between interaction

Analysis:	Post hoc: Compute achieved power		
Input:	Effect size f	=	0.2526456
	α err prob	=	0.05
	Total sample size	=	16
	Number of groups	=	2
	Number of measurements	=	3
	Corr among rep measures	=	0.5
	Nonsphericity correction ϵ	=	1
Output:	Noncentrality parameter λ	=	6.1276607
	Critical F	=	3.3403856
	Numerator df	=	2.0000000
	Denominator df	=	28.0000000
	Power ($1-\beta$ err prob)	=	0.5450294