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The effects of acute and chronic upper and lower body exercise on postural sway and functional balance

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The effects of acute and chronic upper and lower body exercise on postural sway and functional balance

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Abstract

Acute lower body exercise elicits adverse effects on balance performance and subsequent fall risk. However, little information exists for upper body exercise and postural sway. The series of experimental studies presented in this thesis investigated the effects of acute upper and lower body exercise on postural sway in healthy young and older adults and determined whether the acute negative effects of exercise can be removed by an improvement in training status. Chapter 4 examined the effects of maximal and submaximal (absolute and relative exercise intensities) arm crank ergometry (ACE) and cycle ergometry (CE) on postural sway in young healthy adults. Cycling elicited an immediate increase in post exercise postural sway whereas ACE did not. Chapter 5 compared the effects ACE, CE and treadmill walking (TM) on postural sway in healthy older adults. Based on the findings of Study 1, submaximal exercise was performed at the same relative intensity (50 % HR_E). In agreement with Chapter 4, CE and TM elicited post exercise balance impairments lasting for ~ 10 min post exercise. ACE performed at the same relative intensity as the lower body did not elicit post exercise balance impairments in older adults. Collectively, these acute studies suggest that lower limb exercise may acutely increase fall and injury risk in the immediate period after exercise cessation. This is important because practitioners and clinicians should acknowledge that the prescription of conventional training modes might potentially elicit transient impairments in neuromuscular function. However, in this context it appears that seated exercise with the arms may not induce a significant enough stimulus to cause sensorimotor disturbance to postural stability and thus may be a safer alternative exercise mode for fall risk populations or individuals who are very sedentary. Chapter 6 examined differences in balance performance, as measured by quantitative posturography and functional balance tests, among different age groups. Measures of postural sway were able to distinguish between younger ($\leq$ 60 years) and older ($\geq$ 60 years) adults whereas functional performance tests suggested that balance impairments were observed earlier ($\geq$ 50 years). This study enabled a range of tests to be determined for use with subsequent training interventions. Chapter 7 examined the effects of 6-weeks upper or lower body exercise training on postural sway. Upper and lower body training elicited similar improvements in specific (~ 25 %) and cross transfer (~ 12 %) exercise tolerance. Both modes of training elicited favourable balance adaptations. Specifically, upper body training improved mediolateral aspects of postural sway, while lower body training improved anteroposterior aspects of sway. It was proposed that an improvement in cross transfer exercise capacity after upper body exercise reduced the physiological strain experienced during CE, thus reducing post exercise balance impairments and that an increase in abdominal and trunk strength from upper body exercise training reduced sway following ACE. Importantly, ACE also elicited an improvement in functional reach distance and timed up and go test speed. Conversely, CE improved lower limb strength which elicited an improvement in lower body dynamic balance. It is likely that engaging in both upper and lower body exercise will be better that either mode of exercise alone for both health and fitness and balanced incorporated in everyday life. ACE elicits a number of significant benefits to cardiovascular fitness and balance which is important for a number of older subgroups who might have difficulty engaging in lower limb exercise, such as those who are very sedentary, those with lower limb injury/disease or undergoing surgery rehabilitation and individuals who are overweight. Specifically, ACE may offer a pathway from sedentary living to physical activity. Upper body endurance exercise can contribute to a multimodal training stimulus by eliciting favourable adaptations in fitness, functional performance and balance. Such responses are important because this type of exercise may serve a feasible and time-efficient training regime for older adults, which will likely result in improved attrition and adherence to physical activity.

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Publications

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Hill, M.W., Talbot, C., and Price, M. (2014) Predicted maximal heart rate for upper body exercise testing. *Clinical Physiology and Functional Imaging.* Oct 16. doi: 10.1111/cpf.12201. [Epub ahead of print]

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Conference proceedings

Hill, M.W., Goss-Sampson, M., Duncan, M.J., and Price, M.J. (2012). The effects of upper and lower body exercise on postural sway. International Convention on Science, Education & Medicine in Sport (ICSEMIS), Scottish Exhibition & Conference Centre, Glasgow, UK, July 19th – 24th 2012.

Hill, M.W., Oxford, S.W., Duncan, M.J., and Price, M.J. (2013). The effects of arm crank ergometry, cycle ergometry and treadmill walking on postural sway during quiet stance in healthy elderly women. British Association of Sport and Exercise Sciences (BASES), University of Central Lancashire, Preston, UK, Sept 3rd – 5th 2013.

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Abbreviations

Absolute	ABS
Analysis of Variance	ANOVA
Anteroposterior	AP
Anticipatory postural adjustments	APA
Arm crank ergometry	ACE
Arterial-venous oxygen difference	$a\text{-}\overline{v}O_2$
Berg Balance Scale	BBS
Blood lactate	Bla
Blood pressure	BP
Body mass index	BMI
Breath frequency	Bf
Carbon dioxide	CO_2
Cardiac output	Q
Central nervous system	CNS
Centre of pressure	COP
Anteroposterior COP displacement	COP_{AP}
Mediolateral COP displacement	COP_{ML}
Mean velocity of the COP	COP_V
Path length of the COP	COP_L
Centre of mass	COM
Centimetres	cm
Coefficient of variation	CV
Cycle ergmetry	CE
Electromyography	EMG
Gastrocnemius medialis	EMG_{GS}
Tibialis anterior	EMG_{TA}
Rectus femoris	EMG_{RF}
Biceps femoris	EMG_{BF}
Rectus abdominus	EMG_{RA}
Erector spinae	EMG_{ES}
Eyes open	EO
Eyes closed	EC
Functional Reach Test	FRT
Heart rate	HR
Heart rate reserve	HR_E

Hertz	Hz
Intra-class correlation coefficient	ICC
Kilogram	Kg
Lower body exercise training group	LBX
Maximal heart rate	HR_{MAX}
Maximal voluntary isometric contraction	MVIC
Mediolateral	ML
Millimetres	mm
Milliseconds	ms
Minute ventilation	$\dot{V}_E$
Minutes	Min
Multi-Directional Functional Reach	MDFR
Oxygen	O_2
Peak minute power	W_{MAX}
Peak oxygen uptake	$\dot{V}O_{2PEAK}$
Peak minute ventilation	$\dot{V}_{EPEAK}$
Physical activity readiness questionnaire	PAR-Q
Pre exercise trial	Pre
Post exercise trial	Post
Rating of perceived exertion	RPE
Local	RPE_L
Central	RPE_C
Relative	REL
Respiratory exchange ratio	RER
Resting heart rate	RHR
Revolutions per minute	$rev \cdot min^{-1}$
Seconds	s
Standard deviation	SD
Star Excursion Balance Test	SEBT
Stroke volume	SV
Target heart rate	THR
Tidal Volume	Vt
Timed Up and Go Test	TUG
Treadmill walking	TM
Upper body exercise training group	UBX
Watts	W

Chapter 1

Introduction

The ability to maintain a stable upright stance is an essential component of any daily activity (Bisson et al. 2014). In order to remain in an upright position, the central nervous system must integrate information from different sensory systems (somatosensory, visual and vestibular) and modulate commands to the neuromuscular system (Bisson et al. 2011). It has recently been well established that acute lower limb exercise may alter proprioceptive and kinaesthetic properties of the joints (Miura et al. 2004) by increasing the threshold of muscle spindle discharge, therefore disrupting afferent feedback and resulting in altered conscious joint awareness (Gribble and Hertel 2004). Additionally, general exercise engaging several muscle groups decreases muscle force generating capacity (Lepers et al. 1997; Nardone et al. 1997; Taylor et al. 2000). Therefore, both the sensory and motor contributions of the sensorimotor postural process appear to be affected by lower limb exercise (Vuillerme, Anziani and Rougier 2007).

In recent years, growing interest has been directed towards investigating the effects of exercise engaging the lower body musculature on the control of upright bipedal standing. Popular recreational exercise such as cycling and walking elicit acute balance impairments in both healthy young (Demura & Uchiyama, 2009; Gouchard et al., 2002; Mello, de Oliveira, & Nadal, 2010; Vuillerme, & Hintzy, 2007) and older adults (Donath et al. 2013; Egerton, Brauer & Cresswell 2009; Maciaszek, Stemplewski & Osinski 2010; Stemplewski et al. 2012; 2013). Such findings are particularly interesting since these exercises are often prescribed for improving and/or maintaining fitness and exercise rehabilitation (Donath et al. 2013). The reduced balance control, often assessed using centre of pressure (COP) measures of postural sway using a force platform, suggest an increased risk of falling post lower body exercise (Stemplewski et al. 2012). However, the prevalence of upper body tasks in a variety of occupational (i.e., handling heavy machinery) and recreational activities (i.e., kayaking and canoeing) highlights the importance of investigating the effects of upper body exercise on postural sway (Nussbaum 2003). Given that ACE possess the potential to elicit positive adaptations for everyday activities (Groslambert et al. 2008) it is of interest to health practitioners prescribing exercise for older adults to understand if this mode of exercise impacts upon fall risk factors.

Currently, it is unclear whether postural stability is adversely affected by fatigue of none postural muscles. It is possible to examine this question more closely by directly comparing the effects of exercise engaging non-postural muscles (e.g., the arms) with muscles more directly in control of postural stability (e.g., the legs). This approach may provide insight into how exercise using none postural muscles affect the control of non-fatigued postural muscles. Although lower body muscles are more directly involved in postural control during quiet standing (Winter et al. 1996), muscle proprioceptive input operates through a proprioceptive chain linking the eyes to the feet (Roll and Roll 1988). Indeed, a reduction in postural stability has been reported following localised muscle fatigue of the trunk extensors (Parreira et al. 2013; Vuillerme, Anziani and Rougier 2007), forearm (Kennedy 2014), cervical extensor muscles (Duclos et al., 2004; Gosselin et al., 2004; Schieppati et al., 2003; Stapley et al., 2006) and the deltoids (Nussbaum, 2003). Several studies have suggested that changes in postural control after fatigue of non-postural muscles are caused by central mediation of the motor commands rather than limitations within the fatigued muscles (Kanekar et al. 2008; Kennedy et al. 2014; Morris and Allison 2006; Strang et al. 2009). Changes in postural control after fatiguing exercise using non-postural muscles reflect a more cautious and rigid postural strategy. For example, several studies have reported increased incidence of co-contraction of the tibialis anterior and medial gastrocnemius muscles (Kanekar et al. 2008) and decreased COP displacement following upper limb fatigue (Kennedy et al. 2012). According to Kennedy et al. (2014) these changes likely contribute to a more ‘tightly’ controlled posture following arm fatigue. Co-contraction elicits increased stiffness of the ankle which might reduce the COP displacement preventing a drift of the centre of gravity closer to the limits of stability thus limiting fall risk (Kennedy et al. 2014). Nonetheless, it is difficult to rationally generalise the effects of localised muscle fatigue (e.g., a single joint and muscle group, such as shoulder flexion) to more dynamic general exercise (e.g., involving multiple joints and several muscles groups, such as arm cranking) (Paillard 2012). Whether these findings would have been different if the entire upper body musculature were engaged is not clear.

Inconsistent findings have been reported with regards to the effects of ACE on postural sway (Douris et al. 2011; Smith et al. 2010). For example, anaerobic (Wingate) CE reduces postural stability when standing on a single limb to a greater extent than ACE in healthy young adults (Douris et al., 2011). However, following a maximal incremental exercise test, ACE impaired postural stability to a greater extent than CE. Douris et al. (2011) specified that the trunk stabilisers were more fatigued following maximal aerobic ACE compared to CE, and thus the torso musculature was unable to assist in maintaining balance. This type of maximal exercise is dissimilar to the physical activity that older adults are exposed to during daily life or that used during exercise rehabilitation making practical applications difficult. In contrast, when standing on two legs, ACE appears to elicit minimal balance impairments in healthy older adults (Smith et al., 2010). With the current state of knowledge, the field of upper body exercise on postural sway remains in its infancy. Further evidence is required to clarify the effects of ACE on postural sway.

Dynamic arm movement training in a standing position improves standing postural control in patients with chronic stroke (Waller and Prettyman 2012) and seated Kayak training improves seated postural stability in paraplegics (Bjerkefors, Carpenter and Thorstensson 2007). However, the question remains as to whether seated arm training improves standing balance. Bjerkefors and colleagues (2007) reported that Kayak training (which closely mimics movements during ACE) improved trunk stability in a spinal cord injured cohort (injuries ranging from T3 to T12). Therefore, there appears to be transfer effects of abilities trained on the kayak ergometer onto capabilities to control sitting balance. Seated ACE places demands on the trunk for stabilisation because of the asynchronous pushing and pulling of the ergometer handles (Di Blasio et al. 2009). To the author's knowledge, no publication exists on training the upper body in a sitting position on standing balance in able bodied individuals. Furthermore, the reported effects of cycling training on balance and fall risk are scant (Buchner et al. 1997a,b; Hassanlouei et al. 2014; Bouillon, Sklenka and Ver 2009; Rissel et al. 2013). Cycling has been found to increase leg strength and subsequently improves balance performance (Rissel et al., 2013) and reduces the incidence future falls (Buchner et al. 1997a). Determining the adaptations of ACE training on postural stability may help to distinguish differences between upper and lower body exercise and their potential benefits for fall risk and provide important findings for adults who have reduced lower body exercise capacity.

Falls are often multifaceted in origin (Rubenstein and Josephson 2006) and therefore falling is seldom due to a single cause. A number of factors contribute to falling such as environment-related (Rubenstein and Josephson 2006), gait and balance disorders (Maki, Holliday and Topper 1994; Piirtola and Era 2006), muscle weakness (Moreland et al. 2004), cognitive impairment (Muir, Gopaul and Odasso 2012) and visual problems (Lord 2006). A decrease in muscular strength which coincides with reductions in postural balance are of primary interest in the present thesis since such risk factors have been shown to be modifiable by training in older adults (review by Howe et al. 2008). However, there is no single balance tool that could be considered as a 'gold standard' to assess the integrity of the postural control system (Horak 1997). A number of characteristics associated with novel balance tasks have been shown to be associated with an increased risk of falling, such as reduced voluntary displacement of the centre of gravity during leaning tasks (Maki, Holliday and Topper 1994), impaired functional reach distance (Duncan et al. 1990), increased postural sway during quiet stance (Piirtola and Era 2006) and slower timed walking tests (Podsiadlo and Richardson 1991).

Studies determining the acute effects or chronic adaptations of exercise on balance adopt a range of balance tasks, such as single limb stance (Gribble and Hertel 2004), semi tandem stance (Bisson et al. 2010), standing on a compliant surface (Bisson et al. 2011) and dynamic postural control (Gribble and Hertel 2004) however, for comparative purposes most studies adopt a bipedal stance (Vuillerme, Sporbert and Pinsault 2009). Generally, there is no agreement as to which test should be used when determining chronic adaptations to exercise training. Furthermore, much of the available literature focusing on age related reductions in postural balance are limited to either women (Isles et al. 2004; Choy, Brauer and Nitz 2007), men (Nolan et al. 2010; Illing et al. 2010), or on either young or older adults (Nagai et al. 2011). This leaves gaps in the knowledge of the middle decades of life. Identifying the change in postural balance tasks over every decade (e.g., 20 – 80 years) will allow the identification of deterioration in balance and thus the period when interventions should be implemented. Additionally, such data will allow the identification of which aspects of balance deteriorate most among the older population and therefore might be suitable to target with an exercise training intervention.

1.1 Aims and objectives of the thesis

The principal aim of the thesis was to examine how acute and chronic upper and lower body exercise impact upon postural sway and other functional abilities. In order to do this, four main objectives were formulated:

- To examine the effects of upper and lower body exercise effects on postural sway in young healthy adults
- To examine the effects of upper, lower and whole body exercise on postural sway in older but otherwise healthy adults
- To analyse the differences in postural stability, walking speed and dynamic balance in young, middle aged and older adults to develop a range of suitable tests which can be

used to examine potential improvements in postural stability and functional ability following exercise training

- To examine the effects of 6-weeks of exercise training using either the upper or lower body on a range of balance tests

1.2 Organisation of thesis

This thesis is presented as a comprehensive review of the literature regarding upper and lower body exercise and postural sway (Chapter 2) and a general methods section giving details of generic methodologies used in each study and into four experimental chapters (Chapter 3). Chapters 4 and 5 examined the effects of upper and lower body exercise effects on postural sway in healthy young (Chapter 4) and older (Chapter 5) adults. The multifaceted nature of balance was explored and a range of postural sway and functional balance assessments were identified which could be used with a training intervention among older adults (Chapter 6). The final experimental chapter explores the acute effects of ACE and CE on postural sway before and after a period of upper body or lower body exercise training (Chapter 7). The final chapter (Chapter 8) presents a general summary and discusson of all the findings in addition to limitations and directions for future research.

Chapter 2

Literature Review

2.0 Introduction

This chapter aims to initially briefly describe the current knowledge and available research regarding the physiological responses and training adaptations to upper and lower body exercise. The review will then describe the current understanding of postural control and postural sway. The review will conclude with postural sway responses to upper and lower body exercise with particular reference to arm crank ergometry. To the author's knowledge, reviews are available regarding upper body exercise (Franklin 1985; Sawka 1986) but are rather dated and do not explicitly consider the application of ACE for healthy able bodied individuals and more recent developments in clinical groups. A more recent review is available for the effects of muscle fatigue on postural sway (Paillard 2012) but lacks applications to upper body exercise. The principal aim of this review chapter is to present a more comprehensive and specifically focused review integrating the area of upper body exercise with postural sway.

2.1 Advances in arm crank ergometry research

Arm cranking was historically employed in western prisons in the early 1800's as a punishment for petty crime. Prisoners would routinely turn a hand crank with thirty pounds of pressure one thousand times an hour for up to ten hours a day (Godfrey, Barry and Lawrence 2005). The first known research associated with conventional arm crank ergometry (ACE) was conducted by Collett and Liljestrand (1924) who reported a greater physiological strain elicited by arm exercise compared to leg exercise performed at the same metabolic rate. Since this early publication a number of studies were published in the late 60's to early 80's which concentrated on comparing the physiological responses to ACE and cycle ergometry (CE) at a given power output (Astrand et al. 1965; Reybrouck et al. 1975; Pendergast et al. 1979; Secher et al. 1974), the effects of cadence of peak physiological responses (Sawka et al. 1983), thermoregulatory responses to ACE (Pimental et al. 1984; Sawka et al. 1984) and the importance of ACE for adults who do not have the habitual use of their lower body, such as wheelchair users (Hjeltnes 1977). However, these studies were not followed up until more recently.

Research involving ACE has developed over the past decade, focusing upon a broad range of applications. For example, studies have focused on thermoregulatory responses (Price 2006; Price and Campbell 1997; 2003), studies of exercise specificity (DeJong et al. 2009; Forbes and Chilibeck 2007), applications for standing ACE as adopted by America's cup 'grinders' (Neville et al. 2009; 2010), hand cycling (Goosey-Tolfrey, Alfano, and Fowler 2007; Goosey-Tolfrey & Sindall, 2007) and the development of suitable protocols used to examine aerobic and anaerobic capability (Price et al. 2011; Price et al. 2007; Smith, Doherty and Price 2007).

In accordance with the principle of exercise specificity, most exercise rehabilitation regimens have involved the lower body musculature (Hiatt et al. 1995). However, for individual's undergoing hip replacement rehabilitation or those with lower limb ischemic pain, lower body exercise is often not feasible. Accordingly, recent studies have reported that populations with reduced lower body exercise capacity are able to achieve a greater exercise intensity with the arms compared to the legs, thereby eliciting greater potential for cardiovascular training adaptations (Grange et al. 2004; Saxton et al. 2011; Tew et al. 2009; Treat-Jacobson, Bronas and Leon 2009; Zwierska et al. 2007). These applications are still being developed, specifically for healthy adults.

2.2 Importance of arm crank ergometry

Arm crank ergoemtry provides a generic means in which physiological responses to upper body exercise can be examined (Smith and Price 2007; Smith, Doherty and Price 2007). This mode of upper body exercise testing is reproducible (Bar-Or & Zwiren 1975; Lazarus, Cullinane & Thompson 1987) is easily modified using cycle ergometers and is the least physiologically complex form of upper body exercise (Sawka 1986). As a result, this exercise modality has become increasingly popular in clinical rehabilitation and exercise / fitness centres.

Recent research has reported that ACE training elicits improved walking performance of a similar magnitude to cycle ergometry (CE) and treadmill (TM) training in older adults with reduced lower body exercise capability (Tew et al. 2009; Saxton et al. 2011; Treat-Jacobson, Bronas and Leon 2009; Zwierska et al. 2007). Increased exercise tolerance during exercise with the untrained limbs has been referred to as a cross transfer effect, which has previously been interpreted as indirect evidence of the central nature of training adaptations (Pogliaghi et al. 2006). Indeed, Tew et al., (2009) measured muscle tissue O_2 saturation [StO_2] using Near-infrared spectroscopy (NIRS) during a treadmill test before and after 12-weeks of ACE in patients with peripheral arterial disease. It was reported that there were faster oxygen kinetics and an increase in submaximal StO_2 during treadmill walking following ACE training. These results add support that cross transfer improvements in walking performance after ACE training are partly attributable to improved delivery of O_2 to the lower limbs, although the underlying mechanisms remain unclear. Various central and peripheral circulatory adaptations may be associated with enhanced O_2 delivery to the lower limbs following ACE training such as; increased stroke volume, cardiac output and blood volume. Indeed, ACE training improves stroke volume in young women and men (Loftin et al. 1988) and lowers submaximal HR in clinical populations (Walker et al. 2006), which is indicative of an increase in stroke volume and capillarisation.

In healthy but sedentary older male adults it has been shown that ACE and CE training elicits similar improvements in specific (trained muscles) and cross transfer (untrained muscles) exercise tolerance at both maximal and submaximal intensities (Pogliaghi et al. 2006). From a practical perspective, ACE could potentially be an effective alternative form of exercise for both healthy older adults and clinical populations with lower limb impairment. Furthermore, since many recreational and occupational activities require sustained arm work to a greater extent than leg work, albeit it at a lower intensity (e.g., dressing, bathing, cooking) (Franklin 1989; Waller and Prettyman 2012), ACE may possess other important applications which are relevant to daily life (Groslambert et al. 2006). Ultimately, training the arms as well as the legs may become extremely important in later life.

There is a need to investigate the potential applications of ACE in healthy able-bodied adults which will allow future research and applications to be made to clinical groups. One line of enquiry which has received growing interest is the effects of exercise on postural sway. The acute negative consequences of lower limb exercise on postural sway have been documented (Section 2.8). Arm cranking is a mode of exercise which is likely not to elicit such effects of muscle fatigue and subsequent balance impairment. This is a novel area with potential applications for daily life in healthy and movement impaired populations and is an important step in future assessments of the effects of exercise on fall risk.

2.3 Physiological responses to acute upper and lower body exercise

It is well known that physiological responses are different between the arms and legs (e.g., Sawka 1986). For example, at a given submaximal workload, arm exercise elicits a greater cardiac output, oxygen uptake ($\dot{V}O_2$), heart rate (HR), blood pressure (BP), ventilation and blood lactate response compared to leg exercise. More specific responses to maximal and submaximal upper and lower body exercise will be discussed in the following sections.

2.3.1 Peak physiological responses

For the purposes of this thesis it is important to note that the continuation of exercise during ACE is usually constrained by peripheral limiting factors (e.g., the ability to extract and utilise oxygen) as opposed to central (e.g., cardiac output) limiting factors (Sawka 1986). Therefore, maximal oxygen uptake during incremental exercise is referred to as peak oxygen uptake ($\dot{V}O_{2PEAK}$) throughout this thesis (Magel et al., 1975; Smith & Price 2007). As expected, most untrained individuals achieve a lower peak power output (W_{PEAK}) during ACE compared to CE (Sawka et al. 1982). In agreement, peak oxygen uptake during ACE is ~70% of the $\dot{V}O_{2PEAK}$ attained during maximal effort CE in untrained young (Lyons et al., 2007; Reybrouck, Heigenhauser and Faulkner 1975; Pendergast 1989; Pimental et al. 1984; Sawka et al. 1982; Vokac et al. 1975) and older adults (Pogliaghi et al., 2006) with a range of 39 - 89% of CE values reported in the literature (Table 2.1). The greater percentage of lower body $\dot{V}O_{2PEAK}$ reflects those who are trained in the upper body (Seals & Mullin 1982). Maximal cardiac output and heart rate values are ~30% lower during peak ACE compared to CE (Reybrouck, Heigenhauser & Faulkner 1975) which corresponds with differences in peak oxygen uptake (Stenberg et al., 1967).

Most investigations have reported lower blood lactate concentrations upon reaching volitional exhaustion during ACE (e.g., 9.7 $mmol \cdot L^{-1}$) compared to CE (e.g., 13.5 $mmol \cdot L^{-1}$) (Astrand et al., 1986) which likely reflects the amount of muscle mass engaged. Sawka (1986) specifies that while blood lactate concentration per unit of muscle mass may be equivalent, total lactate production during CE would be greater than the blood lactate attained during ACE. The lower peak values achieved during ACE are likely due to peripheral factors limiting exercise such as reduced potential to generate muscular tension due to the smaller muscle mass (Sawka 1986), an increase in peripheral resistance (Sawka et al., 1983), smaller venous return (Toner et al. 1983), a greater isometric component (Sawka, 1986), an increase in sympathetic neural drive

(Vokac et al., 1975) and a reduced oxidative capacity (e.g., greater number of type II fibres) of the upper body muscles (Oliver et al. 2008; Sawka 1986).

Fatigue manifests itself acutely and is linked to localised muscular fatigue rather than central limiting factors (Sawka, 1986). Upon cessation of maximal ACE, local ratings of perceived exertion (working muscles) (RPE_L) are typically greater than central ratings of perceived exertion (ventilatory and circulatory exertion) (RPE_C), reflecting the peripheral limitation of arm exercise (Pandolf et al. 1984; Price and Campbell 1997; Price et al. 2007; Price et al. 2011; Sawka et al. 1983; Smith et al. 2001; Smith, Doherty and Price 2007). As a result, both RPE_L an RPE_C will be reported for upper and lower body exercise.

2.3.2 Submaximal exercise responses

As highlighted previously, the early results of Collett and Liljestrand (1924) suggested that a greater physiological strain is elicited by arm compared to leg exercise performed at the same metabolic rate. These results have since received significant support (Stenberg et al., 1967; Glaser et al., 1980; Hellerstein & Franklin 1984; Sawka 1986) and are further addressed below.

2.3.2 i Energy cost

For untrained individuals, ACE elicits a greater $\dot{V}O_2$ than CE at the same power output (Astrand et al., 1965; Davies & Sargeant 1974; Franklin et al., 1983; Pendergast 1989; Vocak et al., 1975). Previous studies have noted that efficiency is lower during ACE compared to CE at the same absolute (Davis and Sargeant 1974; Toner et al. 1983) and relative (Kang et al. 1997) exercise intensities. The lower efficiency during ACE is a result of greater energy expenditure for unmeasured work, such as torso stabilisation and isometric contraction (Powers, Beadle and Mangum 1984). Furthermore, contractile efficiency varies with skeletal muscle fibre type (Whipp and Wasserman 1969), which is important considering that some studies have reported a greater percentage of energetically inefficient fast-twitch fibres in the upper compared to lower body musculature (Ahlborg and Jensen-Urstad, 1991; Johnson et al. 1973; Susheela and Walton 1969). Increased oxygen requirements for muscular stabilisation of the torso (Stenberg et al., 1967; Toner et al. 1983), gripping of the ergometer handles (Davies & Sergeant 1974), a greater isometric component (Sawka et al., 1982), a smaller muscle mass (Sawka 1986) as well as an increase in the recruitment of type II muscle fibres (Sawka 1986) which have significantly lower metabolic efficiency than type I fibres (Koppo, Bouckaert & Jones 2002) have been suggested as the mechanisms to explain the greater oxygen uptake during submaximal ACE compared to CE at the same absolute power output.

2.3.2 ii Cardiovascular responses

Cardiac output has been shown to be similar between ACE and CE at the same oxygen uptake (Stenberg et al. 1967). However, at any given $\dot{V}O_2$, HR is ~ 20 % higher during ACE compared to CE (Astrand et al., 1965; Reybrouck et al., 1975; Pendergast et al., 1980; Pendergast 1989; Toner et al., 1983; Sawka 1986), while stroke volume is reported to be 10 – 20 % lower during ACE than CE (Stenberg et al. 1967). This likely reflects greater sympathetic stimulation during ACE, increasing myocardial contractility (Sawka 1986). Furthermore, blood pressure is typically greater during ACE compared to CE (Astrand et al. 1967), which is likely a result of greater total peripheral resistance, since cardiac output is similar to upper and lower body exercise. Vessel radius also contributes to peripheral resistance during ACE (Sawka 1986).

Table 2.1: Comparison of peak oxygen uptake and heart rate between arm crank ergometry and cycle ergometry

Author	**Peak oxygen uptake ($L \cdot min^{-1}$ / $ml \cdot kg^{-1} \cdot min^{-1}$)**		**Maximal heart rate ($beats \cdot min^{-1}$)**	
Young adults	**ACE**	**CE**	**ACE**	**CE**
Davis et al. (1976)	2.43 ± 0.69	3.68 ± 0.41	184 ± 12	193 ± 10
Kang et al. (1997)	2.24 ± 0.54	2.98 ± 0.52	170 ± 17	180 ± 14
Koga et al. (1995)	1.58 ± 0.61	2.89 ± 0.26	-	-
Lyons et al. (2007)	2.20 ± 0.25	3.10 ± 0.38	-	-
Schneider et al. (2002)	2.08 ± 0.11	3.10 ± 0.14	180 ± 5	193 ± 2
Tulppo et al. (1999)	2.44 ± 0.27	3.70 ± 0.47	178 ± 11	188 ± 13
Older adults				
Loughney et al., (2014) ($ml \cdot kg^{-1} \cdot min^{-1}$)	11.5 (10.7-13.6)	20.8 (18.2–27.9)	-	-
Pogliaghi et al. (2006) (Control)	1.37 ± 0.25	1.89 ± 0.42	143 ± 12	146 ± 9
Pogliaghi et al. (2006) (ACE group)	1.62 ± 0.24	2.31 ± 0.37	149 ± 19	157 ± 13
Pogliaghi et al. (2006) (CE group)	1.84 ± 0.30	2.18 ± 0.28	154 ± 9	158 ± 5
Clinical populations				
Tew et al., (2009) (ACE group)	13.5 ± 2.7	-	121 ± 23	-
Tew et al., (2009) (Control)	13.3 ± 3.5	-	116 ± 24	-
Zwierska et al., (2006)	1.0	1.10	114	113

The smaller muscle mass engaged during ACE compared to CE results in a smaller vascular cross sectional area being perfused by the same cardiac output, which results in a greater total peripheral resistance (Sawka 1986). A greater total peripheral resistance during ACE may also be a result of the greater mechanical compression of the vasculature, since the smaller muscle mass needs to develop a greater percentage of its maximal tension to produce a given power output (Sawka et al. 1983).

2.3.2 iii Respiratory responses

Minute ventilation ($\dot{V}_E$) is greater during ACE compared to CE at the same submaximal power output (Sawka et al. 1982) and oxygen uptake (Bevegard, Freyschuss and Strandal 1966). The greater $\dot{V}_E$ during ACE is accomplished by a greater breath frequency and a lower tidal volume (Pendergast 1989) and is likely the result of a greater blood lactate concentration (Bevegard, Freyschuss and Strandal 1966), greater isometric component (Wiley and Lind 1975) and/or greater neurogenic drive from the upper limbs (Sawka 1986).

2.3.2 iv Metabolic responses

At the same relative intensity of $\dot{V}O_{2PEAK}$ carbohydrate oxidation is greater (thus lower fat oxidation) during ACE compared to CE (Hooker et al. 1990; Kang et al.1999; Helge et al. 2008). The greater carbohydrate oxidation during ACE compared to CE may partly be explained by a lower training status and difference in fibre type in the arms compared to the leg musculature (Helge et al. 2010). The greater reliance on carbohydrate oxidation during ACE may partly be explained by the greater recruitment of type II muscle fibres with arm exercise compared to leg exercise (Orr et al. 2013). It is generally understood that type II muscles fibres are metabolically less efficient (Schneider, Wing and Morris 2002; Koppo, Bouckaert and Jones 2002)

Previous studies have also reported a greater blood lactate concentration during ACE than CE at the same power output (Astrand et al. 1965; Davies and Sargeant 1974). The greater blood lactate concentration during ACE is probably a result of differences in relative exercise intensity (Sawka 1986). For example, when performed at the same relative intensity ranging between 20 – 80 % $\dot{V}O_{2PEAK}$, blood lactate concentrations are similar between ACE and CE (Sawka et al. 1982).

2.3.2 v Perceptual responses

Collectively, the above mentioned findings suggest that exercising with the arms elicits different metabolic responses compared with leg exercise. Therefore it is reasonable to assume that the same applies to subjective perceptions of exertion. Astrand and Rodahl (1970) suggest that perceived exertion might be associated with the metabolic rate per unit mass of active musculature. Indeed, Kang et al. (1998) suggested that the engagement of the smaller upper body musculature during ACE may concentrate perceptions of exertion more than during leg cycling at 50 and 70 % $\dot{V}O_{2PEAK}$. Additional engagement of the trunk and arm musculature during ACE may result in a greater afferent feedback compared to cycling (Ishida et al. 1994), leading to increased perceptions of fatigue. The greater local perceived exertion during ACE compared to CE does not appear to be influenced by age (Aminoff et al. 1997).

2.3.3 Age related reductions in aerobic fitness

There is little doubt that aging is associated with an increase in physiological stress during submaximal exercise and a reduction in $\dot{V}O_{2MAX}$ by approximately 10% per decade (Betik and Hepple 2008; Burskirk and Hodgson 1987; Hawkins and Wiswell 2003; Perini et al. 2002). Extensive focus has been directed towards understanding the central and/or peripheral mechanisms responsible for the reduction in $\dot{V}O_{2MAX}$. For example, maximal cardiac output (Q) reduces with advancing age (Ogawa et al. 1992), which is accompanied by a reduced fraction of cardiac output delivered to the muscle (Ho et al. 1997). The reduced maximal cardiac output is associated with a reduction in maximal heart rate, which reduces by approximately 3-5 % per decade, irrespective of gender (Jackson et al. 1995; 1996) or training status (Hawkins et al. 2001). Maximal stroke volume (SV) also deceases with age in sedentary individuals (Hagberg et al. 1985). The age related structural changes in the heart and blood vessels result in decreased chronotropic response, impaired cardiac function and increased aortic impedance which are responsible for reduced Q and SV with age (Seals et al. 1994). In contrast, endurance training in older adults has been shown to increase diastolic filling (Levy et al. 1993) and increase left ventricular contractility due to an increase in blood plasma volume (Ehsani et al. 1991; Seals et al. 1994)

Several studies have alluded to the mechanisms with age which contribute to a reduced $\dot{V}O_{2MAX}$. Alterations in body composition, namely decreased lean body mass and increased fat mass (Hawkins et al. 2001; Jackson et al. 1995; 1996) appear to be predominant factors responsible for reduced $\dot{V}O_{2MAX}$. Reductions in maximal a-$\overline{v}O_2$ difference are also reported with age, thus reflecting less oxygen utilisation by the (reduced) active musculature (Rivera et al. 1989). Other factors which contribute to reduced exercise capacity with age include a loss of muscular strength, endurance and muscle contractile speed (Fisher, Pendergast and Calkins 1990). Research indicates that the loss of force production in older adults is due to reductions in the percentage of contractile tissue within the muscle (Frontera et al. 2000) and deficits in motor unit recruitment and firing rates (Williams et al. 2002).

2.3.3 i Age related changes in upper and lower body exercise performance

When exercising at the same % of $\dot{V}O_{2PEAK}$ older adults show equal physiological strain compared to younger adults (Davy et al. 1995). However these studies employed exercise using the large muscle groups of the legs (e.g., cycling). Aminoff et al. (1996) reported that healthy older men, with a similar estimated size of upper and lower body muscle mass, demonstrated reduced maximal cycling capacity compared to young men (3.02 ± 0.20 vs. 3.54 ± 0.24 $L \cdot min^{-1}$). In contrast, maximal aerobic capacity during ACE did not differ between older and younger males (2.09 ± 0.18 vs. 2.19 ± 0.32), indicating that the arms remain well preserved in older adults. Aminoff et al. (1997) later reported that 30-min of ACE and CE at

50% and 75% of mode specific $\dot{V}O_{2PEAK}$ elicited a similar degree of physiological strain (HR, RER, $\dot{V}O_2$) between healthy older and young men. Interestingly, during the ACE trials young and older adults exercised at the same absolute exercise intensity (due to a similar W_{MAX}), indicating a similar ability to perform prolonged exercise using the arms with advancing age (Aminoff et al. 1997). Therefore, the peripheral limiting factors of arm exercise are not necessarily accentuated with age in men with a well retained muscle mass.

2.3.4 Physiological adaptations to upper and lower body exercise training

Previous studies examining the physiological responses to upper body exercise training have mainly explored whether training adaptations are muscle specific (Volianitis et al. 2004), the potential cross transfer effects of the trained limbs (i.e., arms) to the untrained limbs (i.e., legs) (Lewis et al., 1980; Loftin et al., 1988; Tordi et al., 2001; Bhambhani et al., 1991; Pogliaghi et al., 2006) and the use of upper body exercise for clinical rehabilitation (Saxton et al. 2011; Tew et al. 2009; Zwierska et al. 2007). However, there appears some discrepancy between authors with regards to the specific improvements in aerobic capacity when training with the upper body. For example, Loftin et al. (1988) noted that improvements in aerobic capacity were dependent upon central adaptations, such as increased cardiac output and stroke volume. In contrast, Volianitis et al. (2004) suggest that endurance training results in predominantly peripheral circulatory adaptations such as arterial–venous oxygen difference, increased blood flow and cellular and enzymatic adaptations. This would imply that training is limb specific. Despite many studies reporting increases in $\dot{V}O_{2PEAK}$ and reductions in submaximal exercising heart rate on mode specific exercise capacity, there have been some studies which report a cross transfer effect of increased $\dot{V}O_{2PEAK}$ or lower HR during submaximal exercise with untrained limbs providing evidence in favour of the central circulatory adaptations to endurance training. Peripheral adaptations are further evidenced by increases in peak power output (W_{PEAK}) correlating with increased bicep circumference when tensed, suggesting hypertrophy of the bicep may have in part, contributed to an increase in peak power (Bottoms and Price 2014).

2.3.4 i Evidence in favour of the cross transfer effect

Several studies have demonstrated a cross-training effect of arm exercise to leg exercise as evidenced by increased $\dot{V}O_{2MAX,}$ or decreased HR during submaximal exercise with the untrained limbs in young (Lewis et al., 1980; Loftin et al., 1988; Tordi et al., 2001) middle aged (Bhambhani et al., 1991), healthy older (Pogliaghi et al., 2006) and clinical populations (Saxton et al. 2011; Tew et al. 2009; Zwierska et al. 2007). Lewis et al. (1980) investigated the effects of ACE and CE training on the $\dot{V}O_{2PEAK}$ attained during upper and lower body exercise in very sedentary young males. Training comprised of four 30-min exercise sessions per week for 11 weeks at an intensity corresponding to 75-80% $\dot{V}O_{2PEAK}$. It is acknowledged that this represents a heavy training intensity for an untrained group; however the authors did not allude to the rationale for such as high training intensity. The arm training group demonstrated significant specific (35%) and cross-transfer (12%) improvements in $\dot{V}O_{2PEAK,}$ while the leg training group showed smaller specific (15%) and cross-transfer (9%) improvements in $\dot{V}O_{2PEAK}$. Loftin et al. (1988) reported that 5-weeks of endurance training using the arms elicits significant improvements in $\dot{V}O_{2PEAK}$ attained during maximal effort arm (32%) and leg (7%) exercise. Moreover, significant improvements in central (cardiac output and stroke volume) and peripheral circulatory (arterial-venous oxygen difference) function and an increased time to exhaustion during both modes of exercise were also observed (Loftin et al. 1988). The abovementioned studies were conducted in young adults who were either untrained or very inactive. Pogliaghi et al. (2006) investigated the effects of ACE training and CE training on $\dot{V}O_{2PEAK}$ and ventilatory threshold in healthy older adults (age, 67 ± 5 years). It was reported that none of the adults were meeting exercise guidelines (less than three times per week, for less that 20-min at an intensity below 50 % HR_{MAX}) as recommended by ACSM (Nelson et al. 2007). Training comprised of three 30-min exercise sessions per week at an intensity corresponding to 90-110% of heart rate achieved at ventilatory threshold. Following training, significant improvements in W_{PEAK} and $\dot{V}O_{2PEAK}$ were reported in both training groups for both modes of exercise. Arm training elicited an increase in $\dot{V}O_{2PEAK}$ of ~19% and ~8% for maximal arm and leg exercise, respectively. Similarly, leg training resulted in a significant increase in $\dot{V}O_{2PEAK}$ of ~22% and ~9% for maximal leg and arm exercise, respectively. The relative improvements in specific and cross transfer effects of arm and leg training in this older population are therefore generally in accordance with previous work in young adults (Lewis et al. 1980; Loftin et al. 1988; Tordi et al., 2001). These studies suggest that the initial fitness of participants is an important variable when determining the extent of cross transfer benefits in untrained limbs (Lewis et al. 1980).

2.3.4 ii Evidence against the cross transfer effect

Those participants who are relatively untrained are able to gain greater improvements in ACE performance compared to those who are trained, regardless of whether they are specifically trained in the upper body. Low pre-training $\dot{V}O_{2PEAK}$ in young (Loftin et al., 1988), older (Pogliaghi et al., 2006; CE 27.9 $ml \cdot kg \cdot min^{-1}$; ACE 21.3 $ml \cdot kg \cdot min^{-1}$) and clinical populations (Tew et al., 2009; ACE 13.4 $ml \cdot kg \cdot min^{-1}$) may explain the greater potential for transfer effects with the untrained limbs. Some studies have provided evidence that training the legs has no effects on arm performance, or vice versa (Clausen et al., 1970; Stamford et al., 1978) therefore discounting crossover training benefits. It is possible that in trained individuals the intensity of upper body training may not have been high enough to elicit a significant training stimulus for the cardiovascular system (i.e., a transferable effect). For example, Magel et al. (1978) reported that treadmill $\dot{V}O_{2PEAK}$ increased only slightly following arm training in those with relatively high initial aerobic capacity (56 – 57 $ml \cdot kg \cdot min^{-1}$). Similarly, Stamford et al. (1978) investigated the effects of high intensity arm or leg training on $\dot{V}O_{2PEAK}$ of the upper and lower body in fit individuals. Mode specific improvements were observed (~ 15 – 20 %) for ACE and CE, while the untrained limbs remained unchanged. If the initial fitness level is relatively high, the potential benefits of arm training seem to be limited to specific improvements (trained muscles) in exercise tolerance. From this perspective, the limited degree of crossover training effects from one set of limbs to another discount the practice of encouraging walking, jogging or cycling exclusively (Franklin 1989). Instead, individuals should be encouraged to train the arms as well as the legs, with the expected attenuation of cardiorespiratory, hemodynamic and perceived exertion in each mode of exercise (Franklin 1989).

2.3.4 iii Central versus peripheral adaptations

While acute responses to ACE have been reported (section 2.3), little data exist regarding upper body exercise training on determinants of improved aerobic capacity (e.g., SV or a-$\overline{v}O_2$). Improvements in $\dot{V}O_{2PEAK}$ following cycling training appear to be more dependent on central rather than peripheral adaptations (Gledhill et al. 1994). Magel et al. (1978) cited improvements in $\dot{V}O_{2PEAK}$ from 34 to 39 ml·kg^{-1}·min^{-1}, which were attributed to a widened a-$\overline{v}O_2$ difference since peak cardiac output, stroke volume and heart rate remained unchanged. This would imply that endurance training results in predominantly peripheral circulatory adaptations, such as a-$\overline{v}O_2$ difference, increased blood flow and cellular and enzymatic adaptations (Volianitis et al 2004). In contrast, the work by Loftin et al. (1988) specifies that ACE training elicited significant improvements in central (cardiac output and stroke volume) and peripheral (a-$\overline{v}O_2$ difference) adaptations which supported improvements in arm and leg $\dot{V}O_{2PEAK}$. Nevertheless, ACE training may not be as effective as CE training in eliciting general effects, since arm exercise is performed a relatively low oxygen uptake and energy expenditure. Training the arms at ~ 70 % of mode specific $\dot{V}O_{2PEAK}$ would typically require less than 50 % of the cycling $\dot{V}O_{2PEAK}$ (Franklin 1989). However, upper body exercise training may yield significant central circulatory capacity during ACE and CE, primarily in individuals who are sedentary or less fit.

2.4 Postural Sway

Human bipedal standing is inherently unstable (Peterka and Loughlin 2004). Since humans have been able to stand upright they have faced the challenge to balance the bodys high centre of mass over a relatively small base of support. This challenging task is apparent in the postural sway that is always present when attempting to stand as still as possible (Duarte and Sternad 2008). During quiet standing the human body continually oscillates in a somewhat stochastic fashion (Collins and De Luca 1994). The seemingly chaotic motion of sway is understood to reflect the interaction between destabilising perturbations acting on the body and the corrective internal responses made by the postural control system (Pavol 2005). Postural sway has been classically defined as the constant small deviations from the vertical and their subsequent correction to which all human beings are subject when standing upright (Sheldon 1963).

Although postural sway is a phenomenon experienced by all humans, the physiological origin of sway has been for many years, and still remains, a subject of debate. For example, postural sway has been suggested to be the result of haemodynamic functions such as cardiac and respiratory muscular contractions (Caron et al. 2004; Conforto et al. 2001; Gandevia et al. 2002; Hodges et al. 2002) neuromuscular noise (e.g., delays and/or errors in motor output) (Fitzpatrick et al. 1992; Ishida and Imai 1980; Kiemel, Oie and Jeka 2002; Loram and Lakie 2002; Maurer and Peterka 2002), physiological tremor (e.g., force fluctuations of isometric contractions of the lower limbs) (Kouzaki and Masani 2012), or inaccuracies in the modulation of calf muscle activity (Loram et al. 2001; 2005; Loram and Lakie 2002).

2.5 Postural Control

Postural control can be defined as the ability of an individual to maintain position of the body, or more specifically its centre of mass, within specific boundaries of space, referred to as stability limits (Shumway-Cook and Woollacott 2011). This definition of postural control is particularly useful as it accentuates the need to discuss stability in the context of a distinct task, such as standing still. However, regardless of the task it is well documented that maintaining postural stability is a highly complex control task for the central nervous system (CNS) achieved by the integration and interaction of neural, sensory and musculoskeletal systems (Duarte and Sterand 2008). Despite the contribution of many systems, our current understanding is that postural control is primarily a sensory function. Disturbance of postural sway induced by sensory alterations suggest that active feedback control mechanisms contribute to corrective motor output based on movements detected by sensory systems (Peterka and Loughlin 2004). Sensory afferents from visual, vestibular, cutaneous and proprioceptive systems provide unique information regarding the actual orientation of the body axis in relation to the desired orientation of the body axis (Deliagina et al. 2007; Horak 2006; Mergner, Maurer and Peterka 2003). Subsequently, a proportional efferent signal is delivered to elicit a corrective and adequate motor response (Deliagina et al. 2007; Massion 1994).

2.5.1 Sensory contributions to standing balance

As already discussed, in order to maintain an upright stance, the central nervous system (CNS) must integrate afferent information from visual, vestibular, proprioceptive and cutaneous systems and modulate commands to the neuromuscular systems (Bisson et al., 2011). Each sensory pathway has a specific activation threshold and sensitivity (Mallau, Vaugoyeau and Assaiante 2010). The following sections will consider the various contributions of sensory systems involved in balance control.

2.5.1 i Proprioceptive System

Proprioception is classically defined as the perception of joint and body movement as well as position of the body, or body segments in space (Sherrington 1906). The term proprioception refers to the cumulative neural input to the CNS from mechanoreceptors located in the joints, ligaments, muscles and tendons (Ribeiro and Oliveira 2007). Proprioception is typically assessed using sense of limb movement (i.e., detection of passive movement of the limb) (Hiemstra, Lo and Fowler 2001). Lower limb proprioception appears to play a key role in providing a representation of the body's position, informing the CNS about the relative position of body segments in relation to others (Massion 1992). It is well known that proprioceptive signals originating from muscle spindles (type Ia and II) and Golgi tendon organs (type Ib) around the ankle and knee joints are sensitive to changes in muscle fascicle length and muscle tension, respectively (Gandevia 1996; Loram et al. 2009). Receptors located in joints, muscles and tendons provide input regarding the configuration and position of the limbs with respect to each other (Hadders-Algra and Carlberg 2008), allowing a representation of the bodies perpendicular position relative to the ground (Windhurst 2007).

Localised muscle fatigue elicits an internal disturbance by disrupting proprioceptive functioning at the ankle (Forestier, Teasdale and Nougier 2002), knee (Skinner et al. 1986), lumbar spine (Taimela, Kankaanpää and Luoto 1999), cervical spine (Vuillerme, Pinsault and Vaillant 2008), elbow (Walsh et al. 2004), shoulder (Björklund et al. 2000) and the neck (Schieppati, Nardone and Schmid 2003). Roll and Roll (1988) suggested that muscle spindles form a proprioceptive chain from the muscles of the eyes to the feet, since applying a vibration at any level of the chain results in increased postural sway. In accordance with the assumption that sensory inputs operate through a proprioceptive chain, a disruption of proprioceptive information in the upper body induced by fatigue (i.e., torso, shoulder or upper limb musculature) may alter postural stability. However, at present little is known about how this multiple proprioceptive information is integrated. Early research by Eklund (1972) recognised that whole body postural sway could be induced by applying a vibratory stimulus to the ankle muscles. It was found that stimulating the tibialis anterior muscle resulted in a forward tilt, while stimulating the triceps surae musculature induced a backward tilt. Vibratory stimulations applied to postural muscles drives muscles spindles creating an illusionary increase in the muscles length (Latash 2008). This misinterpretation of body orientation is subsequently compensated for by an actual change in body position.

In addition to reduced muscular strength and slower neural processing, older adults demonstrate a significant decrease in both cuataneous vibratory stimulation and joint sensations (Diener et al. 1984) resulting in peripheral neuropathy. Indeed, Richardson, Ashton-Miller and Lee (1996) estimate that one in five adults of 65 years of age has evidence of peripheral neuropathy. Advancing age affects both the quantity and quality of Meissner (tactile) and Pacinian corpuscles (nerve endings in skin) (Shumway-Cook and Woollacott 2011). These proprioceptive losses increase the threshold needed to detect movement and decrease precision in reproducing or matching joint angles (Speers, Kuo and Horak 2002). It is widely assumed that these physiological alterations with age result in poorer postural control, although the relationship remains unclear.

2.5.1 ii Visual System

Visual information is considered to be one of the most important sensory inputs for postural control (Uchiyama and Demura 2008). During body sway, properties of the optic flow pattern at the eye detect the motion and position of body segments relative to each other and the environment (Lord and Menz 2000). For example, when you sway forward, visual afferents provide feedback of surrounding objectives moving in the opposite direction (Shumway-Cook and Woollacott 2011). Visual inputs may also provide a frame of reference for verticality (i.e., right angle to the horizon). This is essential, since we are faced with many structures which are vertically aligned such as doors, walls and windows (Shumway-Cook and Woollacott 2011).

Visual inputs include both peripheral and central visual information. Central vision is assumed to be responsible for detecting the physical characteristics of objects, while peripheral vision is concerned with detecting spatial characteristics of the surrounding environment (Schmidt & Lee, 1999). The contribution of vision for postural control is most easily demonstrated by increases in postural sway of 20 - 70% under eyes closed conditions (Fitzpatrick and McCloskey 1994; Lord and Menz 2000) (Figure 2.1). The contribution of visual input for standing balance is also evidenced by investigations which have employed moving visual fields inducing a sense of self motion, therefore when visual information is erroneous (e.g., sitting in a stationary car and the vehicle next you moves, resulting in reaching for the brake) significant increases in sway have been reported (Lee and Lishman 1975). Therefore, the CNS may also misinterpret visual information.

Advancing age is associated with a reduction in visual performance and has been strongly linked to the observed increase in postural sway among older adults by a number of authors (e.g., Lord and Menz 2000). Litchtenstein et al. (1988) found that poor visual acuity (clearness of vision) was associated with increases in centre of pressure (COP) area. While Lord, Clark and Webster (1991) reported that visual acuity was not a predictor of postural sway on a firm surface but was a strong predictor when standing on a compliant rubber surface. These findings suggest that vision is particularly important in upright stance under more challenging conditions as visual information is relied upon to detect greater body sway.

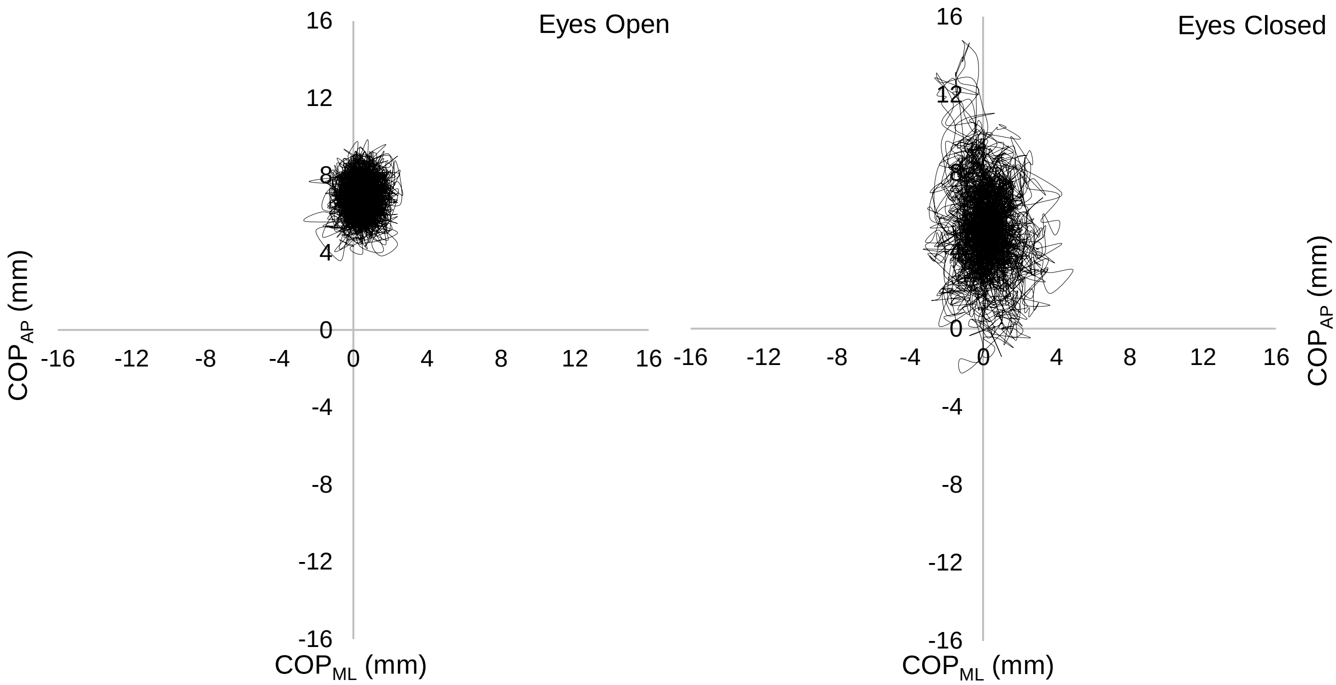

Figure 2.1: Statokinesigram showing the map of the COP displacement in the anteroposterior (AP) and mediolateral (ML) directions during eyes open and eyes closed conditions in a young healthy adult (Note data taken from Chapter 6).

2.5.1 iii Vestibular system

It is widely acknowledged that the vestibular system is strongly linked to the perception of body orientation (Fitzpatrick and McCloskey 1994). The vestibular system provides information related to linear acceleration and angular velocity of the head in relation to gravity (Nashner, Black and Wall 1982). The semi-circular canals provide information regarding angular velocity in three dimensions while the otolith organs are primarily involved in the detection of linear acceleration and the heads position in relation to the earth vertical (Mergner, Maurer and Peterka 2003). However, the role of vestibular information in quiet stance is somewhat controversial, with increasing research suggesting that the vestibular system does not contribute significantly to postural control during quiet standing (Fitzpatrick and McCloskey 1994), in particular when visual and proprioceptive information are readily available (Nashner, Black and Wall 1982). These studies highlight that with only vestibular afferents, the CNS cannot distinguish between whole body sway and simple head movements such as looking at the ground which could be misinterpreted as excessive anterior body sway. Advancing age decreases the number of hair cells in both the semi-circular canals and the otolith organs, in addition to the number of nerve fibres in the vestibular nerve, resulting in a reduction in vestibular excitability (Speers, Kuo and Horak 2002).

2.5.1iv Plantar cutaneous mechanoreceptors

While numerous studies have focused on the role of proprioceptive, visual and vestibular sensory information on balance control, relatively little information is known about the role of plantar cutaneous mechanoreceptors from the soles of the feet. When considering that the cutaneous mechanoreceptors on the surface of the feet are the boundary between the body and the ground, this sensory information might play an important role in balance control (Kavounoudias, Roll and Roll 1998). The information from cutaneous mechanoreceptors that innervate hairless skin are derived from the Meissner corpuscles, Pancnian corpuscles, Markel disks and Ruffini endings (Shaffer and Harrison 2007) and deliver information with reference to the location and force of weight bearing activities (Kavounoudias, Roll and Roll 1998; Perry 2006). The contribution of cutaneous mechanoreceptors for balance control are largely based on methods which either anaesthetise (Meyer, Oddsson and De Luca 2004) or cool the foot (Magnusson, Johnasson and Pyykko 1990) and change characteristics of the support surface, such as standing on compliant foam (Tanaka and Uetake 2005). The former approaches have received significantly less interest due to the potential discomfort caused to participants. In contrast, more recently postural responses to standing on a foam block have received extensive interest, probably due to the relative ease of this approach (Lord and Menz 2000; Vuillerme et al. 2001; Bisson et al. 2014).

Several studies have reported that when proprioceptive input is affected (e.g., when standing on foam) the fatiguing effects of exercise on postural stability can be accentuated (Bisson et al. 2012; 2014) as a result of altered joint position sense (Allen et al. 2010) and force sense (Vuillerme and Boisgontier 2008). Furthermore, some weight bearing exercises such as running or walking may damage the plantar cutaneous receptors because of repetitive compression of the sole of the foot during contact with the ground (Lepers et al. 1997). In this context it is also possible that the mechanoreceptors of the feet might be stimulated and affected differently by ACE and CE. For example, during ACE the feet are typically flat to the ground for the entire duration of exercise. However, during CE the soles of the feet are intermittently stimulated during each duty cycle Therefore, plantar cutaneous mechanoreceptors appear to play a vital role relating the position of foot pressure and subsequent postural sway, by creating an important link between the foot and proximal muscles (Gribble et al. 2004).

2.5.1 v Sensory integration within the central nervous system

Postural control is an automatic process controlled by neural mechanisms (Deliagina et al. 2007). Structures of the CNS which are known to contribute significantly to postural control include the basal ganglia, cerebellum, cerebral cortex, secondary motor cortex, brain stem and the spinal cord (Hadders-Algra and Carlberg 2008; Deliagina et al. 2006; 2007; 2008). Each structure of the CNS has its own unique function and contribution to maintain postural equilibrium. The key function of the nervous system is to receive and interpret all sensory information and to deliver a motor command to elicit corrective postural responses (Deliagina et al. 2007; Massion 1994). Little is known about how afferent information from the various sensory systems is processed and combined to generate appropriate corrective responses when the information provided by the sensory systems is conflicting. According to Peterka (2001), one possibility is that sensory cues are combined in a linear manner. More specifically, each sensory system detects an error indicating a change in body orientation from a normal reference position. The individual errors from these sensory systems are summed to yield a single error and appropriate corrective motor responses are generated as a function of the summed signal (Peterka 2001).

2.5.2 Motor Control of Standing

Three main factors contribute to motor control during quiet stance; (a) intrinsic stiffness of skeletal muscles and muscle tone due to neural contributions, (b) the activation of anti-gravity muscles during quiet stance (Basmajian and Deluca 1985; Shumway-Cook and Woollacott 2011) and (c) specific movement patters. Indeed, all three of these factors could be acutely and chronically affected by exercise bouts.

2.5.2 i Intrinsic Muscle-Tendon Stiffness

While few would argue that the ankle musculature is integral for postural equilibrium (Loram, Maganaris and Lakie 2005), it is suggested that the upright posture is mostly maintained by the passive stiffness of the musculotendinous structure of the human body, which is typically viewed as the body's first line of defence against falling (Duarte & Freitas 2010; Latash 2008; Winter et al., 1998). Research by Winter et al. (1998) suggested that postural adjustments are primarily controlled by the spring like action of the ankle muscles with the stiffness reflecting the mechanical properties of the muscles. The basic premise of this model is that during quiet standing the human biped should be modelled as an inverted pendulum, with the ankle the only moveable joint. However, Loram and Lakie (2002) suggest that without the proactive control of the calf musculature, it can be expected that during quiet standing a person would fall or would need to initiate a step.

The inherent passive stiffness of the human is supplemented with low level tonic muscle activity (continual activation) of a number of postural muscles (e.g., triceps surae and erector spinae) (Nashner 1983). Muscle tone specifically refers to the force at which a muscle resists lengthening (i.e., its stiffness) (Basmajian and Deluca 1985). While a certain degree of muscle tone is always present in a relaxed state, no electromyographic (EMG) activity is recorded (Shumway-Cook and Woollacott 2011). Numerous authors have alluded to the many muscles which are tonically active during quiet standing (Laughton et al. 2003; Loram et al., 2005; Joseph and Nightingale 1954, 1956; Basmajian and DeLuca 1985; O'Sullivan et al. 2002; Woodhull-McNeal 1986). These muscles are depicted in Figure 2.2 and include but are not limited to the triceps surae (soleus and gastrocnemius), tibilais anterior, gluteus medius, tensor fascia late, iliopsoas, thoracic erector spinae and abdominals.

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Figure 2.2: Muscles that are tonically active during the control of quiet stance (Shumway-Cook and Woollacott 2011, pp. 167).

2.5.3 Postural strategies

Since quiet standing is accompanied by spontaneous postural sway, postural alignment and postural tone are unable to provide complete stability and therefore movement strategies are required to maintain the centre of mass within the base of support (Shumway-Cook and

Woollacott 2011). Maintaining a quiet standing position requires coordinated muscle action in order to produce appropriate muscular torque (Era et al. 1996). Indeed, there is little doubt that postural sway is primarily controlled by the lower body and trunk musculature (Laughton et al., 2003; Loram et al., 2005). Early research by Nashner and McCollum (1985) discussed the existence of two strategies which could be used alone or combined by the nervous system to control the centre of mass in the sagittal plane. These strategies were termed the ankle strategy and hip strategy (Figure 2.3). More recently, research has shown that postural control is much more complex than a single segment inverted pendulum, with the body behaving more like a multilink pendulum (legs and trunk) with two coexisting modes of control (Creath et al. 2005). According to Shumway-Cook and Woollacott (2011), the CNS can move back and forth between these control modes, with one strategy predominating over the other depending on the sensory information available and task conditions. During quiet standing, humans exhibit a natural sway in the forward and backwards (anteroposterior) and side to side (mediolateral) directions.

2.5.3 i Control of anteroposterior sway

Ankle strategy

According to the ankle strategy, the whole body is represented as a single segment inverted pendulum and ankle plantar and dorsi flexors act alone to control the inverted pendulum (Fujisawa et al., 2005; Winter 1995). In the case on anterior sway an increase in muscle activity of the posterior leg and trunk muscles are observed. In contrast, during posterior sway, an increase in the activity of anterior muscles is reported (Horak and Nashner 1986) (Figure 2.3). During upright standing, these muscular responses occur at a latency of 90 to 100 ms in soleus and gastrocnemius muscles, followed by activation of the hamstrings 20 – 30 ms later which precedes activation of the paraspinal muscles (Nashner 1989) in the case of anterior sway. When swaying in the posterior direction, muscle activity begins in the distal muscle (tibialis anterior) following by activation of the quadriceps and abdominal muscles (Nashner 1989). This strategy is mostly used when the support surface is large, with minimal internal or external disturbance (i.e., quiet bipedal stance) (Winter 1995; Fujisawa et al. 2005; Horak 2006; Horak and Nashner 1986).

Hip strategy

In contrast, when the postural perturbations are large, or the support surface is narrow or compliant, inclination angles become so great that the ankle strategy is not able to restore the centre of gravity (Fujisawa et al. 2005; Horak and Nashner 1986). Under these conditions, the hip strategy rotates the body about the sagittal plane as a two-segment inverted pendulum, with a smaller torque produced at the ankle joint (Runge et al. 1999). Under the hip strategy the muscle activation occurs in a proximal to distal order (Woollacott and Shumway-Cook 2011).

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Figure 2.3: Standing control during quiet standing using either ankle strategy, hip strategy, or a combination of both (Winter et al., 1995)

2.5.3 ii Control of mediolateral sway

More recent work has described that the control strategy for ML stability is different (Winter et al., 1996). For example, lower limb ankle and knee joints can produce very little ML movement (Shumway-Cook and Woollacott 2011). Therefore, the hip is the primary joint used when controlling ML stability (Day et al., 1993; Winter et al., 1996). Winter and colleagues (1993; 1996) reported that the loading and unloading of the right and left limb appeared as mirror images, with the weight unloaded on one side loaded by the other. A number of authors have reported different musculature controlling ML sway. For example, both the hip abductors (gluteus medius and tensor fascia latae) and adductor muscles have been suggested to be involved in the loading and unloading of the legs (Maki et al., 1994; Moore et al. 1988). In contrast with AP muscle control patterns, ML muscle patterns are organised in a proximal to distal order, where the hip musculature is activated before ankle muscles (Moore et al. 1988).

The abovementioned postural movement strategies are used in the form of compensatory mechanisms (feedback) and anticipatory mechanisms (feedforward) (Horak and Macperson 1996; Park Horak and Kuo 2004). Feedback control mechanisms are activated by different sensory systems in order to compensate for unexpected deviations from a desired position (Horak and Macpherson 1996). Feed-forward mechanisms are used to predict perturbations (i.e., surface translation when standing on a moving bus) and elicit pre-programmed postural responses in order to maintain postural stability (Massion et al. 1997). According to Horak and Macpherson (1996) these two systems normally interact. For example, inaccuracy of the anticipatory mechanisms can be counteracted by the compensatory mechanisms.

2.5.4 Age related changes in motor control

Several studies have reported age-related changes in the pattern of muscle activation of the lower limbs during quiet standing (Laughton et al. 2003; Nagai et al. 2011; 2013; Tucker et al.

2008). Recent research by Nagai et al., (2011) revealed that muscle co-activation at the ankle joint was greater in older compared to younger adults with older adults activated two to five times more of their biceps femoris and tibialis anterior muscles than young adults (Laughton et al. 2003). It is presently unclear however, whether increases in muscle co-activation preclude or elicit postural instability. It has been postulated that increased muscle co-activation in older adults is employed as a compensatory mechanism for increased postural sway accompanied by healthy aging (Tucker et al. 2008). On the other hand, excessive muscle co-activation which increases postural rigidity (Tang and Woollacott 1998) may increase postural sway and the risk of falling. It remains unknown as to whether increased co-activation causes increased postural sway or whether an increase in postural sway elicits a greater co-activation. Increased rigidity induced by disproportionate levels of muscle co-activation reduces the degrees of freedom of the postural control system (Tucker et al. 2008) and may compromise voluntary postural responses (Allum et al. 2002).

2.6 Measuring Postural Sway

2.6.1 Measurement apparatus

Several experimental techniques have been developed to measure postural sway during quiet standing. However, there is no single measurement technique that could be considered as a 'gold standard' to assess the integrity of the postural control system (Winter et al., 1990). During quiet standing postural sway is minimised by movements of different body segments and accomplished through the neuromuscular system (Bisson et al., 2011). Therefore, three distinct measurement techniques are adopted in the literature and include; (a) kinematic analysis (motion analysis) which is interested in the movements rather than the forces (e.g., Benvenuti et al., 1999) (b) neuromuscular changes associated with muscle activation (electromyography) (e.g., Nagai et al., 2011) and (c) kinetic data (force platform) recording the resultant ground reaction forces ensued by kinematic movements (e.g., Pinsault and Vuillerme 2009).

2.6.1 i Motion Analysis

Using kinematic motion analysis it is possible to observe single joint angles, joint velocities and accelerations during quiet standing and the data can be used to reconstruct the body's movement in space (Winter 2009). However, few studies adopt motion analysis for balance assessment likely because the data collection procedures, acquisition and analysis are labour intensive and take a lot of time. In postural control studies motion analysis systems have been used for the purpose of tracking the position of body segments (Benvenuti et al. 1999) and estimating the position of the centre of gravity (Winter et al. 1998). This method requires modelling of the body which during quiet standing is particularly well suited for such modelling (Kuo 1995).

There are several limitations which should be considered when using motion analysis to record body movements; positioning of the markers relative to anatomical landmarks and/or joints and the use of an adequate number of markers. Unlike centre of pressure (COP) data recorded using a force platform there are no representative or reliability data available for kinematic analysis during quiet standing. Furthermore, multiple camera systems are expensive and often cannot be used in daylight and therefore have to be collected in blacked out room conditions (Winter 2009).

2.6.1 ii Electromyography

Electromyographic data acquisition is used to assess the electrical signal associated with muscle contraction (Winter 2008). Surface EMG is typically used to identify muscles which are active during quiet standing and the time and intensity of contraction (Shumway-Cook and

Woollacott 2011). Studies measuring EMG during quiet standing have focused on muscle co-activation (Nagai et al., 2011; 2013), the magnitude of muscle activation during a particular task (Laughton et al., 2003), muscle latencies in response to platform perturbations (Nashner 1989; Ntousis et al., 2013) and anticipatory postural adjustments prior to a voluntary movement (Santos et al., 2010a; 2010b; Vuillerme, Nougier and Teasdale 2002) or during voluntary sway (Loram et al., 2004). Although the EMG can be relatively easily acquired a range of factors can alter the EMG amplitude. For example, the amount of subcutaneous tissue, fibre types / diameter and blood flow within muscle tissue are unable to be controlled for (Rainoldi et al., 2004). As a result, it is necessary to normalise the EMG signal to a standard value (Burden and Bartlett 1999). While it is agreed that EMG amplitudes must be normalised, little consensus exists in the literature for how this should be accomplished. For example, research has investigated the use of maximal voluntary isometric contraction (MVIC) submaximal isometric contraction and peak and mean amplitude during dynamic movement (Soderberg and Knutson 2000). Those studies which have investigated the amplitude of muscle activation during quiet standing have typically normalised data to MVIC (Laughton et al., 2003; Nagai et al., 2011; 2013; Ntousis et al., 2013).

2.6.1 iii Force Platform

Postural stability during quiet standing is most commonly assessed with centre of pressure (COP) measures recorded using a force platform (Doyle et al., 2007; Lafond et al., 2004; Lin et al., 2008; Pinsault and Vuillerme 2009; Santos et al., 2008). The basic premise of force platform assessment is to measure displacements of the centre of pressure (COP) that reflect the horizontal location of the centre of mass (COM). The use of force platform for balance assessment will be discussed in section 2.8.2.

2.6.2 Centre of pressure as a representation of postural sway

Although postural sway is a kinematic term it is frequently assessed using centre of pressure (COP) measures calculated using a force platform (Piirtola & Era 2006). In the context of postural sway, the COP can be viewed as the resultant of multiple forces occurring over the plantar surface of the foot as a result of sway (Hamilton and Luttgens 2002), and reflects the activity of the postural control system. It is generally agreed that increased magnitude (COP displacement) and amount (COP path length) of postural sway represents a decline in postural stability. Changes in the COP position change the moment of the resultant forces acting on the body resulting in rotational actions of the proximal body segments (Latash 2008). As a result, we can view shifts in the COP as the means of moving the COM. Different characteristics of postural sway have been studied including standard one-dimensional ML and AP centre of pressure variables (i.e., standard deviation of each variable, range, mean and maximal velocity) and two-dimensional COP variables (i.e., area, path length and mean velocity) over a fixed time interval. More sophisticated measures (i.e., time series analysis, wavelet transformation, multiscale entropy and fractal analysis) address the temporal structure of postural sway (Collins and De Luca 1994; Kirchner et al., 2012) (Figure 2.6). The former are traditionally used as an index of postural sway, particularly those who have investigated the effects of fatigue.

Small amplitude, low velocity shifts in the COP during quiet standing are considered to indicate the effectiveness of the postural control system as little effort is required to maintain equilibrium (Era et al. 1996). There is no general agreement regarding which variables relating to COP should be adopted for the assessment of postural sway (Duarte & Freitas 2010). Both COP_{AP} and COP_{ML} have been shown to be able to predict future risk of falling (Stel et al. 2004) and have both been shown to be altered by muscular fatigue.

According to Piirtola and Era (2006) ML sway variables (range and velocity) may have a predictive value for subsequent falls in older adults and are used in many clinical assessments. Centre of pressure velocity (COP_V) is a time-dependent sway variable which refers to the total distance covered by the COP path (COP_L) divided by the sampling duration. According to Maki et al. (1990) sway velocity is proportional to the amount of muscle activity required to maintain postural stability. Sway velocity is generally considered to be the most reliable and sensitive parameter for detecting changes in postural sway, particularly among older adults (Lafond et al. 2004; Raymakers et al. 2005). Moreover, good postural stability during standing is characterised by a small path length (low COP velocity) which indicates low energy consumption (Gauchard et al. 2002). Centre of pressure path length has been shown to be a strong predictor of falls in previous prospective studies (Lord, Clark and Webster 1991).

Several studies have reported the use of a confidence ellipse (Rocchi et al., 2005). The most accepted method is the analysis of 95% confidence ellipse (a region that covers the COP with a given probability) of postural sway (Pinsault and Vuillerme 2009). However, the confidence ellipse is subject to random fluctuations and is also dependent on the sampling duration and sampling frequency (Batschelet et al., 1981).

Previous studies which have examined the effects of exercise on postural sway in young and older adults have generally reported standard one and two-dimensional COP variables. Therefore, the present thesis will report standard measures of postural sway in an attempt to compare results across studies as these measures are considered sensitive and robust enough to explore postural sway performance following exercise (e.g., Donath et al. 2013; Gauchard et al. 2002; Mello, Oliveira and Nadal 2010; Nardone et al. 1997; Stemplewski et al. 2012; 2013; Vuillerme and Hintzy 2007). Moreover, the aforementioned variables have received extensive interest in the context of protocol standardisation and reliability (Doyle et al. 2007, Lafond et al. 2004, Le Clair and Riach 1996, Pinsault and Vuillerme 2009; Santos et al. 2008).

2.6.3 Reliability of Centre of Pressure Measures

Although COP measures are most commonly used for assessing standing balance, little standardisation exists in data collection procedures and the way in which COP data are analysed. As a result, it is very difficult to compare data across studies due to differences in adopted protocols. As such, reliability represents an essential requirement for any postural sway outcome measure to ensure that differences in the COP reflect real changes in postural control rather than random or systematic error in the measurement procedure (Pinsault and Vuillerme 2009).

Few studies have investigated the test-retest reliability of COP measures of postural sway (Table 2.2). There are three explicit parameters of the experimental procedure which can affect the reliability of the COP measures; (a) instructions to the individual, (b) sampling duration and (c) number of trials recorded (Pinsault and Vuillerme 2009). With regards to instructions to participants, Zok Mazzà and Cappozzo (2008) reported that stating 'stand as still as possible' elicited a more consistent COP displacement than 'stand quietly', suggesting that the former instructions might be more suitable. While sampling durations in the literature range from 10 to 120 s it is generally agreed that a minimum of 30 s is required to achieved acceptable reliability (Doyle et al., 2005; Goldie, Bach and Evans 1989; Lafond et al., 2004; Le Claire and Riach 1996; Lin et al., 2008; Santos et al., 2008). To the author's knowledge, only one study has determined the test-retest reliability of sampling duration and number of trials (Pinsault and Vuillerme 2009). It was reported that three, 30 s trials are required to achieve excellent reliability of 12 COP measures widely adopted in the literature (namely COP area with a 95% ellipse, total COP range, anteroposterior and mediolateral COP displacement, mean and maximal COP velocity and COP path length).

The placement and orientation of the feet are seldom standardised for postural sway assessment. Most studies ensure repeatability of foot placement between trials by outlining the feet on a paper template placed on top of the force platform (Lin et al., 2008). Pinsault and Vuillerme (2009) requested participants to stand barefoot with feet abducted at 30 degrees and heels separated by 3 cm. The International Society of Posturography (ISP) recommends that participants stand with their feet at 30 degrees with heels together (Kapteyn et al., 1983). Another widely adopted approach is the Romberg stance, which requires feet to be placed together side by side (Black et al., 1982). However, these positions generally fall outside commonly reported comfortable standing positions. McIlroy and Maki (1997) suggest that constraining the feet outside of the participants preferred position could affect postural stability. At present foot position remains open to debate and currently there is no universally accepted standing position. Most studies appear to adopt pelvic and shoulder width position which may not be reproducible in repeated measures.

2.6.4 Postural sway characteristics in older adults

There are several characteristic changes in postural sway with advancing age that are reported to be linked with increased fall risk. Numerous studies have reported that dynamic components of postural sway, such as COP mean velocity, are faster in older compared to younger adults (Kouzaki and Shinohara 2010). The occurrence of falls however, appears to be very strongly associated with ML stability (Maki, Hollida and Topper 1994). It has been extensively reported that ML sway amplitude and velocity is an independent predictor of falls in older adults (Maki, Holiday and Topper 1994; Piirtola and Era 2006; Raymakers et al. 2005; Stel et al. 2003). Mediolateral instability is also associated with previous falls (Lord et al., 1999). Falling in the ML direction often causes hip fractures, thus resulting in a reduced quality of life (Tinetti, Speechley & Ginter 1988). Although there is growing evidence for the existence of a directional vulnerability to ML postural control and fall risk the underlying mechanisms are currently unclear. Video observations of naturally occurring falls have revealed that older adults have problems controlling mediolateral stability during sideway falls (Rogers and Millie 2003). In their review, Rogers and Millie (2003) suggest that aging reduces the ability to produce rapid and appropriately controlled muscle actions and postural movements involving ML hip joint adductor and abductor musculature with reduced control of the trunk segment. Therefore, it is reasonable to assume that specific musculoskeletal factors changing with age compromise ML stability.

2.6.5 Other measures of postural control

The measurement of postural sway during bipedal stance is the most popular assessment technique to determine the effects of fatigue on postural control (Bove et al. 2007; Derave et al. 1998; Derave et al. 2002; Donath et al. 2013; Gauchard et al. 2002; Lepers et al. 1997; Mello, Oliveira and Nadal 2010; Stemplewski et al. 2012; 2013). The popularity of posturography is likely because this type of assessment can overcome many of the draw backs of functional clinical examinations, such as large variability within and between sessions, subjective nature of scoring and poor sensitivity to small changes (Visser et al. 2008). In contrast, despite its excellent sensitivity (Mancini and Horak 2010) postural sway has poor specificity (Diener et al. 1989). As postural sway is a complex behaviour involving many central and peripheral nervous systems and musculoskeletal systems, it can often be difficult to determine why postural sway characteristics have changed (Mancini and Horak 2010). Other postural measurements have therefore been used to render the balance task more difficult. For example, some studies have reduced the size of the base of support (Bisson et al. 2010; Bruniera, Rogerio and Rodacki 2013; Douris et al. 2011), decreasing proprioceptive feedback (compliant surface) (Bisson et al. 2012) or maintaining single limb stance while completing a dynamic task (Gribble et al. 2004).

Table 2.2: Summary of available literature regarding reliability of COP measures used to assess postural sway during quiet standing

Author(s)	Cohort	Condition	COP Variables	Duration (s)	Trials	Statistics	Results
Bauer et al., (2008)	Healthy older adults, N = 64	Feet together and bipedal standing, EO & EC	Path length, area, AP displacement	30	3	ICC	EO; ICC = .841 – .945 EC; ICC = .710 – .946
Goldie, Bach and Evans (1989)	Healthy young adults, N = 28	One leg and two leg standing, EO & EC	AP and ML displacement	32	1	Linear regression	Two leg EO; 0.30 (ML) and 0.11 (AP)
Doyle et al., (2005)	Healthy young adults, N = 30	Bipedal standing foam and firm, EO & EC	Maximum velocity, area, range	10	3	ICC *G*-coefficient	Firm surface EC; ICC = 0.51
Lafond et al., (2004)	Healthy older adult, N = 7	Bipedal standing EO	Area, range, RMS, mean velocity	30,60, 120	9	ICC	1 x 30 s trial; ICC =area (0.22), ML displacement (0.44), AP displacement (0.29), ML (0.87) and AP (0.73) velocity
Le Claire and Riach (1996)	Healthy older adult, N = 25	Feet together and Romberg, EO & EC	Mean velocity	10,20, 30,45, 60	1	Reliability coefficient	Mean velocity (0.84)
Lin et al., (2008)	Healthy young and older adults, N = 32	Bipedal standing, EC	Mean velocity, RMS, area	60	3	ICC	ICC= Young (0.41 – 0.91), older (0.57 – 0.95)
Pinsault and Vuillerme (2009)	Healthy young adults, N = 10	Bipedal standing, EO & EC	Area, range, AP and ML displacement, mean and maximal velocity and path length	30	10	ICC	3 x 30 s ICC = area (0.82), ML displacement (0.76), AP displacement (0.87), velocity (0.84)

Note: Not all variables and results are presented in this table. Abbreviation; EO = eyes open; EC = eyes closed; AP = anteroposterior; ML = mediolateral; ICC= Intra-class correlation coefficients; RMS = Root mean square

Typically, balance assessments can be divided into two main approaches; functional assessment and quantitative assessment (Horak 1997). Functional balance assessments are helpful to document changes in balance with intervention (Mancini and Horak 2010). Functional balance testing are typically subjective and usually involve rating performance of motor tasks on a three to five point scale or stop watch (Horak 1997). The most widely cited functional balance tests include the Activities-Specific Balance Confidence Scale (ABC) (Powel and Meyers 1995), Berg Balance Scale (BBS) (Berg et al. 1992), Tinetti Balance and Gait Assessment (Tinetti 1986), Timed Up and Go Test (TUG) (Mathias, Nayak and Isaacs 1986) and the Functional Reach Test (FRT) (Duncan et al. 1990). While the aforementioned assessments are reliable and valid and typically quick, simple and easy to administer and can detect whether a balance problem exists, many of these tests are subject to ceiling effects, are not comprehensive, not objective and cannot detect the type of balance problem. While functional balance assessment may be useful to monitor balance status following an intervention and can detect whether a balance problem is present, the low sensitivity and often time extensive nature of these tests do not particularly lend themselves to studies determining the acute effects of exercise on balance. Due to the transient nature of the effects of exercise on postural sway lasting only a few minutes, it is not feasible to measure multiple postural tasks post exercise. Nevertheless, the author acknowledges the need to measure multiple aspects of balance which might better lend itself to exercise training interventions.

2.7 Postural sway responses to acute exercise

Postural stability is considered a sensorimotor process (Horak 2006) which includes (a) the integration of sensory afferent information from visual, vestibular and somatosensory systems, (b) central processing of sensory information, and (c) selecting motor responses (Vuillerme and Hintzy 2007; Lepers et al., 1997). Exercise affects all three levels of this sensorimotor process (Lepers et al., 1997). In recent years, a growing number of studies have explored the acute effects of muscular fatigue on the regulation of postural sway during quiet upright standing (review by Paillard 2012). It has been shown that cycling (CE) (Derave et al. 1998; Gauchard et al. 2002; Mello, Oliveira and Nadal 2010; Nardone et al. 1997), treadmill running and/or walking (Bove et al. 2007; Derave et al. 2002; Lepers et al. 1997; Nardone et al. 1997, 1998; Strang, Choi and Berg 2008; Thomas, Van Lunen and Morrison 2012), rowing (Springer and Pincivero 2009), triathlon events (Burdet and Rougier 2004; Nagy et al. 2004) and, more recently, arm crank ergometry (Douris et al., 2011) negatively affect postural stability.

2.7.1 Lower Body Exercise

The immediate effects of cycling exercise on postural sway during quiet bipedal standing have received growing interest in recent years. Although comparisons across studies are challenging due to the differences in the exercise protocols (e.g., intensity and duration) and postural sway measurement procedures, it is commonly observed that CE impairs the ability to minimise postural sway in healthy young adults (Derave et al., 1998; Demura and Uchiyama 2009; Gouchard et al. 2002; Lion et al., 2010; Mello, Oliveria and Nadal 2010; Nardone et al., 1997; Vuillerme and Hintzy 2007). The magnitude of effect on postural sway appears to be proportional to exercise duration and/or intensity. For example, Gouchard et al. (2002) reported significant increases in COP_L (~ 20 %) following 45 min cycling at 60% $\dot{V}O_{2MAX}$ with larger increases (100 %) observed following a maximal oxygen uptake test ($\dot{V}O_{2MAX}$). Similarly, Mello, Oliveria and Nadal (2010) reported that cycling for 60 min at a power output corresponding to 70% of ventilatory threshold resulted in small destabilising effects on mean COP_V (~ 1.20 %), with larger negative effects reported following a $\dot{V}O_{2MAX}$ test (~ 7.90 %). The former likely reflects normal movement variation during quiet standing. Furthermore, Nardone et al. (1997) reported that 25 min of cycling just above the lactate threshold elicited small destabilising effects on COP_L during eyes open (~ 115 %) and eyes closed conditions (~ 125 %). No significant effects of cycling were observed following cycling for 25-min below 60 % HR_{MAX}. Postural sway responses to cycling therefore appear to be proportional to the intensity and/or duration of exercise. According to Paillard (2012), large negative effects of cycling on postural sway may be observed when the exercise intensity is moderate, but the duration must be relatively long. For example, Derave et al., (1998) reported a ~ 16 % increase in COP velocity following 2 h of cycling at 60 % $\dot{V}O_{2MAX}$. In contrast, the same exercise test which included the ingestion of 1.9 L of a carbohydrate-electrolyte solution removed the negative effects of exercise on sway. Mean fluid loss was 2.7 ± 0.4 and 0.5 ± 0.5 % body mass following non-hydration and hydration trials, respectively. The discrepancy in their findings were not explained. Currently, it appears that a loss of vestibular function linked to exercise results at least in part due to dehydration (Lion et al. 2010; Paillard et al. 2012).

In all of the aforementioned studies, increases in COP_{AP} were reported. This is not surprising when considering that cycle ergometry recruits sagittal plane muscles of the lower extremity (Ericson et al., 1985) which also control AP sway. Therefore, fatiguing exercise of the sagittal plane movers of the lower body musculature (i.e., ankle plantar and dorsi flexors) impair postural sway in the sagittal plane only, supporting the evidence of directionally sensitive activity of postural muscles (Winter et al., 1996).

For a given exercise intensity, treadmill exercise deteriorates postural stability to a greater extent than cycling. Nardone et al. (1997) reported that 25 min of treadmill running above the anaerobic threshold had a greater negative effect on COP_L (~ 132 %) compared to cycling at the same intensity (~ 115 %). It has been suggested that running induces stronger eccentric contractions of the ankle musculature, while the calf muscles mainly experience concentric contraction during cycling (Nardone et al. 1997). Eccentric muscle contractions induce greater muscle damage (Vissing et al. 2008) and proprioceptive alterations (Paschalis et al. 2007) compared to concentric contractions, therefore inducing greater disturbances to postural stability.

The impact of exercise on postural sway in older adults may be more pronounced than in young adults since their proprioceptive and neuromuscular systems are less efficient (Bisson et al. 2014). Surprisingly, the effects of acute lower body exercise on postural sway in healthy older adults have only received interest in recent years. Generally, a deterioration in postural stability has been reported following acute cycling (Maciaszek, Stemplewski & Osinski 2010; Stemplewski et al. 2012; 2013) and treadmill exercise (Donath et al. 2013). While postural sway increases with age, exercise does not appear to affect postural stability more in older compared to younger adults (Bisson et al. 2014; Granacher et al. 2010). Older adults use compensatory strategies as effectively as younger adults to maintain stability after fatiguing exercise (Bellew et al. 2009). While young adults preferentially adopt an ankle strategy during quiet standing (Horak and Nashner 1986), older adults tend to adopt a hip strategy (Woollacott and Shumway Cook 1986). Therefore the particular aspects of balance which are affected by exercise may not be the same.

To the best of the author's knowledge, Egerton and colleagues were the first to explore the effects of moderate intensity exercise on postural sway in healthy older adults (Egerton, Brauer & Cresswell 2009). It was reported that 14 min moderate intensity exercise (walking, stepping onto blocks in forward and lateral directions, lunges, mini-squats, stepping over obstacles and carrying bags) increased COP_{ML} by 5 %, while COP_{AP} remained unchanged relative to pre exercise conditions. It was further demonstrated that fallers compared to non-fallers responded to the exercise in a similar manner. A follow up study demonstrated that the same exercise protocol did not impair dynamic postural stability (Egerton, Brauer & Cresswell 2010). While these findings are useful, it is not possible to determine which specific modality of the exercise protocol impaired or did not impair postural sway.

Stemplewski et al. (2012) examined the effects of cycling at 60% of each individual's heart rate reserve for 10-min (118 beats·min^{-1}) on COP displacement during quiet bipedal standing in healthy older male adults (68.4 ± 2.9 years). The COP_V, and its components in the AP and ML directions significantly increased by ~34 %, ~ 35 % and ~ 30 %, respectively. A follow up study showed that the level of habitual physical activity may have an impact on the effects of exercise on postural sway (Stemplewski et al., 2013). This study comprised of two cohorts of either high level of physical activity or lower level physical activity. Habitual physical activity was assessed by an accelerometer over one week. The results indicated a significantly greater increase in the COP_V in those with a lower habitual physical activity (~ 57 %) compared to those who were more physically active (~ 22 %) when performed at the same relative intensity. These findings suggest that being more physically active is characterised by smaller increases in postural sway.

Donath et al. (2013) examined the effects of a treadmill $\dot{V}O_{2MAX}$ test and 2-km treadmill walk (76 ± 8 % $\dot{V}O_{2MAX}$) on single limb eyes open (EO) and double limb eyes closed (EC) postural sway in healthy older adults. It was shown that for both standing conditions COP_L increased more following $\dot{V}O_{2MAX}$ (~ 35 – 45 %) compared to a 2-km walking trial (~ 13 – 15 %). While difficult to rationally compare due to differences in exercise and posturographic protocols, comparative alterations in postural sway following exercise between young and older adults appear to be similar. However, the increases in COP_{ML} observed after exercise in older adults might suggest a temporary increase in fall risk immediately after exercise (Piirtola and Era 2006).

2.7.2 Disturbance to afferent and efferent information

Muscular fatigue, defined as an acute impairment in the ability to produce force, regardless of whether or not the task itself can be performed successfully (Enoka and Stuart 1992), can have a number of profound effects on the proprioceptive system responsible for postural stability (Forestier, Teasdale and Nougier 2002). Proprioceptive alterations due to exercise could be induced by changes in the discharge patterns of muscle afferents due to the build-up of metabolites (Paillard 2012). Such alterations may impair muscle spindle information (Hiemstra, Lo and Folwer 2001), disrupt central processing of proprioceptive information (Sharpe and Miles 1993) and efferent pathways thus decreasing muscular system efficiency and force generating capacity (Taylor et al. 2000). The accumulation of metabolic products as a function of exercise leads to less accurate commands of postural muscles and subsequent impairment in postural stability (Nardone et al., 1997). Muscular system efficiency relates to anticipatory postural adjustments (APA's) initiated by feed forward mechanisms (Schepens & Drew, 2004) and compensatory postural adjustments triggered by sensory feedback (Park et al., 2004).

Submaximal muscle contractions create modifications of muscle properties including the action potential, extracellular and intracellular ions (Allen et al. 2008) which decrease muscular excitability and increase the fluctuation of force production (Hunter, Duchateau and Enoka 2004). According to Surenkok et al. (2008) a decrease in pH, caused by lactic acid disassociating into lactate and hydrogen ions affects postural stability. However, these authors were unable to demonstrate the existence of any relationship between lactic acid accumulation and the magnitude of disturbance to postural sway. Exercise also elicits an increase in metabolic and energy demands which augments cardiac and respiratory contractions (i.e., heart beating and diaphragm contracting), thus exacerbating postural sway (Bove et al., 2007; Hodges et al., 2002; Paillard 2012). While Bousisset & Duchene (1994) suggested that sway disturbances from respiratory movements are weak, there is little doubt that these respiratory movements are counteracted by small angular displacements of the trunk and lower extremity (Hodges et al., 2002).

As previously discussed, running affects postural stability to a greater extent than cycling exercise. Visual, proprioceptive and vestibular sensory inputs are highly stimulated by running (Lepers et al. 1997). For example, otolithic receptors (which convey information with regards to the heads position relative to gravity), are sensitive to linear accelerations of up to at least 6 $cm \cdot s^{-2}$ (Fitzpatrick and McCloskey 1994). It is possible that excessive horizontal movements during running raises the detection threshold of otolithic organs, affecting integration of sensory information (Lepers et al. 1997; Paillard 2012). It was further shown that running affects postural sway to a greater extent than walking when matched for energy expenditure (Derave et al. 2002). Kinematic analysis revealed larger vertical displacements and acceleration of the head during running compared to walking (Derave et al. 2002). Stationary cycling does not elicit significant enough accelerations to alter vestibular afferent information (Paillard 2012).

2.7.3 Time course of effects

The duration of exercise induced fatigue on postural sway appears to be dependent on the type of exercise (e.g., intensity, duration and type of muscle contraction). Typically, the time course effects of cycling or treadmill exercise on postural sway in young healthy adults are short lasting, returning to baseline levels within 5 – 20 min (Nardone et al. 1997; 1998; Fox et al. 2008; Yaggie and Armstrong 2004). It is not yet clear how long older adults require for recovery. Bove et al. (2007) reported a linear relationship between $\dot{V}O_2$ and COP path length (R^2 = 0.82). The rapid recovery of $\dot{V}O_2$ might therefore explain the relatively short lasting effects of exercise on body sway. The authors did not allude to the potential mechanisms of this association.

Muscle histological factors between participants (i.e., percentage of fatigable type II fibres), differences in exercise protocols and type of posturographic assessment can at least partly explain the discrepancy in the duration of exercise effects on postural sway between studies. Despite the practical importance of understanding the open window of an exercise induced increased risk of falling, few studies consider the time course of effects. Addressing the recovery time frame after exercise will allow estimations to be made with respect to how long individuals should warm down before attempting to initiate a voluntary movement (e.g., stepping off an ergometer).

2.7.4 Gender effects

While early research suggested that the magnitude of fatigue effects on postural sway were not different between males and females (Nelson and Johnson 1973), more recent findings have suggested that postural sway is more adversely affected in males compared to females (Springer and Pincivero 2009; Wojcik et al. 2011). Previous findings have noted an apparent gender advantage in muscle fatigability, in favour of females, attributed to the lower absolute force produced performed at the same relative workload as males (Pincivero and Gandaio 2003). Females possess a greater capacity for utilising oxidative metabolism, therefore reducing the reliance of glycolytic pathways (Russ et al. 2005). According to Paillard (2012), these gender differences are only valid for local muscular fatigue. Indeed, men demonstrate greater increases in postural sway after local ankle fatigue (calf heel raises) compared to general exercise (rowing), while females experience greater postural sway following general exercise compared to local fatigue (Springer and Pincivero 2009). Therefore, gender differences appear to exist depending on the type of muscular exercise (e.g., local vs general exercise).

2.7.5 Upper body exercise

Exercise involving the proximal musculature (e.g., triceps surae) deteriorates postural control more than fatigue of the distal musculature (e.g., trunk and arms) (Bizid et al. 2009; Gribble and Hertel 2004). There are indications that afferent input from the upper extremity plays an important role in the control of upright stance as evidenced by an increase in postural sway following localised muscle fatigue of the trunk extensors (Vuillerme, Anziani, & Rougier, 2007), neck (Schieppati et al. 2003) and deltoids (Nussbaum 2003). However, the negative effects of non-postural muscle fatigue may be counteracted by muscle activation of postural muscles. Kanekar et al., (2008) examined the effects of fatiguing deltoid contractions on anticipatory postural adjustments and reported that activation of lower limb muscles (e.g., soleus, gastrocnemius, semitendinosus, biceps femoris and erector spinae) were able to compensate for the perturbations to maintain postural stability in bipedal stance. The earlier onset of anticipatory postural activity reported by Kanekar et al. (2008) may represent a functional adaptation by the central nervous system to preserve postural stability in the presence of fatigue.

The effects of general upper body exercise (e.g., involving several upper body muscles) on postural sway remains poorly understood. During ACE, isometric work of the abdominal and back extensors provides trunk stability to maintain balance and posture in the seated position (Sawka 1986). These isometric contractions of the trunk may become a limiting factor because the abdominal and intercostal muscles may compete for ventilation and balance control (Smith et al. 2010). Recently, Douris et al. (2011) noted that anaerobic arm cranking exercise perturbed single limb balance to a greater extent than anaerobic cycling in young adults. In contrast, the opposite findings were reported for maximal aerobic exercise. The scoring system adopted by Douris and colleagues (Dynamic Stability Index) is dissimilar to conventional COP measures of postural sway measured using a force platform. Therefore, it is difficult to rationally compare their findings with previous investigations. Furthermore, the exercise protocols were explicitly maximal in nature, therefore limiting the practical applications and generalizability of their findings, particularly for populations at risk of falling, such as the elderly. The authors specified that the trunk stabilisers were more fatigued during ACE compared to cycling. This is a reasonable explanation when considering that during single limb stance postural adjustments are highly dependent on movements at the hip and trunk (Tropp & Odenrick, 1988).

Surprisingly, Douris and colleagues did not allude to the potentially negative effects of upper body exercise effects of respiratory muscles. Arm crank ergometry decreases the ventilatory contribution of some of the inspiratory muscles of the rib cage as they have to contribute in non-ventilatory functions (e.g., upper torso and arm positioning), resulting in a greater contribution of the diaphragm and abdominal muscles (Celli et al. 1988). Since postural activity of the trunk muscles (e.g., diaphragm and abdominals) are altered when respiratory demand is increased (Hodges et al. 2002), there may be competition between the contribution of the trunk musculature to maintain balance and respiration, thus comprising the ability to minimise sway. Ultimately, increased activity of superficial abdominal muscles increases trunk stiffness and is likely to reduce the contribution of the trunk musculature movements for postural control (Smith et al., 2010). Douris et al. (2011) specified that since the entire upper body musculature was fatigued, there were no other upper body muscles which would have been able to compensate, as previously demonstrated by Kanekar et al. (2008). Douris et al. (2011) suggests that ACE may have been a novel mode of exercise which participants were not accustomed to. In addition, participants were not familiarised to exercise which is an important limitation with respects to postural sway responses to exercise.

Smith et al., (2010) investigated the effects of ACE on postural sway in older adults with chronic obstructive pulmonary disease (COPD) compared to a healthy older control group. Exercise intensity was based upon ratings of breathlessness using a modified Borg Scale (Borg 1982). Exercise was stopped when the COPD patients reported ‘very severe’ breathlessness and was then repeated to maintain ‘moderate’ (3 / 10) to ‘severe’ (7 / 10) breathlessness (duration not stated). Exercise dose was matched between pairs of participants with and without COPD who were matched for age and gender. While postural sway was affected in those with COPD, sway was minimally affected in the control group. The effects of exercise engaging the entire upper body musculature on postural sway remains poorly understood, especially in comparison to lower body exercise.

Conversely, there are several potential mechanisms as to why postural sway may not be affected following ACE. Several studies have reported that the central nervous system develops compensatory strategies after exercise. For example, following exercise anticipatory postural adjustments are reported where the activation of postural muscles occurs earlier

(Strang et al. 2009) and last longer (Strang and Berg 2007). Strang et al. (2008) noted that earlier onset of anticipatory postural adjustments gives muscles more time to achieve the force required to minimise postural sway. This results in a greater activation (EMG amplitude) of some postural muscles and weaker activation of others (Morris and Allison 2006). Therefore, it is possible that fatiguing the upper body musculature can be easily compensated for by the recruitment of new motor units, or the activation of previously inactive muscles (e.g., the lower limbs) (Strang et al. 2009). Since ACE does not fatigue muscles directly involved in postural stability (e.g., triceps surae), the CNS may compensate for fatigue of the upper body musculature by increasing the amplitude of non-fatigued muscles, such as the gastrocnemius and soleus.

2.8 Postural sway response to aerobic lower body exercise training

Postural sway increases with advancing age, owing to reduced strength of the ankle musculature and a decrease in tactile sensitivity, joint position sense and proprioception (Lord, Clark and Webster 1991). Proprioception has been reported to be the most important sensorial system for maintaining postural stability during normal fixed surface conditions (Horak 2006; Peterka 2002). It appears that being more physically active increases the use of these stimuli, thus allowing for more efficient postural adaptation (Perrin et al. 1999). Intuitively, some investigators advocate that balance abilities are specific to a particular task, and that these abilities can only be improved by task specific training (Nitz and Choy 2004). Research examining the efficacy of training interventions to reduce fall risk typically consist of lower body strength training (Bellew, Yates and Gater 2003; Bird et al., 2009), balance, co-ordination or tai-chi training (Wolf 1997; 2001; Zhang 2006), or a combination of the above (Lord et al., 1995; 2003; Nelson 2004; Ramsbottom 2004). In general, all of the abovementioned studies improve postural control.

It is widely recommended that older adults do endurance training activities to promote cardiovascular fitness. The American College of Sports Medicine published recommendations that every adult should accumulate 30 minutes or more of moderate-intensity physical activity on preferably all days of the week (Nelson et al. 2007). Although 30 min of balance training each day may be recommended to improve fall risk factors (Gschwind et al. 2013), it would be beneficial if 20 - 30 minutes of endurance training could simultaneously train cardiovascular fitness and balance (Donath et al. 2014). In this regard, multimodal training regimes including strength, endurance and balance tasks are considered important to reduce the risk of falls and cardiovascular and metabolic disease (Gardner et al. 2000). It is likely that for older adults, adherence to physical activity interventions would be better if balance training is integrated into endurance training (Buchner et al. 1997b).

While cycling requires less balance capacity than other exercises such as walking there is growing evidence that cycling training is associated with increased leg strength, muscle endurance and postural stability which are important risk factors for falls (Hassanlouei et al. 2014; Bouillon, Sklenka and Ver 2009; Rissel et al. 2013). From a strength and cardiorespiratory perspective, cycling training might provide an appropriate training stimulus and thus may be a feasible training modality for improving general fitness and balance among older adults.

A recent study by Rissel et al. (2013) found that cycling for one hour a week for 12-weeks was associated with significant improvements in timed single leg standing (145 ± 90 s to 175 ± 98 s). In contrast, Buchner et al. (1997b) reported that cycling training increased dynamic balance (distance walked on a narrow balance bean) by 3 % and therefore cycling exercise appeared to possess limited usefulness for balance improvements. Indeed, a further study by Buchner et al. (1997a) reported that 25-weeks of cycling for 1 hour, three days per week at an intensity corresponding to 75 % heart rate reserve had no effects on measures of gait and both static and dynamic balance. However, during an 18 month follow up individuals who participated in the cycle group reported to falling significantly less than the control group. Therefore, Buchner and colleagues specified that cycling improved fall risk factors by mechanisms other than improving aspects of gait and balance (i.e., improved confidence) or factors not ascertained by that particular study, such as behavioural changes (e.g., frequency of walking on ice).

Overall, cycling appears promising for improving fall risk factors, although this research domain remains poorly explored. Indeed, while several studies have reported the acute negative effects of cycling on postural sway (Maciaszek, Stemplewski & Osinski 2010; Stemplewski et al. 2012; 2013), no studies have explored whether these negative effects are mitigated by exercise training. Recent findings by Stemplewski et al. (2013) suggest that older adults with low levels of habitual physical activity were characterised by greater increases in the COP displacement after cycling exercise in comparison with older adults who were highly physically active. Furthermore, numerous review studies have reported an association between habitual physical activity history and postural sway (Gouchard et al. 2003; Kiers et al. 2013; Perrin et al. 1999). Collectively, these studies suggest that physically active older adults have better baseline postural control than their inactive peers and possess a greater ability to minimise sway after lower body exercise. Physically active individuals are able to regulate somatosensory inputs more efficiently than inactive counterparts, which leads to a smaller COP path length (conveying a small energy consumption) and small COP area (conveying precision) (Perrin et al. 1999).

2.9 Potential improvements in postural balance following ACE training

While cycling may have a more generic effect on postural control the use of the upper body musculature during ACE may elicit different balance adaptations. Rotational movements of the torso during ACE increase during high intensity exercise (Price et al., 2007) or when fatigued (Talbot 2012). The isometric work of the core is central to ACE as it is used for upper body stabilisation (Sawka 1986), allowing the arms to generate propulsive forces to the cranks

(Smith et al., 2008). While little to no research has examined the effects of ACE on trunk muscle activity, it is clear that there is a need for stabilisation of the trunk due to asymmetric force application in response to the rotating effects of the arms (Grigorenko et al. 2004). Rotational movements of the torso during ACE may improve the control of directionally sensitive postural muscles, such as the hip (adductor / abductor) and trunk musculature which during quiet bipedal standing are responsible for mediolateral balance adjustments (Winter et al., 1993, 1996). Increased trunk stability following ACE training might elicit an increase in neural drive to postural trunk musculature, which might in turn, induce the activation of atrophied muscles, which is particularly relevant to older untrained adults (Bjerkefors, Carpenter and Thorstensson 2007). In addition, Douris et al. (2011) reported that ACE significantly impaired single limb postural stability following maximal incremental exercise and Wingate trials. The authors discussed that the trunk stabilisers were fatigued following ACE and therefore the trunk musculature was less able to assist in maintaining balance. The reduced single limb balance control after acute ACE elucidate on the important role of the upper body in maintaining balance, which may have specific applications to muscles involved in mediolateral sway.

There is a lack of studies investigating the relationship between athletes who use their upper body (e.g., canoeists and kayakers) and postural stability. This is somewhat surprising when considering that the unstable water support during kayaking and/or canoeing pose a great postural challenge along the mediolateral plane in these athletes. Indeed Stambolieva et al. (2012) reported that when standing on fixed surface young athletes trained in kayaking and canoeing demonstrated greater sway amplitude and velocity than controls. However, when standing on a compliant surface athletes trained in the upper body demonstrated significantly less sway than controls. The authors suggested that the poorer postural performance observed on a fixed surface was a result of incorrect re-adaptation of sensory information returning to stable ground from water in kayakers and canoeists. Indeed, sickness of disembarkment occurs when rhythmic movements of the boat resists adaptations to stable conditions, persisting for up to 3 days (Cha 2009). However, Stambolieva et al. (2012) reported that trained upper body athletes coped more effectively with multisensory integration challenges (e.g., loss of proprioception and vision) compared to none athletes when standing on an unstable surface. It is of interest to determine the effects of seated upper body exercise training on a stable surface (e.g., arm crank ergometry) on postural sway to eliminate difficulties of adapting from unstable water to standing on a stable support surface.

2.10 Summary

It is apparent from the literature that there are several questions which require answers regarding the effects of acute and chronic upper body exercise on postural sway in both healthy young and older adults. Therefore, the aims and objectives of this thesis were to examine the effects on exercise postural sway in healthy young participants (Chapter 4), to follow this up in healthy older adults (Chapter 5) and to determine a range of tests which can be used in intervention studies in the older population pre and post training (Chapter 6). We also sought to establish the effects of upper and lower body exercise training on postural sway and other functional abilities in a group of otherwise healthy older adults (Chapter 7).

Chapter 3

General methods and Validation

This chapter will outline the methods which were undertaken within each study. In the case of different methods, specific designs and procedures are described in the relevant chapters.

3.1 Recruitment

All participants under the age of 60 years were recruited from the Coventry University student and staff populations. Recruitment was undertaken using e-mails sent through the University E-dition and poster notices located in common areas within the Department of Biomolecular and Sport Sciences. Adults over the age of 60 years were recruited from Coventry Council Leisure Centres or from collaborative links with Coventry Cathedral Choir. Interested participants voluntarily contacted the lead investigator for further information through face to face meetings, e-mail correspondence or telephone conversation.

3.2 Ethics Approval

All procedures of the studies were reviewed and ethical approval was granted from Coventry University Ethics Committee. The authors had no financial or personal relationships with anyone that could influence the outcome of the ethics committee.

3.3 Participant Information

Prior to participation in any of the studies, participants were provided with a detailed information sheet outlining the purpose and nature of the study. Where requested, a meeting was arranged to discuss the procedures of the particular study. Prior to any involvement all participants provided written informed consent to the experimental procedures as required by the Helsinki declaration (1964) and institutional ethics committee. When visiting the laboratory each participant completed a departmental health screening questionnaire to determine suitability for participation, their health on that particular day and, for the elderly participants or where appropriate, to obtain medical and exercise history.

3.4 Data Storage

All information collected during the course of the research was strictly confidential. Only the principal investigator was permitted access to any data with participant names on it (i.e., medical questionnaires). Any data stored electronically used unique codes to retain anonymity. Data collected was immediately transferred and subsequently deleted from University computers to a personal storage device. Personal computers were restricted to access of the principal investigator only by using a personal password. It was not possible to identify any of the participants from any published outputs. All data guidelines were followed and were in accordance with the Data Protection Act, 1998. On completion of each study, a summary of individual performance was sent to participants to those who requested it.

3.5 Exclusion Criteria

All participants were asked to continue with their normal daily diet and refrain from any exercise 24 hours prior to any testing sessions. The exclusion criteria for each study included the use of an assistive ambulation device, obesity (BMI >30), neurological disorders, musculoskeletal problems, balance disorders, uncorrected vision, inner ear disorder, diabete mellitus, peripheral vascular disease, cognitive impairment or dementia. Given that older people are exposed to more diseases and often take a range of medications, they are at a greater risk of the adverse reactions to pharmacological treatments (Macdonald 1985). The following drugs have been shown to have an association with falls risk and were therefore used as exclusion criteria in all adults over the age of 50; Psychoactive medication: hypnotics, anixolytics, antidepressants, antipyschotics; Cardiovascular medication: antihypertensive agents, diuretic, vasodilators and cardiotonics; Anti-inflammatories and analgesics: corticosteroids and nonsteroidal anti-inflammatories (Lord, Sherrington and Menz 2007).

3.6 Exercise Apparatus

3.6.1 Arm Crank Ergometer

Upper body exercise was conducted using either a mechanically braked modified cycle ergometer (Monark, 824E, Ergomedic, Sweden) (Study 1) or an electronically braked arm ergometer (Lode Angio BV, Gronngen, Netherlands) (Study 2 & 4). When using the modified cycle ergometer, the ergometer was clamped onto a sturdy table with foot pedals replaced with hand grips (Figure 3.1). Additional weights were added to the table to prevent movement of the ergometer at higher power outputs. All participants performed arm cranking exercise in an unrestrained seated position. The crankshaft of the ergometer was horizontally aligned with the centre of the glenohumeral joint (Smith & Price 2007). Chair height was adjusted using rubber matting. Each participant was required to sit at a distance from the ergometer so that when the back was vertical the arms were slightly bent when the crank arm was at the furthest horizontal point of the cycle. The participant was positioned so that their knees were flexed at approximately 90º. This was done in an attempt to reduce the lower body musculature from generating tension to assist turning the crankshaft. However, it is acknowledged that restricting the use of the legs during ACE is a difficult task since the legs are used to maintain balance during arm cranking (Sawka 1986). The Lode ergometer was connected to an external workload programmer to apply resistance proportional to the appropriate power output. The electronically braked ergometer was used with older adults as this allows lower initial power outputs and small incremental increases in power output suitable for this population to be performed.

Figure 3.1: The upper body exercise set up for the modified cycle ergometer (left) used in study 1 and electronically braked arm crank ergometer (right) used in studies 2 & 4

3.6.2 Cycle Ergometer

Lower body exercise tests undertaken in all studies were performed using a stationary mechanically braked cycle ergometer (Monark, 824E, Ergomedic, Sweden) (Figure 3.2). The ergometer was equipped with an adjustable handlebar and saddle. Seat height was adjusted so that the knee was slightly flexed at full extension of the crank (Myers et al. 2009). To increase intra-tester reliability the seat height was recorded and kept the same for all tests. The ergometer was also equipped with adjustable foot straps attached to each pedal, which were adjusted according to participant’s footwear. All participants were asked to wear the same footwear for all trials. During incremental exercise tests, all participants were instructed to maintain a seated position until the prescribed cadence could no longer be maintained. As recommended the cycle ergometer was calibrated on a monthly basis by the departmental technician to ensure power output was accurate (Myers et al. 2009). During calibration, the flywheel was cleaned using low grade sandpaper to dispose of any corrosion or deposits of dust or any other abrasive particles which can result in spring-like movements of the cradle. Digital displays were checked prior to commencement of exercise to ensure that correct unit of measurement was displayed (rpm) and the battery was functioning correctly for easy reading.

Figure 3.2: The lower body exercise set up on the cycle ergometer used in studies 1, 2 & 4

3.7 Exercise Protocols

3.7.1 Peak Oxygen Uptake

All studies utilised either a cycle ergometer (824E, Ergomedic, Sweden) or an electronically braked arm ergometer (Lode Angio BV, Gronngen, Netherlands). To determine percentages of maximal power output (W_{MAX}) in study 1, 2 (prediction) and 4, a $\dot{V}O_{2peak}$ test was performed by each participant. Tests were undertaken at the same time of the day, but on different days separated by at least 72 hours in a randomised order. Exercise trials consisted of an incremental exercise test. All exercise tests involved exercising until volitional exhaustion, which was defined as a reduction in the desired cadence by 5 rev·min^{-1} for 5 sec (Price et al., 2011).

3.7.1i Arm Crank Ergometer

Prior to all ACE protocols each individual completed a 5-min warm up using the unloaded ergometer or the 0 W setting. In study 1 (young participants) the ACE protocol involved an initial power output of 35 W, with increments of 20 W every 4 minutes for the first 4 stages, followed by 2 minute increments thereafter until volitional exhaustion. Four minute stages were adopted in order to elicit steady state exercise responses and to reduce the effects of premature fatigue and to aid the prediction of submaximal intensities. A cadence of 70 rev·min^{-1} was employed throughout trials (Smith and Price 2007). In study 2 & 4 (older participants) the ACE protocol started with an initial power output of 20 W, with increments of 5 W every 3 minutes for the first two stages, followed by increments of 5 W·min^{-1} until 85 % HR_{MAX} (Study 2) or volitional exhaustion (Study 4). A cadence of 60 rev·min^{-1} was employed throughout trials for older adults. The latter protocols have previously been successfully used to elicit peak responses in healthy older adults (Pogliaghi et al., 2006).

3.7.1ii Cycle Ergometer

Prior to all CE protocols each individual completed a 5-min warm up at 0 W (unloaded cradle setting). In study 1, the cycle ergometry (CE) protocol started at an initial power output of 70 W with increments of 35 W every 4 minutes for the first 4 stages, followed by 3 minute increments until volitional exhaustion. In study 2 & 4 participants exercised at an initial power output of 40 W with increments of 10 W every 3 minutes for the first two stages, followed by increments of 10 W·min^{-1} until volitional exhaustion. The crank rate was set at 60 rev·min^{-1} (Pogliaghi et al. 2006).

3.7.2 Criteria for volitional exhaustion

In young adults, peak oxygen uptake was successfully established when two of the following criteria had been reached, as outlined by The British Association of Sport and Exercise Science (BASES, 1997);

i. A respiratory exchange ratio (RER) of 1.15 or above
ii. Final heart rate within 10 beats·min^{-1} of age predicted maximum (CE; 220 – age, ACE; 200 - age)
iii. Post exercise ($\leq$ 5 min) blood lactate concentration of 8.0 mmol L^{-1} or greater
iv. A rating of perceived exertion (RPE) of 19 or 20 on the Borg Scale
v. Volitional exhaustion

To the authors knowledge specific requirements for attaining $\dot{V}O_{2peak}$ in the older adults have not been reported, but have been reviewed and are generally contrasting. For example, attainment of an RER of 1.0 (Patterson et al., 1999), 1.10 (Katzel, Sorkin & Fleg 2001) and 1.15 (Tanaka et al., 1997) have been used as a marker of exhaustion. Moreover, there is no agreement in the attainment of maximal heart rate, with 90 % HR_{MAX} (Borg 1982), 95 % HR_{MAX} (Katzel, Sorkin & Fleg 2001), or a HR within 5 beats·min^{-1} (Patterson et al., 1999) or 10 beats·min^{-1} (Cress et al., 1996) of predicted HR_{MAX} used as criteria. As a result, peak oxygen uptake was considered to be reached when two of the following criteria were met in the older cohorts;

i. A respiratory exchange ratio (RER) of 1.10 or above
ii. A final heart rate within 10 beats·min^{-1} of age predicted maximum using an age adjusted prediction equation (208 – 0.7 * age, Tanaka 2001)
iii. A rating of either local or central perceived exertion (RPE) of 18 or above on the Borg Scale

3.7.3 Submaximal Exercise Tests

For all submaximal exercise tests, with the exception of Study 2, exercise was performed at 50% of peak minute power (W_{PEAK}). Pilot work indicated that young untrained adults were not able to maintain 30-min of exercise at 60% W_{PEAK} or $\dot{V}O_{2PEAK}$. An example of the calculation for peak minute power is illustrated below (Smith et al. 2004).

Maximal exercise end time	=	18 min 30 sec		
Completed full 2-min of previous stage (i.e., 180 W) and 30 sec of next stage (i.e., 200 W)				
30 sec / 120 sec	=	(0.25 x 100) of next stage complete (200 W)		
25% of 20 W				
(Increase from 180 to 200)	=	5 W		
Peak minute power	=	180 + 5 W	=	185 W
50% W_{MAX}	=	185 / 2	=	92.5

3.7.4 Exercise Cadence

During arm crank ergometery, cadence has been shown to elicit significant effects on the attainment of peak oxygen uptake ($\dot{V}O_{2peak}$) with cadences of 70-80 rev·min^{-1} eliciting greater $\dot{V}O_{2peak}$ values compared to 60 rev·min^{-1} (Sawka et al. 1983; Price and Campbell 1997; Price et al. 2007; Smith et al. 2001). Therefore, a cadence of 70 rev·min^{-1} was employed for upper body exercise tests to determine $\dot{V}O_{2peak}$ in study 1. However, no cadence recommendations were found for older adult protocols. Lower cadences ranging from 50 – 60 rev·min^{-1} have been reported in older and clinical groups (Grange et al. 2004; Pogliaghi et al., 2006; Tew et al. 2009). Therefore, for upper body exercise testing in the older populations we utilised a cadence of 60 rev·min^{-1} which allowed direct comparisons of physiological responses with the only known study in healthy older adults (Pogliaghi et al., 2006). Cadence was matched for both ACE and CE protocols with respect to age.

3.7.5 Familiarisation

It is acknowledged that ACE is a unique and likely novel mode of exercise for most individuals (Smith et al., 2007). This is particularly likely in the older population and it was expected that the majority of participants would not be familiar with this mode of exercise. In study 1, 2 and 4, prior to the first visit participants were asked to visit the laboratory for an introductory meeting. During this time the principal investigator requested each participant to perform 10-min of ACE and 10-min of CE at 0 W (unloaded cradle setting) at a cadence consistent with the protocols for the particular study. All practice trials were performed in a counter balanced order. No physiological variables were monitored or recorded since the aim of this session was to acquaint participants with the unique movement of arm cranking.

3.8 Physiological Measurements

3.8.1 Online Breath-By-Breath Analysis System

Expired gas was collected using a breath-by-breath online gas analysis system (Metalyser 3B, Cortex Biophsik, Borsdorf, Germany) and subsequently analysed using MetaSoft v.3.9.7 software (Cortex, Lepzig, Germany). Before each test, the analyser was calibrated for barometric pressure, volume and gas concentrations. Barometric pressure was calibrated against pressure determined using a mercury barometer (F Darton & Co. Ltd, UK). Calibration of the gases was determined by sampling known concentrations of oxygen (15%) and carbon dioxide (5%) using calibration gas obtained from portable gas canisters (BOC Ltd, UK) as well as ambient air. The volume transducer was calibrated with a 3-litre capacity syringe (Hans Rudolph, USA). All calibration procedures for expired gas analysis were followed in accordance with the manufactures instructions. To minimise error, the system was recalibrated prior to each exercise trial. All trials were relatively short (20 – 30 min) therefore little drift was expected or occurred between trials.

During each trial, a rubber mask was secured over the mouth and nose of the participants ensuring that expired and inspired gas passed through the flow sensor and sample line. Different sized face masks were used according to the size of the participants face. The gas sample line transported expired gas to the oxygen and carbon dioxide cells for the determination of the appropriate gas concentration. The MetaMax® 3B Base Unit was connected to a PC running MetaSoft® Studio Software to continually record and monitor all expired gas. Prior to all exercise trials, baseline measurements were recorded for at least 5 mins. Data were then imported to Microsoft Excel®. Data were averaged in the final 20 s of each particular time period for values of oxygen consumption ($\dot{V}O_2$), carbon dioxide ($\dot{V}CO_2$), respiratory exchange ratio (RER) and minute ventilation ($\dot{V}_E$).

3.8.2 Douglas bag technique

Due to malfunction of the breath-by-breath gas analyser system the MetaMax was unavailable for nine final submaximal exercise trials in study 1. As a result expired gas was collected in 200L Douglas Bags (Harvard Apparatus, UK). All valves, tubing and Douglas bags were routinely checked for leaks. All participants (n = 4) were naive to the Douglas bag technique. Therefore, participants were habituated to breathing through the mouth piece prior to data collection. As this technique was used only for steady state protocols, participants were required to exercise with the mouth piece for 30s prior to gas collection, which allowed sufficient time to clear dead space in the tubing (Eston & Reilly 2009). Prior to measurement of expired gas, O_2 and CO_2 analysers were calibrated and dry gas meters checked. The dry gas meter (Seromex, model 1400, Crowthrone, UK) was calibrated with 100% nitrogen to set the zero for both O_2 and CO_2 analysers and concentrations of oxygen and carbon dioxide which span the working range. Douglas bags were analysed for fractions of oxygen (O_2) and carbon dioxide (CO_2) using a dry gas analyser (Seromex, model 1400, Crowthrone, UK). The total volume (L) and temperature (ºC) were also measured using a gas meter and thermometer. Values for oxygen uptake ($\dot{V}O2$), minute ventilation ($\dot{V}_E$) and respiratory exchange ratio (RER) were analysed using an internal gas analysis programme.

3.8.3 Body Mass and Height

Body mass was recorded prior to each experimental trial using both electronic (SECA, Model 877, gmbh & Co., Germany) and mechanical column (SECA, Model 710, gmbh & Co., Germany) weighing scales. For experimental chapters involving exercise trials (study 1, 2 & 4) body mass was also recorded following exercise to ensure there was no significant weight loss due to sweating.

3.8.4 Heart Rate

Heart rate (beats·min^{-1}) was continuously monitored at rest, during exercise and recovery using a Polar heart rate monitor (Polar Electro, Oy, Finland). The wrist unit (Polar FS2C) displayed heart rate by telemetry received from the polar transmitter which was secured around the participant's chest using an adjustable elasticised strap. Heart rate was continually monitored and averaged over a 30s period. For study 1 the standard equation of 220 – age was utilised to predict HR_{MAX} for CE (Fox and Haskell 1971). For all theoretical predictions of HR_{MAX} in older adults (study 2 & 4) an age adjusted regression equation was used (208 – 0.7 * age) (Tanaka et al. 2001). This particular equation does not underestimate HR_{MAX} in older adults, unlike some of its counterparts (i.e., 220 – age) (Tanaka et al. 2001). Moreover, Tanaka et al., (2001) reported that HR_{MAX} is predicted primarily by age alone, and does not require adjustment between genders.

The above regression equation proposed by Tanaka et al. (2001) is based upon lower body exercise. To the authors knowledge no equations have been developed for use with upper body exercise, which naturally requires adjustment. Maximal HR achieved during ACE has been reported to be ~10 – 20 beats·min^{-1} lower than those reported during CE in healthy older males (67 ± 5 years) (Pogliaghi et al. 2006).

3.8.4 i Maximal heart rate

Maximal heart rate (HR_{MAX}) decreases substantially with age (Tanaka et al. 2001). Therefore, age adjusted regression equations have been developed (Fox and Haskell 1971; Nes et al. 2012; Tanaka et al. 2001). However, validity of such equations are based on exercise with the lower body musculature, and have not been established for upper body exercise testing. Maximal HR in study 1 and 4 was measured during maximal incremental exercise tests using a cycle ergometer (CE) and arm crank ergometer (ACE). Age-based HR_{MAX} was predicted using the Fox (1971) (220 – age) and Tanaka (2001) (208 – 0.7 * age) equations. One-way ANOVA was used to determine differences between measured HR_{MAX} and age prediction equations. Measured HR_{MAX} for ACE was significantly lower compared to the Fox (P = 0.009) and Tanaka (P = 0.001) equations for both age cohorts (Table 3.1). These preliminary findings show that age adjusted equations based on lower body exercise over estimate measured HR_{MAX} for ACE. Subtracting 20 beats·min^{-1} from the Tanaka equation provides a reasonable estimate of HR_{MAX} for ACE compared to a graded exercise test.

3.8.5 Perceived Exertion

Ratings of perceived exertion (RPE) were measured using the 6-20 point Borg Scale (Borg 1982). A rating of perceived exertion was recorded for both local (RPE_L) and central (RPE_C) perceptions of exertion. The central RPE pertained to subjective feelings of cardiorespiratory stress, while local RPE were associated with sensations from the working muscles. To ensure all participants understood the RPE scale, the verbal anchors associated with each number were explained fully (Borg 1998).

Table 3.1: Measured maximal heart rate (beats·min^{-1}) compared to commonly used age adjusted prediction equations for upper and lower body exercise in young and older adults

Cohort	Mode	Measured HR_{MAX}	Tanaka (2001) (208 -0.7 * age)	Difference (Tanaka)	Fox (1971) (220 – age)	Difference (Fox)
Study 1	ACE	173 ± 9*†	191 ± 3	18 ± 8	195 ± 5	22 ± 9
(N = 10)	CE	188 ± 4	191 ± 3	3 ± 5	195 ± 5	7 ± 7
Study 4	ACE	144 ± 16*†	162 ± 4	18 ± 7	154 ± 6	10 ± 4
(N = 20)	CE	159 ± 15	162 ± 4	3 ± 6	154 ± 6	- 5 ± 6

* Note that N for each study is different to those reported in Chapters 4 (N = 9) and 7 (N = 18). The present cohorts are based on samples prior to drop outs. * Singificant between measured and Tanaka. † Significant between measured and Fox.

3.9 Postural Sway Measurements

3.9.1 Piezoelectric Force Platform

Posturographic ground reaction forces in study 1 were tested by means of a stationary force platform (Kistler Force Platform 9281B, Kistler Instruments, Switzerland). The force platform was embedded and set in concrete and was fixed into a walkway foundation according to the manufacturer's specifications. The dimension of the rectangular force platform was 600 x 400 x 60mm and weighed 410 N. Signals from ground reaction forces (Fx, Fy, Fz) acting on each of the four transducers were amplified and converted from analogue-to-digital (AD) signal using a 16 bit AD converter. All data was collected at a sampling frequency of 1000 Hz and recorded using Vicon Workstation® (Vicon Peak Workstation®, Oxford Metrics, UK) and subsequently calculated using a custom built virtual instrument (LabView 6.0).

The Kistler 9281B force platform contained 4 pizometeres in each load cell, one located in each corner of the force platform. Pizometeres located within each load cell generate an electric charge when subjected to mechanical load. The force measured as an electrical signal is transmitted through type 9807 amplifiers. It is acknowledged that postural sway rarely exceeds 1 – 2 Hz. Based on Nyquist's theorem, a sampling rate of 10 Hz would likely be sufficient for analysis of sway. However, it is generally considered that this is too low a frequency, and most authors use sampling rates at least 5 – 10 times the highest sway frequency (i.e., 25 Hz – 200 Hz) (Mello, Oliveria and Nadal 2010; Gouchard et al. 2002; Vuillerme & Hintzy 2007). During study one, raw EMG signals were simultaneously recorded with COP data and therefore higher sampling rates of at least 1000 Hz were required to synchronise data with EMG.

3.9.2 Hall Effect Force Platform

For studies two, three and four posturographic ground reaction forces were examined by means of a portable force plate (AMTI, AccuGait, Watertown, MA) and sampled at 100 Hz. All forces were filtered with a 4th order low-pass Butterworth filter with a cut off frequency of 6 Hz. Data were recorded using the accompanying real time data acquisition software package (AMTI, Netforce®, Watertown, MA) and subsequently converted into COP data using the accompanying balance analysis software package (AMTI, BioAnalysis, Version 2.2, Watertown, MA). Analysis of COP data lead to the computation of main sway parameters; COP path length and COP displacement in the anteroposterior and mediolateral directions. These parameters of postural sway were used for comparative proposes with previous investigations (e.g., Stemplewski et al. 2013). The validity and reliability of these parameters computed with a force platform have been accepted (Pinsault and Vuillerme 2009). The AMTI force platform differs with the Kistler force platform in the context of the measurement of forces only. In the portable force platform, Hall Effect sensors and magnets are arranged within each force transducer located in each corner of the platform. Each sensor measures magnetic field changes which occur when integrated spring elements deflect due to external forces acting upon the surface of the platform (AMTI, User Manual). This particular system has a built in amplifier, therefore an AD converter was not required. The force platform was connected to an external interface box using Ethernet connection. The interface box was then connected to a PC using an RS-232 serial port allowing signals to be processed.

3.9.3 Postural Sway Measurement

Participants were instructed to stand barefoot on a single force platform in an upright and bipedal position (Figure 3.3). To minimise movements of the upper body participants were instructed to let their arms hang by their sides (Doyle et al. 2007). When standing quietly with eyes open participants were instructed to look straight ahead at a target which was adjusted to the eye level of each participant, thus preventing vestibular disturbance. The target was a black circle of 150 mm diameter mounted on either a board or wall 2 m from the centre of the force platform. All participants were asked to distribute their body mass symmetrically on both feet, avoiding any extraneous movements. Each participant was requested to 'stand as still as possible' during each trial (Zok, Mazza and Cappozzo 2008). At rest, each participant performed trials alternatively with eyes open (EO) and eyes closed (EC) for a total of 3 trials for each condition. The measurement time was 36 s with a resting period of 30 s between each trial where participants were asked to remain on the force platform (Pinsault and Vuillerme 2009). For all exercise trials, postural sway measurements following exercise were recorded under EO conditions at 1, 3, 5, 10, 15 and 30 minutes, alternated with EC measurements at minutes 2, 4, 6, 11, 16 and 31 minutes (Studies 1, 2 & 4). The first and final 3 s of each trial were removed to avoid potential movements upon commencing and ending the trial (Le Clair & Riach 1996). Therefore, the total analysed sampling time was 30 s.

Figure 3.3: Standard upright bipedal stance position on a single force platform

3.9.3i Foot Position

Standardised foot positions have been proposed in the literature for postural sway measurement (e.g., feet abducted by 30º, heels separated by 3 cm) (Pinsault & Vuillerme 2009). In the present thesis foot positioning was standardised for all participants during each quiet standing test using individual foot templates (Buckley et al. 2005). A partially standardised foot position, with the heels separated by 3 cm from the medial extremity of the posterior side of the calcaneus was adopted. The angle of the feet was left to participant's discretion. Participants were thus asked to stand on top of the force platform with feet separated by 3 cm and a preferred angle base of support, a common procedure in the literature

(Stel et al. 2003). A sheet of A3 plain white paper was placed over the surface of the force platform and the principal investigator traced around the feet so that position was consistent (Du Pasquier et al., 2003).

3.9.3 i Pilot Study: Foot Positioning

There is little standardisation in the adopted foot position for the assessment of postural sway in the literature, which makes responses difficult to interpret and compare between studies. Many traditional centre of pressure (COP) variables have been shown to have a moderate to large variability, both within and between-subjects (Chiari, Rocchi and Cappello 2002), possibly owing in part to foot position. Constraining the feet outside of the individual's comfortable position could potentially affect the recording of postural responses (McIlroy and Maki 1997). In line with this, many studies have allowed participants to adopt a stance according to their own comfort (Chiari, Rocchi and Cappello 2002; Doyle *et al.* 2007). In study 1, participants stood on a template which separated the heels by 3cm, as measured from the medial extremity of the calcaneus, with the foot angle left to participant's discretion. Although, most previous studies did not standardised foot position (Egerton, Brauer & Cresswell 2009; Gouchard et al., 2002; Mello, de Oliveira & Nadal 2010; Stemplewski et al., 2012), it was recognised that there was a need to compare the position adopted in study 1 with those previously and commonly adopted. Existing recommendations for foot position include; (a) feet abducted at 30º, heels separated by 3 cm (Pinsault and Vuillerme 2009), (b) feet abducted at 30º, heels touching (International Society of Posturography) and (c) Romberg stance where heels and toes are placed together (Black et al. 1982). Specifically, this pilot study sought to explore whether absolute postural sway values and standard error (coefficient of variation) differed between a previously standardised foot placement protocol (Pinsault and Vuillerme 2009) and a partially standardised foot position protocol as used in study 1.

Methods

Thirty male (n = 16) and female (n = 14) participants between the ages of 18 - 80 years, volunteered to take part in the data collection procedures. Participant demographics are presented in Table 3.2. The age ranges used in this case study have previously been used in studies examining age related changes in postural control (Liaw *et al.* 2008).

Table 3.2: Demographics of participants

Cohort (Years)	N	Gender (M/F)	Age (years)	Height (cm)	Mass(Kg)
18-39	10	5/5	27.3 ± 5.4	173.6 ± 12.1	77.0 ± 15.3
40-59	10	5/5	44.7 ± 3.7	171.8 ± 7.4	76.9 ± 10.7
60-80	10	6/4	63.1 ± 2.9	163.8 ± 6.0	73.9 ± 13.1

The standardised testing procedure was designed to mimic the methodological approaches utilised in previous reliability studies (Bauer et al. 2008; Lafond et al. 2004; Pinsault and Vuillerme 2009). Participants were instructed to stand with their heels at a distance of 3 cm apart, as measured from the medial extremity of the calcaneus, with a foot angle of 30º. A standard template was designed and secured to the portable force platform. For the self-selected foot position, participants were requested to stand in a comfortable position, with the angle of the feet left to participant's discretion but with the heels separated by 3 cm. All participants were asked to remain barefoot. A tracing of the participant's feet was made on A3 paper after they had chosen their preferred foot position (Figure 3.4). The angle of the feet was calculated from the medial border of the calcaneus and the distal end of the great toe. The great toe was used to define foot angle, since the great toe is more pronounced and could be identified more reliably from the foot tracings. A two-way analysis of variance (ANOVA) with repeated measures on both factors (age [18 – 39 years, 40 – 59 years and 60 – 80 years) × position [self-selected and standardised) was conducted to examine differences in postural sway between stance positions. In order to measure the relative variability of centre of pressure variables, the coefficient of variation (CV) was calculated (SD / Mean * 100).

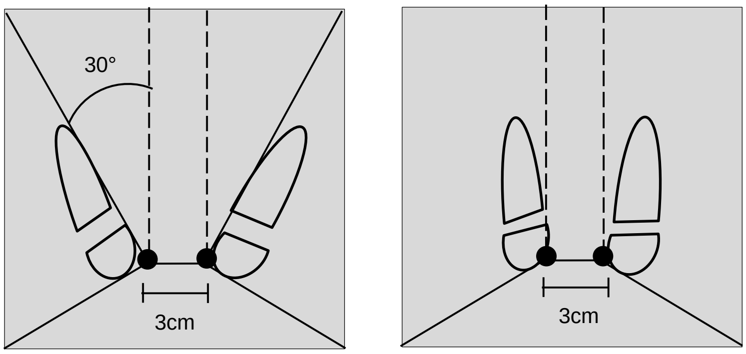


Figure 3.4: Foot tracing of a participant during a standardised foot position (A) and a self-selected foot angle (B)

Results

The overall mean ± standard deviation for the self-selected foot angle was 17.6 ± 3.7 º for the whole group. The range between the smallest and greatest foot angle was rather small (range 15 – 22 º). No differences in foot angle were observed between age groups ($P \geq 0.05$) (Figure 3.5). The standardised position as employed by Pinsault and Vuillerme (2009), the standard Romberg test (width 0 cm, angle 0°) and the International Society of Posturography (ISP) (width 0 cm, 30°) all fall outside the range of present data. The current data shows that participants chose a foot angle between the two extremes (0 - 30 º). These data show that recommended protocols constrain the feet outside of more comfortable foot angles, however the adopted foot position was similar between all participants. No significant interactions of age x stance were observed for COP_L ($P = 0.821$), COP_{AP} ($P = 0.591$), COP_{ML} ($P = 0.234$) or COP area ($P = 0.338$). However, main effects of age and stance were observed for COP_L (age; $P = 0.001$, stance; $P = 0.041$), COP_{AP} (age; $P = 0.032$, stance; $P = 0.007$), COP_{ML} (age; $P = 0.001$, stance; $P = 0.002$) or COP area (age; $P = 0.003$, stance; $P = 0.001$) (Table 3.3).

Absolute sway values were generally greater during the standardised position, in addition to the CV. Therefore, this data suggests that the standardised foot position elicits greater absolute COP values and a greater variance than a partially standardised foot position.

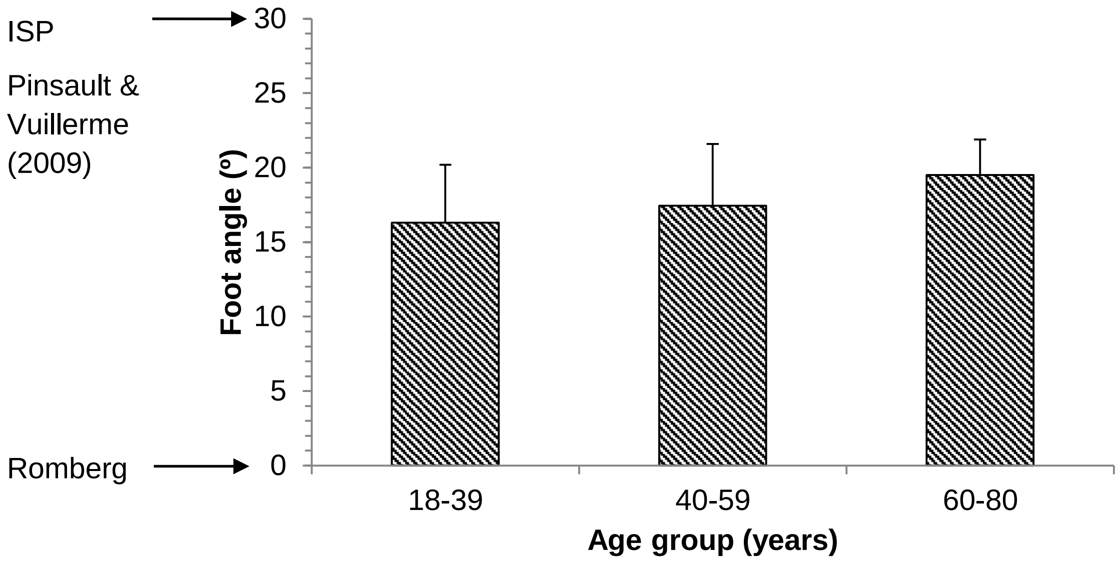


Figure 3.5: Self-selected foot angle for each of the age groups.

Discussion

Absolute postural sway and coefficient of variation (CV) were greater for all COP measures when standing in a standardised foot position (heels 3 cm apart, with a foot angle of 30º). As a result, to standardise the foot position all future posturographic tests required participants to stand with their feet separated by 3 cm and a self-selected angle, as the differences were quite small between groups and therefore it is not expected that foot position will differ in future studies. To the authors knowledge's, most studies in this research domain do not standardise foot position (Egerton, Brauer & Cresswell 2009; Gouchard et al., 2002; Mello, de Oliveira & Nadal 2010; Stemplewski et al., 2012).

3.9.3 ii Familiarisation to Postural Sway Measurement

To familiarise participants with the standing tasks for postural sway measurement and to validate their understanding of the instructions specified by the principal investigator, all participants were required to perform two eyes open and two eyes closed practice trials in an attempt to eliminate a learning effect (Nordahl et al. 2000). Reliability data presented in section 3.9.5 demonstrates that any practice effects were minimal, if not evident at all. To avoid extraneous postural movements in stabilising the body after stepping onto the force platform, trials were commenced after an initial 10s period, as recommended by Benvenuti et al. (1999).

3.9.4 Postural sway during standing conditions

3.9.4 i Compliant Surface

Participants who volunteered for studies 3 & 4 were asked to stand barefoot on a commercially available manufactured foam block (Balance-pad Plus, Alcan Airex AG, Switzerland) (apparent density, 55 kg/m^3; compression resistance, 18 kPa and 70 kPa for 25% and 50% compression respectively; tensile strength, 240 kPa). The dimension of the foam was 50 x 41 x 6 cm. This particular product is widely used in the literature (Tanaka and Uetake 2005). The foam was placed on top of the force platform and a length of non-slip rubber was positioned under the foam to prevent the foam from moving. When standing on foam, the principal investigator stood next to the participant to provide support due to an increase risk of falling.

Table 3.3: Mean ± SD and CV for COP parameters of postural sway in the standardised (A) and partially standardised (B) foot position in young, middle aged and older adults

Variable	18 – 39 Years		40- 59 Years		60 – 80 Years	
	A	B	A	B	A	B
COP Area 95% (cm^2)	1.57 ± 0.59	0.69 ± 0.30	2.04 ± 0.79	0.81 ± 0.53	2.92 ± 1.57	1.24 ± 0.49
CV %	37.8	30.3	38.8	35.3	43.9	37.9
P value	0.001		0.001		0.002	
COP_L (cm)	57.73 ± 7.83	54.87 ± 8.48	61.74 ± 5.70	56.46 ± 5.96	68.33 ± 6.67	65.64 ± 5.58
CV %	13.6	10.5	12.2	9.6	11.8	8.5
P value	0.002		0.001		0.06	
COP_{AP} (cm)	1.67 ± 0.59	1.31 ± 0.27	2.38 ± 1.40	1.46 ± 0.50	2.59 ± 1.04	1.91 ± 0.88
CV %	25.1	20.7	28.7	24.0	30.1	20.9
P value	0.021		0.035		0.02	
COP_{ML} (cm)	1.27 ± 0.36	0.68 ± 0.26	1.44 ± 0.44	0.92 ± 0.46	2.34 ± 0.55	1.33 ± 0.58
CV %	28.2	18.9	30.5	22.0	23.5	20.6
P value	0.001		0.001		0.007	

3.9.4ii Single Limb Stance

In studies 3 and 4 participants were asked to maintain a static standing position with the metatarsophalangeal joints of the load foot positioned in the centre of the force platform. It was instructed that the unloaded leg should not touch the supporting leg and the knee should be flexed to 90º. A total of three trials were recorded on each foot. Participants were asked to stand in this position for 30 s or as long as possible. Each trial was separated by a self-determined recovery period. Termination of the test was recorded if 1) the foot touched the support leg; 2) hopping occurred; 3) the foot touched the floor, 4) the arms touched something for support or 5) 30 s measurement time was reached.

3.9.4iii Load Carriage

In all trials in study 3 & 4 participants were required to hold a standardised load in their right and left hand, or a load equally distributed in both hands. The load used consisted of metal weights held in a reusable shopping bags made from woven synthetic fibres (dimension 34 cm x 38 cm x 16 cm). The mass of the loads were confirmed prior to each condition. Each participant held a bag weight which corresponded to a percentage of individual's body mass for each condition. Each condition was repeated three times, yielding a total of 9 trials.

3.9.5 Centre of pressure normal variation pilot study

Reliability represents an essential prerequisite for any outcome measure of postural sway to ensure that any observed differences in COP measures reflect a true change in balance control, rather than systematic or random error (Dijkers et al., 2002). Although the reliability of COP measures have previously been addressed in the literature (Pinsault and Vuillerme 2009), there was a need to investigate the within and between session variation in COP based measures of postural sway. This data will allow identification of the magnitude of differences between trials and days due to normal biological variation and subsequently allow us to determine the absolute effects of exercise on postural sway in the present thesis.

Methods

Eleven healthy male participants (mean ± SD age, 29.1±5.9; height, 1.81±0.07 m; mass, 79.1±10.3 kg) volunteered to visit the gait analysis laboratory on three consecutive days (Tue, Wed & Thur in the same week).

Each individual completed three postural sway measurements with eyes open and three measurements with eyes closed. All measurements were carried out at the same time of day and by the same principal investigator. The experimental procedures are outlined in section 3.9.3.

The reliability of COP measures were determined using intra-class correlation coefficient (ICC) and typical percent 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. The ICC has previously been used to report relative reliability of COP measures (Pinsault and Vuillerme 2009). Mean ± SD for the within and between session variation are also reported.

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.

No differences in sway measures were observed when trials were averaged over three separate days ($P \geq 0.05$) (Figure 3.6). The standard error, range and magnitude of effects are reported in the following sections. As with trials between days, no differences in sway measures were observed when trials were analysed over three consecutive trials (Figure 3.7).

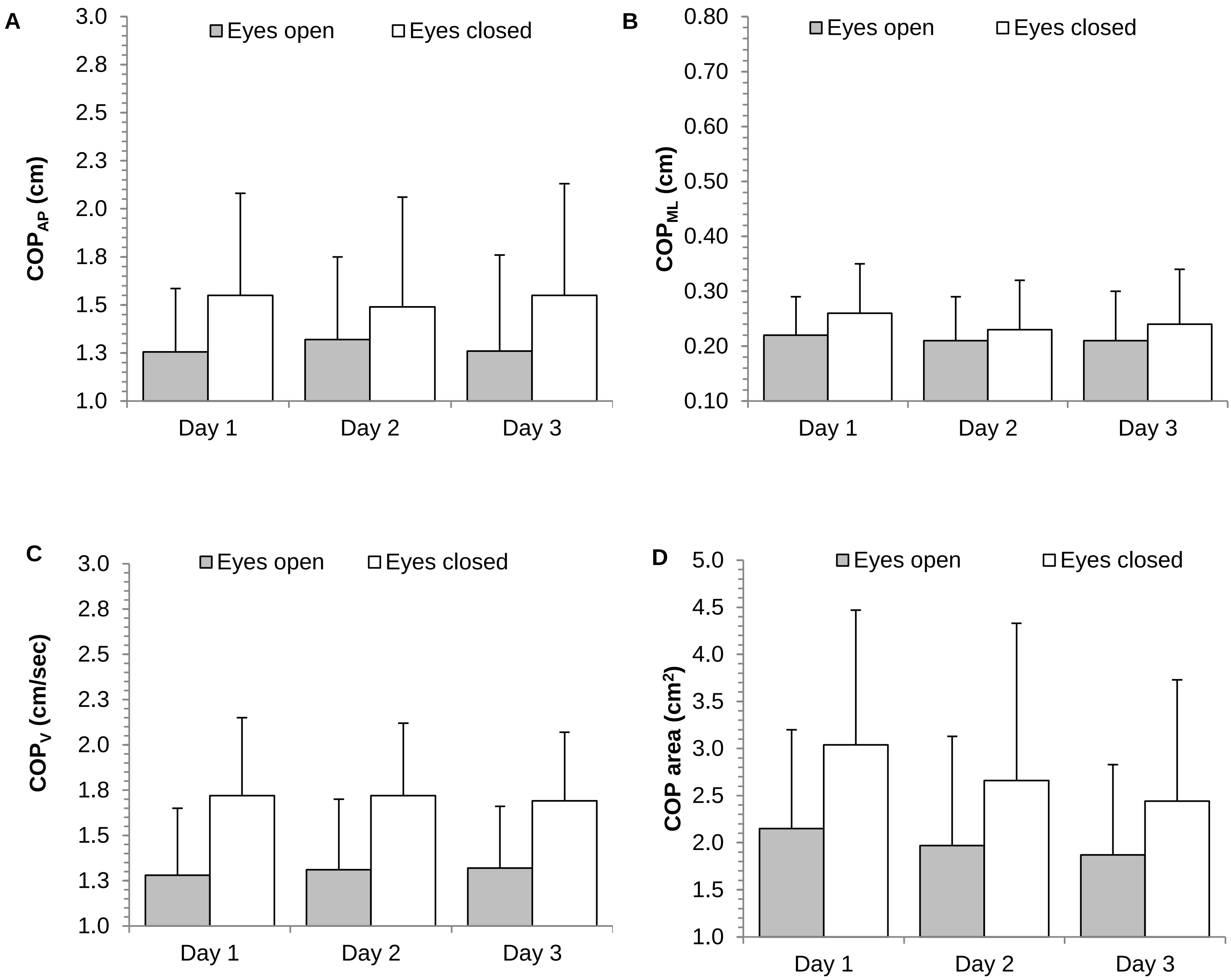


Figure 3.6: Between session variations of four COP measures of postural sway

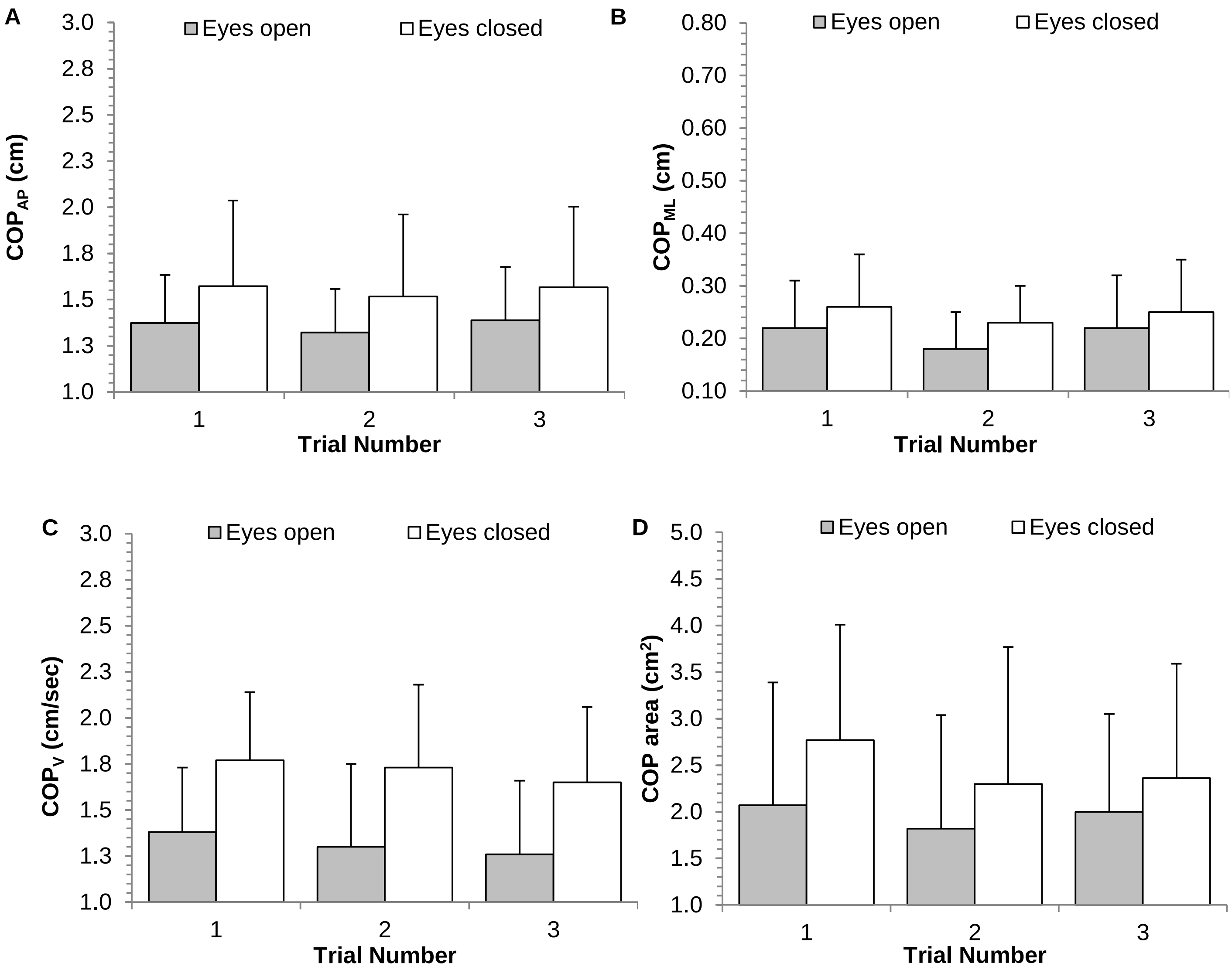


Figure 3.7: Within session variation of four COP measures of postural sway

Reliability of COP measures of postural sway (within session)

Anteroposterior and mediolateral COP displacement, average velocity (path length x time) of the COP and sway area with a 95% ellipse all showed large to nearly perfect ICC's. The coefficient of variation (CV) and intra-class correlations coefficient (ICC) are presented in the following tables for each variable.

COP_{AP}

The present data demonstrates that COP_{AP} is reliable, with ICC values ranging from 0.72 – 0.91 (Table 3.4). The CV ranged from 17 – 23 % with both eyes open and eyes closed.

Table 3.4: Intra-class correlation coefficient, coefficient of variation and range for COP_{AP} across three days

	Day	Range (cm)	CV (%)	ICC	ICC Description
EO	1	0.18	23	0.833	Very large
	2	0.16	19	0.849	Very large
	3	0.15	22	0.797	Very large
EC	1	0.23	20	0.846	Very large
	2	0.19	17	0.914	Near perfect
	3	0.22	21	0.725	Very large

COP_{ML}

Mediolateral COP displacement was generally consistent with ICC values ranging from 0.70 – 0.90 (Table 3.6) which were very similar to those observed in the anteroposterior direction and considered to be large.

Table 3.6: Intra-class correlation coefficient, coefficient of variation and range for COP_{ML} across three days

	Day	Range (cm)	CV (%)	ICC	ICC Description
EO	1	0.01	22	0.907	Near perfect
	2	0.04	23	0.882	Very large
	3	0.05	18	0.896	Very large
EC	1	0.04	23	0.886	Very large
	2	0.04	18	0.704	Very large
	3	0.11	19	0.856	Very large

COP_{V}

The COP_{V} was the most consistently realiable COP measure of postural sway recorded in this study (Table 3.7). ICC values ranged from 0.81 – 0.95, most of which were considered as near perfect correlations. This variable also showed the consistently smallest CV, ranging between 12 – 15 %.

Table 3.7: Intra-class correlation coefficient, coefficient of variation and range for COP_V across three days

	Day	Range (cm)	CV (%)	ICC	ICC Description
EO	1	0.17	16	.817	Very large
	2	0.08	15	.950	Near perfect
	3	0.10	16	.822	Very large
EC	1	0.09	13	.934	Near perfect
	2	0.06	12	.914	Near perfect
	3	0.08	15	.901	Near perfect

COP 95% elliptical area

The COP area with a 95% ellipse was the second most consistent measure of postural sway. Two of the six outcomes showing near perfect correlations, with the remaining four showing very large correlations ranging from 0.83 – 0.96 (Table 3.8). However, this variable showed the consistently greatest CV, ranging between 44 – 55 %. Values ranged from 0.65 – 3.55 cm^2.

Table 3.8: Intra-class correlation coefficient, coefficient of variation and range for COP area with a 95% ellipse applied to data across three days

	Day	Range (cm^2)	CV (%)	ICC	ICC Description
EO	1	1.10	44	.864	Very large
	2	0.65	47	.865	Very large
	3	0.74	54	.832	Very large
EC	1	0.96	55	.961	Near perfect
	2	1.55	46	.830	Very large
	3	0.98	52	.945	Near perfect

Discussion

As part of this reliability study we measured anteroposterior and mediolateral COP displacement (cm), average velocity of the COP ($cm \cdot sec^{-1}$) and the COP area with an ellipse applied to 95% of the COP data points (cm^2). In general, all variables had very large or near perfect ICC's. This data is consistent with the findings of Pinsault & Vuillerme (2009) which showed that three, 30-s postural sway measurements were sufficient to ensure excellent reliability of standard COP measures widely used in research, namely 2-dimensional COP variables (area, COP range, mean velocity). Similar findings were expected, as this study adopted the same instructions to participants (Zok et al., 2008), sampling duration and number of trials in each visual conditions. The average velocity of the COP was the most reliable postural sway variable showing very large or near perfect correlations, regardless of visual condition (0.81 – 0.95). This variable also demonstrated the consistently smallest CV across the three days. Moreover, COP displacements in the mediolateral and anteroposterior direction also showed very large correlations and CV values which are consistent with the literature. However, while the COP area also demonstrated very large ICC and upper limit CI of 0.9 or above, this variable also exhibited the greatest CV values, SD's and large ranges both within and between sessions. Hufschmidt et al., (1990) observed an intra-subject variation of COP area to have a high CV of 58.9 % following 10 trials. It is generally accepted that good reliability can be accepted with CV values of ~ 20 % (Hopkins et al. 2009). In light of this, with the exception of COP area, the remaining variables show good reliability and therefore would be appropriate to use in future studies.

3.10 Motion Analysis

Motion analysis was recorded during study 1 only. The Vicon motion analysis system (Vicon 512, Oxford Metrics, UK) was used to measure movements of separate body segments along the anteroposterior and mediolateral axis during upright bipedal stance. This system was integrated with the Kistler force plate. Six Vicon motion capture video cameras operating at a sampling frequency of 50 Hz set on tripods (Manfrotto, Italy) were positioned around the participants. Video cameras were connected to the workstation using ethernet connection and then connected to a personal computer. The size of the capture volume was reduced to the smallest area possible since it affects the system resolution and therefore precision (Milner 2008). Motion cameras were positioned so that camera one, two and three captured the participant’s right view and cameras four, five and six captured the left view. Each camera was ~2m away from the recorded movements in order to minimise the amount of dead space surrounding the cameras field of view within the capture volume.

3.10.1 Motion Analysis Calibration

Prior to testing the system was calibrated to allow the software to calculate the relative location and orientation of cameras and to define the 3-dimensional coordinate system. Initially, four reflective markers were placed around the force plate to ensure there was no background interference (i.e., sun light, reflective material). Static calibration used a rigid L-frame with four markers mounted in known locations. The static L-frame was used to define the location of the origin and orientation of the reference frame. Dynamic calibration used a calibration wand (2x 50 mm reflective spheres, 500 mm apart). The technique involved working around the outside of the area waving the wand in a figure of eight motion, gradually working in towards the centre of the force plate. Calibration was only accepted if all cameras gave residual error readings of less than 1.0 mm and less than 2% for static reproducibility, which was in accordance with manufacturer guidelines. A residual error of 1.0 mm ensures that a markers position in space will be located within 1 mm of its true position (Milner 2008). Camera sensitivity was adjusted to minimise background interference however, all cameras were required to have a sensitivity level of 9 out of 10 (unitless value).

3.10.2 Marker Sets

Eight retroreflective caption markers (8 mm diameter) (Vicon, Oxford Metrics, UK) were placed unilaterally on the lateral malleolus (ankle), lateral femoral epicondyle (knee), mid femur (mid-thigh), sacrum (base of spine), anterior superior iliac spine crest (pelvis), olecranon (elbow) and acromion process (shoulder). Anthropometric locations of the markers were made according to bony landmarks determined by palpation according to the procedures proposed by Cappozzo et al. (1995) and manufacturer's guidelines. The distance between each marker and the markers position relative to the bony landmark was measured and recorded for subsequent trials. In keeping with recommendations (Milner 2008) each marker was viewed by at least two cameras during data recording to facilitate three dimensional reconstructions. Moreover, the distance between markers was sufficient so the system could differentiate between markers during reconstruction. Reflective markers captured were subsequently identified and labelled. While it is acknowledged that minimising skin marker artefact must be the main concern in marker set design (Cappozzo et al. 1995), during quiet standing skin marker movement error is minimal.

3.10.3 Data Processing

Data were recorded using Vicon Workstation® (Vicon Peak Workstation®, Oxford Metrics, UK) at a sampling rate of 200 Hz. Data were filtered in Vicon Workstation® using a low pass Butterworth filter at a cut of frequency of 6 Hz. Following filtering, data were exported into an Excel file and the maximal amplitudes (range) of segments were calculated as the difference between minimum and maximum trajectories during the measurement period in all cardinal axes.

3.10.4 Motion Analysis System Error

System error was analysed by measuring the 'noise' of a static marker placed in the middle of the capture volume. A single marker was placed on the middle of the force plate in the same position as where participants stood while video cameras measured relative 'noise'. The measurement was repeated three times. The magnitude of marker displacement in the X and Y axis was calculated as the difference between minimum and maximum trajectories. One way ANOVA revealed no differences over trials in the X or Y axis ($P \geq 0.05$). Marker noise in the Y-axis ranged from 0.85 – 0.87 mm, and 0.84 – 0.87 mm in the X-axis.

3.11 Electromyography (EMG)

All surface electromyographic (sEMG) signals during studies 1 and 4 were registered from disposable stud electrodes (40.8 x 35 mm diameter) (Ambu, Blue Sensor M, Denmark) and recorded using a 16-channel portable biomonitor (Mega Electronics Ltd, ME6000, Finland). Surface EMG (sEMG) was chosen to study superficial muscle activity during quiet standing in preference to more invasive methods, since sEMG provides a global measure of muscle activity (Soderberg & Knutson 2000).

3.11 i Site preparation and electrode placement

As recommended by Hermens et al. (2000) body hair was removed from all EMG sites using disposable razors. Additionally, each site was cleaned using isopropyl alcohol wipes to remove dirt and oil to reduce skin impedance and external noise. Participants were assured that this process would not cause discomfort. Surface electrodes were taped to the skin surface, with all accompanying wires also taped down which reduces both movement artefact and the likelihood of electrodes detaching during exercise. However, since EMG was only recorded before and after exercise, any electrodes which detached were replaced at the earliest time post exercise. Surface electrodes were placed bilaterally over the belly of the muscles (Table 3.9) with an inter-electrode spacing (centre to centre) of 3 cm, which was in accordance with the manufacture recommendations. The reference electrode was place 6 -7 cm laterally from the recording electrodes. The location of the electrodes on the participants muscle belly was marked on the participant's skin using a surgical marker pen to enable the electrodes to be replaced at the exact same site for subsequent tests.

In healthy older adults, the seven muscles of interest were the tibilais anterior (TA), gastrocnemius medialis (GM), biceps femoris (BF), rectus femoris (vastus lateralis) (RF), rectus abdominus (RA) and erector spinae (ES) muscles on the right side of the body. The right side was chosen because all 18 participants in study 4 reported the right foot as the dominant side. Electrode placement and orientation were standardised and adhered to the European project "Surface EMG for non-invasive assessment of muscles" (SENIAM) procedures reported in previous literature (Table 3.9).

3.11 ii EMG processing

Data were sampled at 1000 Hz and smoothed using a low pass Butterworth filter with a cut off frequency of 6 Hz (Figure 3.13). The highest frequency of sEMG signals ranged between 400 – 500 Hz (Besajian and De Lica 1985). Raw EMG signals were converted from analogue to digital using a 14-bit band pass filter. All signals were acquired in real time and analysed using MegaWin (v. 3.0.1) software. The average root mean square for each muscle was calculated over a 10 s period from the middle of a 30 s trial. A 10 s measurement time has previously been used (Panzer, Bandinelli & Hallett 1989). The average muscle activity was calculated during this 10 s period.

3.11 iii Maximal Voluntary Isometric Contraction

Normalisation of the EMG signal using maximal voluntary isometric contraction (MVIC) is widely used with EMG studies and has successfully been used in both younger (Ntousis et al., 2012) and older adults (Nagai et al., 2011; 2013) when examining standing balance. To enable normalisation of the EMG amplitudes during quiet standing tests, MVIC's were carried out before balance trials. The principal investigator demonstrated all isometric contractions and a demonstration card was presented to all participants for review. Strong verbal encouragement was given during each contraction to promote maximum effort. The EMG data from the MVIC's were used to normalise the EMG amplitude (% MVIC) during postural sway trials. Following data smoothing the peak force was calculated from the best of three muscle contractions.

Table 3.9: Brief description of selected muscles, rationale and function of each muscle, and recommendations

Muscle	Rationale / Function	Electrode placement (SENIAM)	MVIC
Tibilais anterior	Dorsi flexor which counteracts posterior body sway (Latash 2008). Three times greater activity in old compared to young adults (Loughton et al. 2003)	The electrode was placed 25% of the distance between the tip of the fibula and tip of the medial malleous, just lateral to the tibia	Participants were asked to stand up and perform a unilateral dorsi flexion at a 90º ankle position. The feet were manually restrained.
Gastrocnemius medialis	Plantar flexor to counteract anterior sway (Loram et al., 2005).	Electrode was placed on the most prominent bulge of the medial muscle belly. The electrode was oriented in the direction of the leg.	Participants were positioned in a seat which was secured to the ground and asked to perform a unilateral plantar flexion at a 90º ankle position.
Biceps femoris	Counteracts posterior body sway, but only when feet are close together (Joseph and Nightingale 1954).	The electrode was placed at 50% on the line between the ischial tuberosity and lateral epicondyle of the tibia.	Participants lay in a prone position. The hip was manually restrained to the ground. Participants were asked to perform a unilateral knee flexion at 20 – 30º knee flexion.
Vastus lateralis	Older adults show significantly more activity compared to young adults (Loughton et al. 2003)	The electrode was placed 2/3 on the line from the anterior spina iliac superior to the lateral side of the patella.	Participants were seated on a chair (height 46 cm) and asked to perform single leg knee extension which was restrained between 70 - 90º knee flexion position.
Rectus abdominis	Control gross trunk movement and provide trunk stability for postural control (O'Sullivan et al., 2002).	The electrode was placed ~ 2cm lateral and across from the umbilicus and oriented parallel with muscle fibres.	Participant lay in supine position with feet secured to the ground. Participants were requested to flex the spine by ~30º. Torso manually restrained
Erector spinae (Longissimus)	Active during standing, maintaining thoracic and lumbar spine in a neutral kyphosis against gravity (Andersson et al., 1974)	The electrode was place at 2 fingers width lateral to L1. Electrodes oriented vertically.	Participants lay in a prone position and were asked to raise the upper torso from the ground. Arms were restrained.
Trapezius	An increase in sway has been observed following cervical muscle fatigue, which may be experienced following upper limb exercise.	The electode was place 50% on the line from the acromion to the spine on vertebra C7.	Shoulders were manually restrained and participants were instructed to shrug the shoulders.

Chapter 4

The Effects of Upper and Lower Body Exercise on Postural Sway in Healthy Young Adults

4.1 Introduction

The ability to maintain and control bipedal stance is an essential prerequisite for many physical and daily activities, such as gait initiation and reaching tasks (Lepers et al., 1997). Maintaining postural control during quiet standing has long been known to be a complex process of positional adjustments of the muscles acting over joints of the lower extremity and is controlled by the integration of afferent information from visual, vestibular and proprioceptive information within the central nervous system (Enoka, 2008). There are also indications that afferent input from the upper extremity plays an important role in the control of upright stance as evidenced by an increase in postural sway following localised muscle fatigue of the neck (Schieppati, Mardone & Schmid, 2003), deltoids (Nussbaum, 2003) and the trunk extensors (Vuillerme, Anziani & Rougier, 2007).

Postural control may not only be important for daily activities, but also sporting activities (Alderton, Moritz, & Moe-Nilssen, 2003). An inevitable consequence of physical activity is muscle fatigue, defined as a decrease in the muscles force generating capacity (Kent-Braun 1999). Muscle fatigue disrupts both the peripheral proprioceptive system and central processing of proprioception (Sharpe and Miles 1993). Several studies have reported that exhaustive exercise, such as a maximal oxygen uptake test on a cycle ergometer (CE) impairs the ability to minimise postural sway during quiet bipedal stance (Gouchard et al., 2002; Mello, de Oliveira, & Nadal, 2010). However, at submaximal intensities the effects of exercise are less clear. For example, no changes in post exercise postural sway were observed when CE corresponds to 60% of maximal heart rate for 25-min (Nardone et al., 1997) or 60-min at 70%

ventilatory threshold (Mello, de Oliveira, & Nadal, 2010). However, disturbances are observed when CE was performed at 50 % of maximal aerobic power for 30 min (Demura & Uchiyama 2009) or at 200 W for 15 min (Vuillerme & Hintzy 2007). Fatiguing exercise, which utilises the lower body musculature, likely deteriorates the quality of sensory proprioceptive information and/or integration (Paillard, 2012) and also decreases muscular system efficiency (i.e., increases work rate) (Nardone et al., 1997), thus disturbing postural sway post exercise.

Furthermore, while postural sway responses to treadmill and CE exercise have been examined, little information exists for arm exercise. Arm crank ergometry (ACE) training elicits improved walking distance in patients with reduced lower body exercise capacity (Tew et al., 2009; Zwierska et al., 2007) and as such this mode of exercise may have other important applications. For example, ACE and CE have been shown to elicit a similar improvement in both specific (trained muscles) and cross transfer (untrained muscles) effects following training in both young (Loftin et al., 1988) and older adults (Pogliaghi et al., 2006). On a practical standpoint, this work has demonstrated that ACE could potentially be an effective alternative form of exercise for healthy adults. Upper body exercise may subsequently provide an effective training stimulus without fatiguing the lower limbs and increasing the risk of falls immediately following exercise which would have applications for elderly and clinical groups. However, research concerning the effects of ACE on postural sway has provided less clear findings than those reported for CE. For example, Douris et al. (2011) observed a greater increase in single limb postural sway following maximal aerobic ACE compared to CE, while the opposite findings were observed following short duration high intensity anaerobic exercise. Both exercise protocols were explicitly maximal in nature. To the author's knowledge, the immediate effects of a submaximal upper body exercise protocol on postural sway are yet to be established and have not been compared with lower body exercise. Therefore, comparing the effects of submaximal upper and lower body exercise on postural sway would allow applications to be made to other clinical and older cohorts which are comparable to the level of exertion experienced during training or daily life in these populations.

In addition, previous investigations of postural sway responses to CE have adopted a bipedal stance (Demura & Uchiyama, 2009; Gouchard et al., 2002; Mello, de Oliveira, & Nadal, 2010; Vuillerme, & Hintzy, 2007). It is well known that during quiet bipedal stance, sway is primarily controlled by ankle plantar and dorsi flexors (Winter 1995). However, during single limb stance postural adjustments are made at the hip (Tropp & Odenrick 1988). Therefore, it is reasonable to assume that ACE might not have the same effects on postural sway when standing in a bipedal compared to a single limb stance. This is supported by findings in healthy older adults which showed that ACE does not disturb postural sway when standing in a bipedal position (Smith et al., 2010).

The aim of the present study was carried out to more thoroughly investigate the effects of ACE on postural sway by determining whether upper limb exercise perturbed postural sway to the same extent as lower limb exercise at both maximal and submaximal exercise intensities. This work will build on prior studies (Douris et al. 2011; Smith et al., 2010) in that it will allow more comprehensive comparisons of upper body exercise to be made with previous literature, in the context of exercise and posturographic protocols. This remains a novel area which will allow applications to be made to populations at an increased risk of falling, such as the elderly.

Research Hypothesis (H_1): Upper body exercise will disturb postural sway but the effects will be less than for lower body exercise which will elicit an increase in postural sway post exercise in healthy younger adults.

Null Hypothesis ($H0_1$): No significant changes in postural sway following upper or lower body exercise in young healthy adults.

4.2 Methods

4.2.1 Participants

Nine healthy male participants (mean ± SD age, 24.1±4.8y; height, 1.77±0.05 m; mass, 75.6±13.9 kg) volunteered to take part in the study, which had received ethical approval by Coventry University Ethics Committee. All participants reported being physically active at least 3 hours each week at moderate to vigorous intensities (50 - 85% VO_{2PEAK}) in a range of sports (e.g., cycling, football, rugby and/or running), as recommended by ACSM's guidelines (Franklin, Whaley, & Howley, 2000). None were specifically trained in either upper or lower body exercise. None of the participants reported cardiovascular or pulmonary diseases, neurological and vestibular disorders, orthopaedic pathology or musculoskeletal problems. All participants provided written informed consent.

4.2.2 Exercise Trials

4.2.2 i Peak oxygen uptake

To determine each individual's ergometer specific peak oxygen uptake (VO_{2PEAK}) and peak power (W_{MAX}) participants performed incremental exercise tests on both an arm crank ergometer (ACE) and a cycle ergometer (CE). Maximal tests were also used to determine the effects of exhaustive exercise on postural sway. Tests were undertaken at the same time of the day, but on different days, separated by at least 72 hours in a counterbalanced order.

Throughout each protocol participants were verbally encouraged to exercise for as long as possible until volitional exhaustion or until the prescribed cadence of 70 rev·min^{-1} could not be maintained (>5 rev·min^{-1}) for longer than 10 s. Expired gas was collected by a breath-by-breath analyser and was averaged over the final 20 s of each incremental exercise stage and the 20 s prior to volitional exhaustion. Heart rate was recorded in the final 10s of each increment and at volitional exhaustion. A rating of both RPE_L and RPE_C was obtained at the same time as HR and immediately upon reaching volitional exhaustion. Blood lactate samples were obtained at rest, volitional exhaustion and after 5 minutes of passive recovery.

4.2.2 ii Experimental trials

At least 72 hours after the $\dot{V}O_{2PEAK}$ trials participants visited the laboratory on three further occasions to perform a 30 min ergometer specific submaximal exercise test. Two sub-maximal trials involved participants working at 50% of ergometer specific W_{MAX} (ACE_{REL}; 53 ± 8 W and CE_{REL}; 109 ± 16 W, respectively). Due to lower W_{MAX} achieved during the upper body maximal exercise trial, a further experimental trial was performed on the CE at the same absolute power output (CE_{ABS}) as the ACE_{REL} trial (i.e., 53 ± 8 W). Prior to all trials, participants were required to perform a 3 minute warm up on the unloaded ergometer at a cadence of 70 rev·min^{-1}. Expired gas was continually recorded and averaged in the final 20 s of minutes 5, 15 and 30. Heart rate was also continually monitored and recorded at the same times as expired gas samples. Ratings for both RPE_L and RPE_C were obtained at 5, 15 and 30 minutes. Blood lactate concentration was determined from a capillary blood sample at rest and 5, 15 and 30 minutes of exercise.

4.2.3 Physiological Measurements

During the preliminary and experimental trials expired gas was continuously analysed using a breath-by-breath gas analyser system (MetaMax, Cortex Biophsik, Borsdorf, Germany) for oxygen uptake ($\dot{V}O_2$), minute ventilation ($\dot{V}_E$) and respiratory exchange ratio (RER). Heart rate was continuously monitored using a Polar heart rate monitor (Polar Electro, Oy, Finland). A rating of both local (RPE_L; working muscles) and central (RPE_C; cardiorespiratory stress) perceived exertion (RPE) using the 6-20 point Borg scale (Borg 1973) was used to record subjective feelings of effort during exercise. To ensure participants understood the scale the verbal anchors associated with each number was explained (Borg 1998).

4.2.4 Blood samples

Ear lobe arterialised capillary blood samples were collected for determination of blood lactate concentration (mmol·L^{-1}). Blood was collected and mixed in 20 µL capillary tubes and analysed for blood lactate using an automatic lactate analyser (Biosen C_Line, EKF Diagnostic, Germany).

4.2.5 Postural sway measures

4.2.5 i Force platform

Participants were instructed to minimise postural sway by standing as still as possible for 30s on force platform mounted in the ground (Kistler Force Plate 9281B, Kistler Instruments, Switzerland). Centre of pressure path length (COP_L) and displacements in the anteroposterior (COP_{AP}) and mediolateral (COP_{ML}) directions were used to quantify postural sway in pre-test and post-test conditions. For specific postural sway assessment procedures refer to section 3.9.3.

4.2.5 ii Motion analysis

The Vicon motion analysis system (Vicon 512, Oxford Metrics, UK) was used to measure joint centre movements of separate body segments along the *X* and *Y* axis during quite stance. Eight retroreflective markers (19mm) were placed bilaterally on the lateral malleoli (ankle), lateral femoral tuberosity (knee), mid femur, anterior suprailiac spines (pelvis), sacrum, ulnar styloid (wrist) olecranon (elbow) and acrominon process (shoulder) for analysis of the magnitude of marker displacements (Figure 4.2).

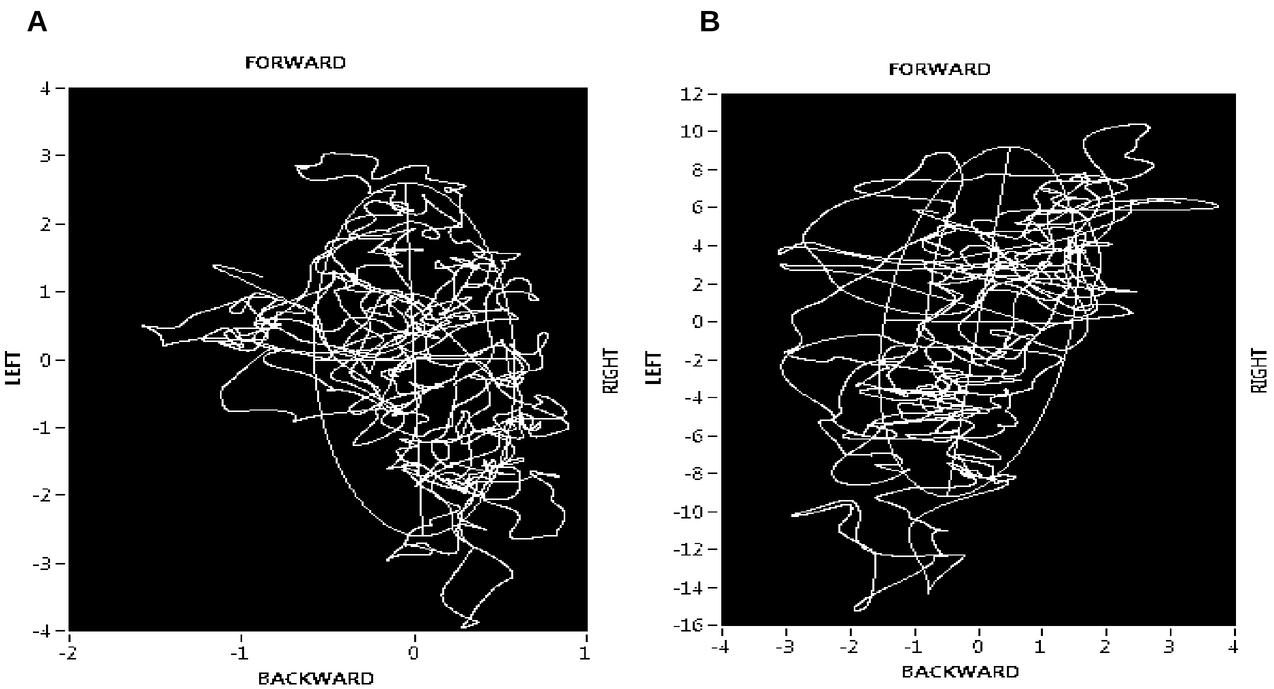


Figure 4.1: Examples of the trajectories of the COP, of a representative participant before (A) and immediately after (B) fatiguing cycle ergometer exercise. Note differences in the scale.

Anatomical position measurements were made according to bony landmarks determined by palpation according to the procedures proposed by Cappozzo *et al.* (1995). The distance between each marker and the markers position relative to the bony landmark was measured and recorded for subsequent trials. Reflective markers captured were subsequently identified and labelled. Data were then exported as an ASCII file from the Vicon software to Excel® where kinematic data for each segment for minimum and maximum coordinates of the joint centres were calculated. The first and last 300 co-ordinates were excluded to avoid analysing any unnecessary movements from the participants which may have occurred at the start and end of the trials (Section 3.10).

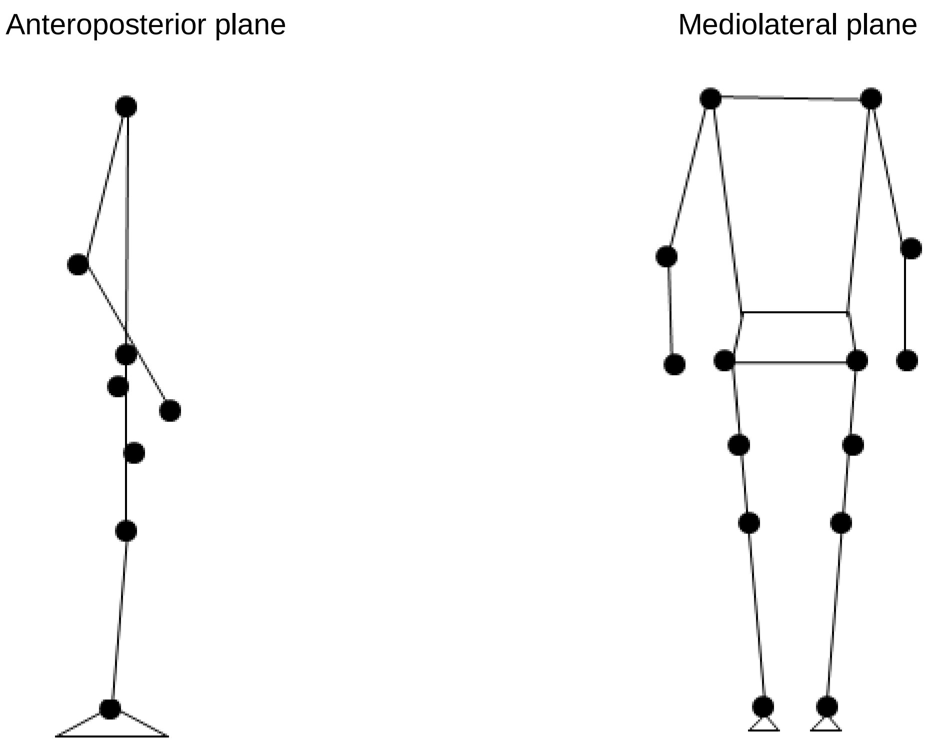


Figure 4.2: Kinematic representation for the analysis of joint movements

4.2.5 iii Electromyography case study

Since EMG data was obtained for only one participant in study 1 (three experimental trials), data are presented within a case study. All data presented are for ACE_{REL}, CE_{REL} and CE_{ABS}. EMG data were obtained for the following muscles; gastrocnemius lateralis, tibialis anterior, biceps femoris, rectus femoris, erector spinae and trapezius. For lower extremity sites, right and left limb EMG data were obtained, while just the right side was collected for upper extremity sites (Section 3.11).

4.2.6 Statistical Analysis

Data were tested for normality (Shapiro Wilk Test) and homogeneity of variance (Levenes Test). A two-way analysis of variance (ANOVA) with repeated measures on both factors (time $\times$ mode) was conducted to examine changes induced by exercise on outcome measures obtained from the COP data (e.g., modes: ACE_{REL}, CE_{REL} & CE_{ABS}; time: pre (0), 1, 3, 5, 10, 15 and 30 min) (Figure 4.3). The visual conditions of EO and EC were analysed separately. Paired t-tests were carried out to examine differences in peak physiological values for the incremental exercise tests. Data was analysed using PASW version 17.0 (SPSS Inc., Chicago, IL). Two-way ANOVA was also conducted to examine differences in physiological responses between upper and lower body exercise (ACE_{REL}, CE_{REL} & CE_{ABS}). Statistical significance was set at $P \leq 0.05$ level. Where the result of the ANOVA was statistically significant, Tukey's Honestly Significant Difference (HSD) post hoc analysis was conducted.

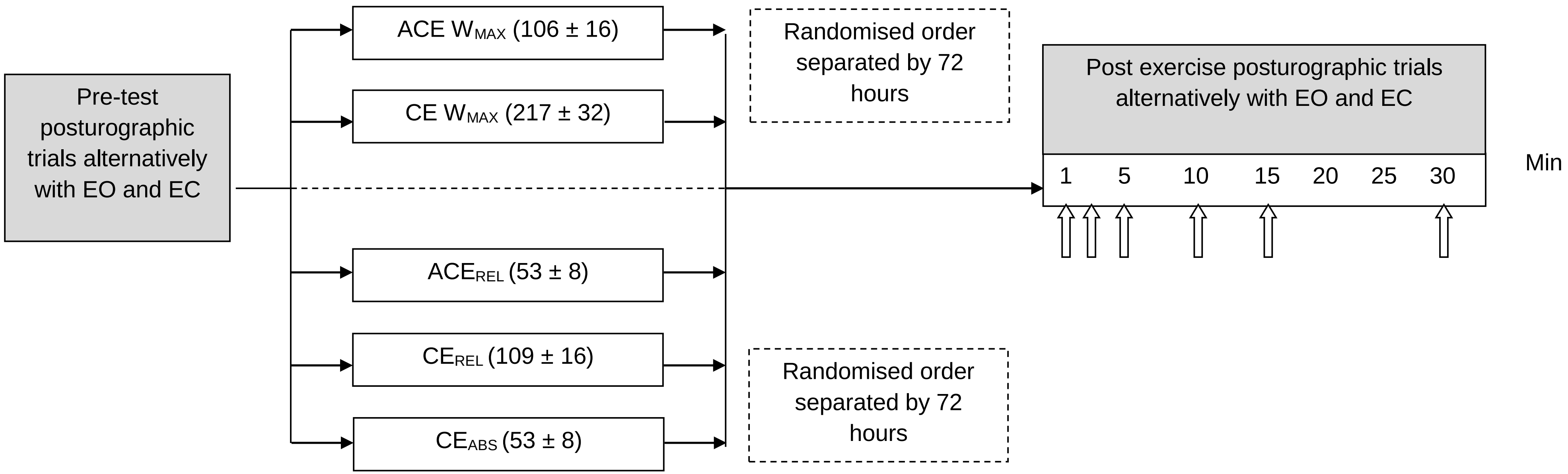


Figure 4.3: Experimental setup indicating posturographic trials before and after maximal and submaximal ACE and CE. Posturographic trials were alternated between eyes open (EO) and eyes closed (EC). Maximal exercise tests were performed first in a counterbalanced order. Submaximal exercise trials were performed after maximal exercise tests in a randomised order.

4.3 Results

4.3.1 Peak physiological responses

Peak physiological responses to maximal ACE and CE are presented in Table 4.1. Significant differences were observed between modes for $\dot{V}O_{2PEAK}$ (P = 0.009), $\dot{V}_{EPEAK}$ (P= 0.032), W_{MAX} (P = 0.001), HR_{MAX} (P = 0.031), and RPE_C (P = 0.001). Data obtained for all variables were significantly greater for CE compared to ACE, with the exception of peak RPE_L and peak blood lactate where no differences were observed (P = 0.598).

Table 4.1: Peak physiological responses during maximal arm crank ergometry (ACE) exercise and cycle ergometry (CE) exercise protocols (N=9)

	ACE	**CE**	**% of CE**
$\dot{V}O_{2PEAK}$ (L·min^{-1})	2.62 ± 0.3*	3.23 ± 0.5	82.2 ± 11.2
$\dot{V}O_{2PEAK}$ (ml/min/kg)	35 ± 7*	45 ± 8	77 ± 9
$\dot{V}_{EPEAK}$ (L·min^{-1})	77.60 ± 13.3*	95.02 ± 17.9	82.5 ± 11.4
Peak Power Output (W_{MAX})	106 ± 16*	217 ± 32	49.2 ± 8.0
Time to exhaustion (min)	16.29 ± 3.18	18.54 ± 3.17	88.9 ± 14.2
HR_{MAX} (beats·min^{-1})	173 ± 9*	182 ± 8	94.9 ± 5.5
Respiratory exchange ratio	0.90 ± 0.12*	0.96 ± 0.11	93.64 ± 4.9
Blood lactate peak (mmol·L^{-1})	9.4 ± 1.4	9.3 ± 1.7	102.8 ± 19.7
Blood lactate 5 min post exercise (mmol·L^{-1})	9.2 ± 1.3	9.2 ± 1.9	102.9 ± 22.7
RPE_L	20 ± 0.0	20 ± 0.0	100.0 ± 0.0
RPE_C	17 ± 1.0*	20 ± 1.0	85.9 ± 7.4

*Significantly different from CE (P <0.05)

4.3.2 Submaximal physiological responses

Physiological responses to sub-maximal exercise protocols are presented in Table 4.2. Significant interactions were observed between all three trials (ACE_{REL}, CE_{REL} and CE_{ABS}) for

$\dot{V}_E$ (P = 0.043), HR ($P \leq$ 0.01), and RPE_C (P = 0.048). At 30 min of exercise $\dot{V}O_2$ (P = 0.011), $\dot{V}_E$ (P = 0.013), HR (P = 0.001) and RPE_C (P = 0.03) were significantly greater during CE_{REL} compared to ACE_{REL}. For CE_{REL}, $\dot{V}O_2$ (P = 0.019), $\dot{V}_E$ (p =0.001), HR (P = 0.012), BLa (P = 0.001) and both RPE_L (P = 0.001) and RPE_C (P = 0.008) were significantly greater than those reported for CE_{ABS}. Similarly, $\dot{V}O_2$ (P = 0.017), HR (p =0.018), BLa (P = 0.001), $\dot{V}_E$ (P = 0.001), RPE_C (P = 0.008) and RPE_L (P = 0.001) were greater during ACE_{REL} compared to CE_{ABS}. In general, the greatest responses were observed during CE_{REL} and the lowest responses were observed during CE_{ABS} (Table 4.2). No significant interactions were observed between trials for RER ($P \geq$ 0.05) and RPE_L ($P \geq$ 0.05). However, main effects were observed for mode and time for RPE_L (P = 0.01) and RER (P = 0.01).

Table 4.2: Physiological responses during sub-maximal ACE_{REL} 50% W_{MAX}, CE_{REL} 50% W_{MAX} and CE_{ABS} at the same absolute workload ACE_{REL} exercise

	ACE_{REL}	**CE_{REL}**	**CE_{ABS}**
Power Output (W)	53 ± 8	109 ± 16	53 ± 8
$\dot{V}O_2$ ($L \cdot min^{-1}$)	1.28 ± 0.19 ^{a}	1.61 ± 0.33 ^{b}	1.19 ± 0.17
$\dot{V}O_2$ ($L \cdot min^{-1}$) as % of VO_{2peak}	49.78 ± 10	50.6 ± 11	37.3 ± 6
$\dot{V}_E$ ($L \cdot min^{-1}$)	31.8 ± 3.30 ^{a}	37.2 ± 7.00 ^{b}	27.5 ± 5.5 ^{c}
Heart Rate ($beats \cdot min^{-1}$)	122 ± 15 ^{a}	135 ± 12 ^{b}	106 ± 9 ^{c}
Heart Rate ($beats \cdot min^{-1}$) as % of HR_{MAX}	71 ± 9	74 ± 5	58 ± 5
Respiratory exchange ratio	0.93 ± 0.14	0.88 ± .011	0.95 ± 0.15
Blood lactate ($mmol \cdot L^{-1}$)	3.43 ± 1.22	2.97 ± 1.17 ^{b}	1.75 ± 0.35 ^{c}
Blood lactate 5 min recovery ($mmol \cdot L^{-1}$)	3.02 ± 1.0 ^{a}	2.53 ± 1.07 ^{b}	1.74 ± 0.40 ^{c}
RPE_L	14 ± 2	13 ± 2	10 ± 1
RPE_C	10 ± 2 ^{a}	11 ± 2 ^{b}	9 ± 1 ^{c}

* Data presented are based on mean values during final 30 seconds of the exercise bout ^{a} Significantly different between ACE_{REL} and CE_{REL} (P <0.05) ^{b} Significantly different between CE_{REL} and CE_{ABS} (P <0.05) ^{c} Significantly different between ACE and CE_{ABS} (P <0.05)

4.3.3 Postural sway responses to maximal incremental exercise

4.3.3i Kinematics

Upper and lower body maximal exercise had different effects on postural sway (Table 4.3). Normal sway exhibited during the resting conditions did not change following maximal ACE ($P \geq 0.05$). Absolute postural sway values in both the anteroposterior and mediolateral planes increased at all segmental levels following maximal lower body exercise when compared with pre exercise conditions. In the anteroposterior plane, with the exception of the ankle segment, maximal CE increased the displacement of all joint segments (all $P \leq 0.05$). The trend for greater amplitudes increased in a distal to proximal order (from feet the shoulder) following CE. Kinematic variables generally returned to pre exercise conditions by 10 and 15 minutes post exercise for EO and EC visual conditions respectively.

Table 4.3: Mean ± SD of maximal joint segment displacements in the anteroposterior and mediolateral planes before and after upper and lower body maximal exercise

	ACE		**CE**	
	Pre-Exercise	**Post Exercise**	**Pre-Exercise**	**Post Exercise**
Shoulder				
AP (mm)	12.60±3.60 †	10.79±3.11 †	12.39±3.78 *†	18.23±10.29 †
ML (mm)	6.32±1.40	6.65±3.56	6.35±1.84	12.04±4.35
Pelvis				
AP (mm)	8.13±2.30 †	8.81±4.13 †	8.12±3.26 *†	16.91±8.98 †
ML (mm)	6.32±1.611	6.19±1.32	5.37±1.54	15.37±6.62
Knee				
AP (mm)	6.30±1.09 †	6.67±1.52 †	6.22±1.58 *†	10.19±5.41 †
ML (mm)	3.63±1.07	3.98±1.34	3.43±0.78	4.78±1.62

*Significantly different between pre and post exercise; † significantly different with the next segment

4.3.3ii Centre of Pressure

Mode × time interactions were observed between the maximal exercise trials for COP_L (EO; *P* =0.008, EC; *P* =0.001), COP_{AP} (EO; *P* =0.001, EC; *P* =0.001) and COP_{ML} (EO; *P* =0.001, and EC; *P* =0.001) (Figure 4.3). Post hoc analysis revealed that CE elicited an increase in postural sway, while no changes were observed for COP measures following ACE. Immediately following maximal CE, there was a mean increase in the COP_L of 16.5 ± 9.1 and 15.7 ±7.8 cm for EO and EC conditions respectively, an increase in anteroposterior COP displacement of 1.45 ± 0.80 cm and 0.90 ± 0.39 cm for EO and EC conditions and an increase in COP_{ML} of 0.21 ± 0.15 and 0.25 ± 0.18 cm for EO and EC when compared to pre exercise values. Post hoc analysis revealed that COP_L and COP_{AP} were significantly greater than pre exercise values under both EO and EC conditions up to 15 min post exercise and 5 min post exercise for COP_{ML} following maximal CE. Although absolute sway variables were greater during EC conditions ($P \leq 0.05$), a similar trend was observed in both visual conditions, and no significant interaction between visual condition and trials were observed ($P \geq 0.05$).

4.3.4 Postural sway responses to absolute and relative submaximal exercise trials

4.3.4 I Joint Kinematics

No changes in joint centre movements were observed following ACE ($P \geq 0.05$). In contrast, significant interactions were observed for all joint segments along the AP plane following CE_{REL} and CE_{ABS} during both EO and EC conditions ($P \leq 0.05$), with the exception of the ankle, where no changes were observed for either exercise intensity. Disturbances to postural control were proportional to CE intensity. Postural sway along the sagittal plane returned to baseline values within 5-11 min following CE_{REL} and 3-6 min following $CE_{ABS.}$

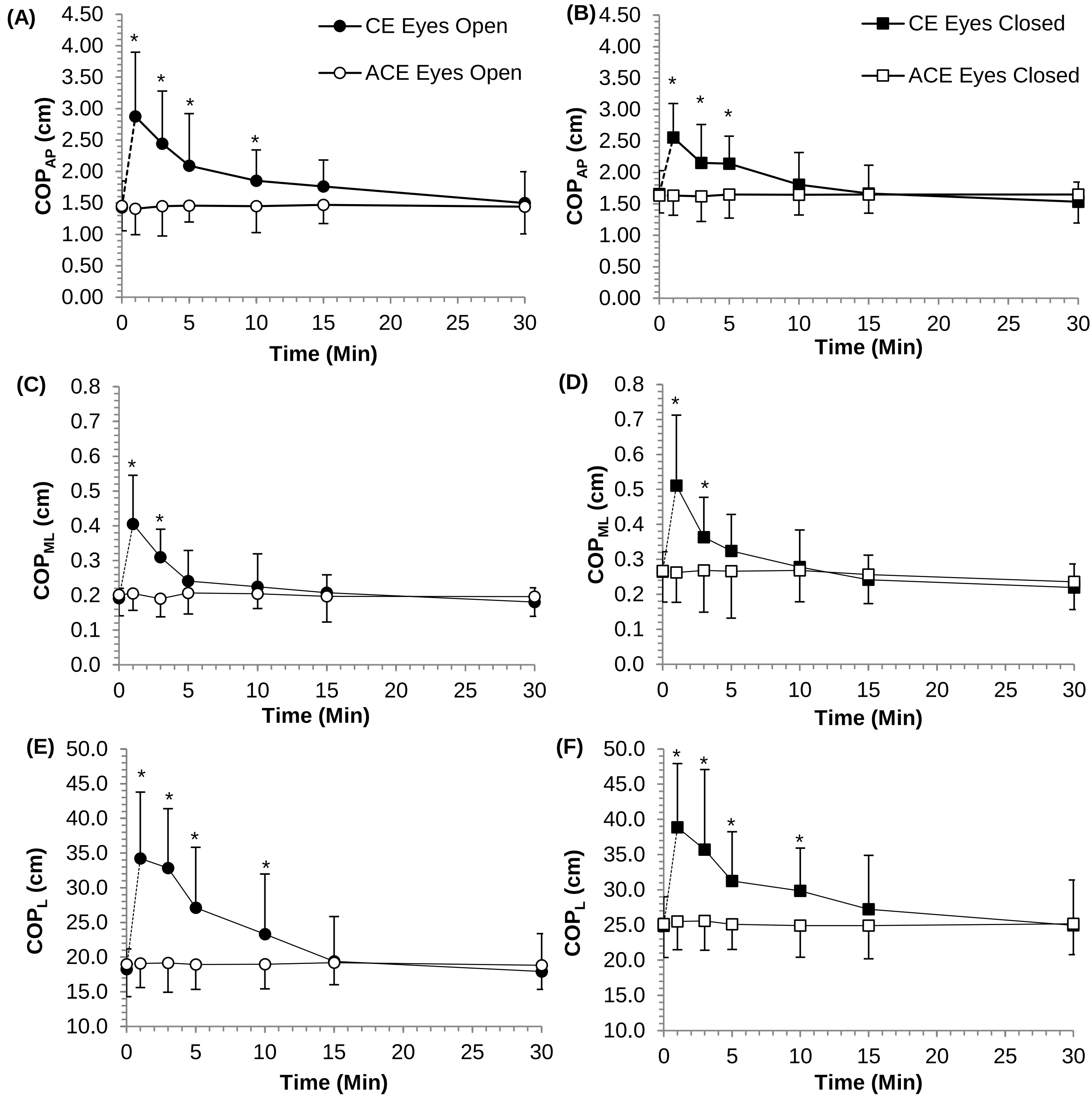


Figure 4.4. Effects of incremental upper and lower body exercise to exhaustion on COP measures of postural sway. The plots show the time course effects of exercise on the COP measures of postural sway during eyes open (left) and eyes closed (right) conditions. * Indicates significantly different from ACE ($P \leq 0.05$). Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

4.3.4 II Centre of Pressure

Mode $\times$ time interactions were observed for COP_L during EO (P = 0.047) and EC (P = 0.003) conditions and COP_{AP} for EO (P = 0.005) and EC conditions (P =0.001). The data presented in Figure 4.4 shows that CE disturbed postural sway, while no changes were detected following ACE.

Immediately following CE_{REL}, there was a mean increase in the COP_L of 11.26 ± 4.62 cm and 9.15 ± 6.62 cm for EO and EC conditions respectively, and an increase following CE_{ABS} of 4.84 ± 2.77 and 6.86 ± 2.44 cm for EO and EC conditions, respectively. In addition, immediately after CE_{REL} the COP_{AP} increased by 1.23 ± 0.60 cm and 1.14 ± 0.65 cm during EO and EC conditions respectively. The COP_{AP} increased to a lesser extent following CE_{ABS} during EO (0.95 ± 0.38 cm) and EC (1.0 ± 0.21 cm) conditions.

There were no significant main effects for time or mode for COP_{ML} following submaximal exercise. The COP_L and COP_{AP} were significant with respect to pre exercise conditions up until 5 min post exercise. Typically, increases in postural sway after cycling were proportional to the exercise intensity (Table 4.4). Furthermore, relative changes in postural sway were greater from pre to post exercise during EO compared to EC conditions. Despite greater relative percentage changes, absolute sway values were greater during EC compared to EO conditions. Individual changes in postural sway after submaximal exercise trials are illustrated in Figure 4.5.

Table 4.4 Percentage changes in COP measures from pre (0) to immediately post (1) CE

Measure	**Visual Condition**	**CE_{MAX} (% Δ)**	**CE_{REL} (% Δ)**	**CE_{ABS} (% Δ)**
COP_L	EO	90.3 ± 54.5*	60.0 ± 32.8	24.0 ± 11.7
	EC	57.2 ± 31.8	39.7 ± 37.0	28.0 ± 12.9
COP_{AP}	EO	104.3 ± 53.4*	98.4 ± 50.9	78.8 ± 48.7
	EC	56.9 ± 32.2	98.0 ± 42.7	67.0 ± 25.6
COP_{ML}	EO	116.8 ± 88.8*	22.2 ± 35.4	16.4 ± 38.1
	EC	92.4 ± 66.1	14.8 ± 34.3	5.7 ± 51.2

*Sig. difference between visual conditions

4.3.4 iii Electromyography

Electromyographic responses (n = 1) to submaximal upper and lower body exercise are illustrated in Figure 4.6 – 4.7. Muscle activation of the EMG_{GS} increased immediately after CE. The increased muscle activation returned to baseline values after ~ 5-min of exercise completion. Interestingly, an increase in the amplitude of the EMG_{GS} was also observed after ACE for both legs. Muscle activation appeared to return to below baseline levels after ~ 3-mins. For the EMG_{TA} and EMG_{RF}, no increases in muscle activation were obvious after ACE. However, in general, the level of muscle activation after CE was proportional to the exercise intensity. For the EMG_{BF}, muscle activation remained consistent with baseline levels following all modes of exercise, suggesting little contribution of this muscle. As expected, an increase in muscle activation of the EMG_{ES} and EMG_{TR} was observed after ACE, remaining elevated for the entire 30-min recovery period. No changes in muscle activation of the EMG_{ES} and EMG_{TR} were observed after CE.

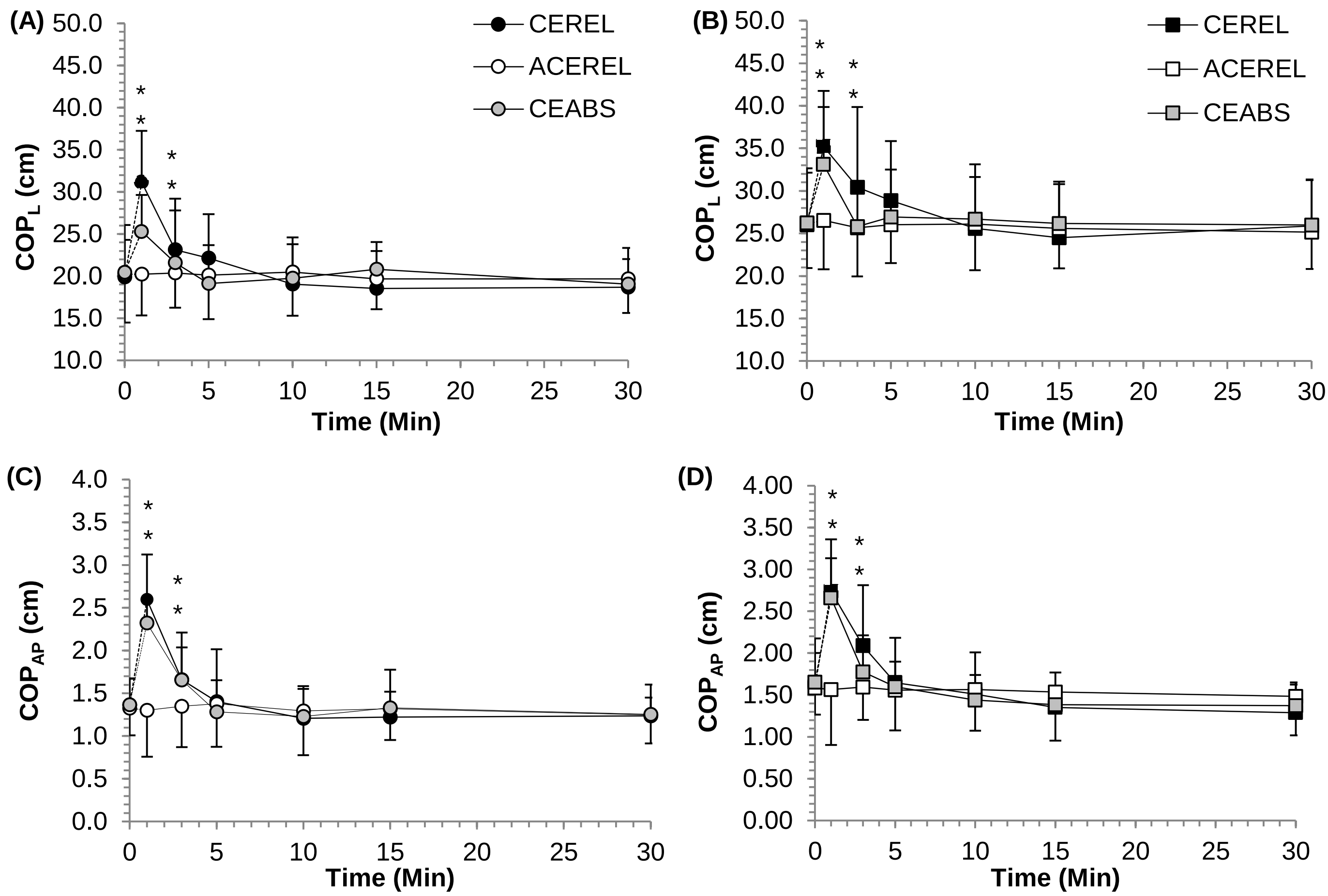


Figure 4.5: Effects of submaximal exercise on COP measures of postural sway. Plots of left shows time course effects of exercise with eyes open, while the plots on the right shows time course effects of exercise with eyes closed. * Indicates significant effects for COP$_L$ and COP$_{AP}$. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

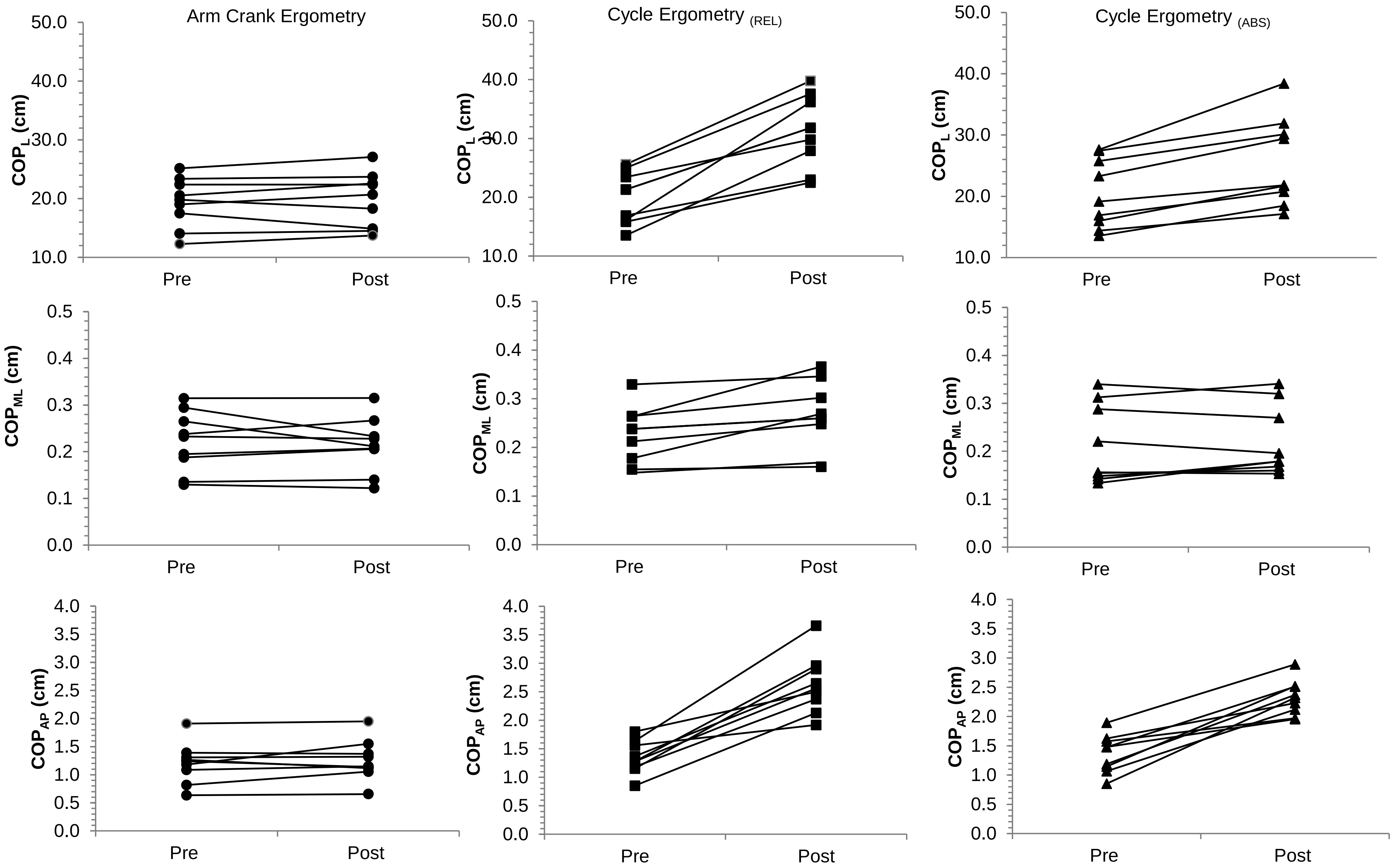


Figure 4.6: Single data line plots between pre (0) and immediately post exercise during EO conditions for ACE_{REL} (left), CE_{REL} (middle) and CE_{ABS} (right)

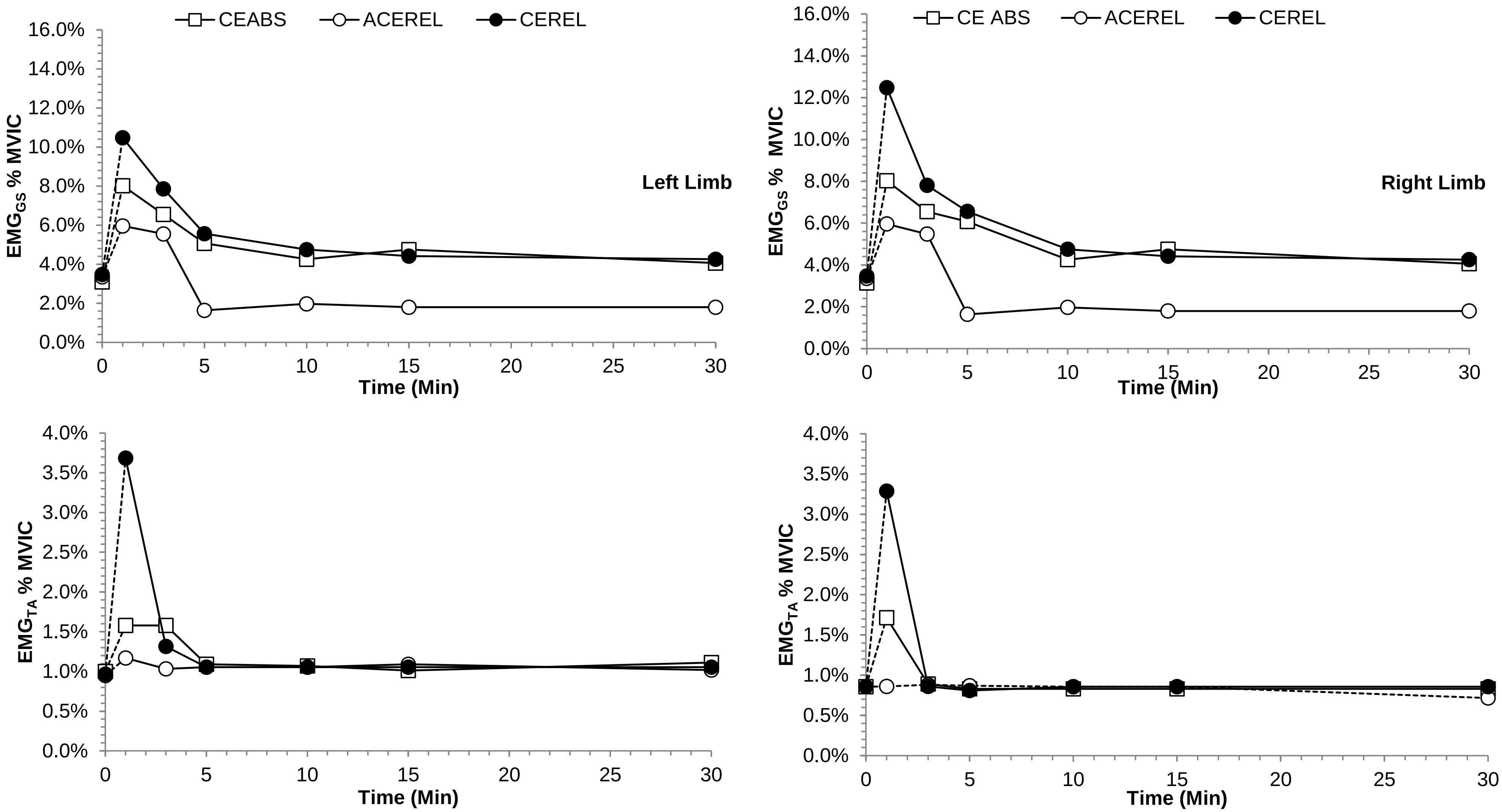


Figure 4.7: Gastrocnemius medialis (top) and tibialis anterior (bottom) muscle activation before (0) and during a 30 min recovery from submaximal exercise. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

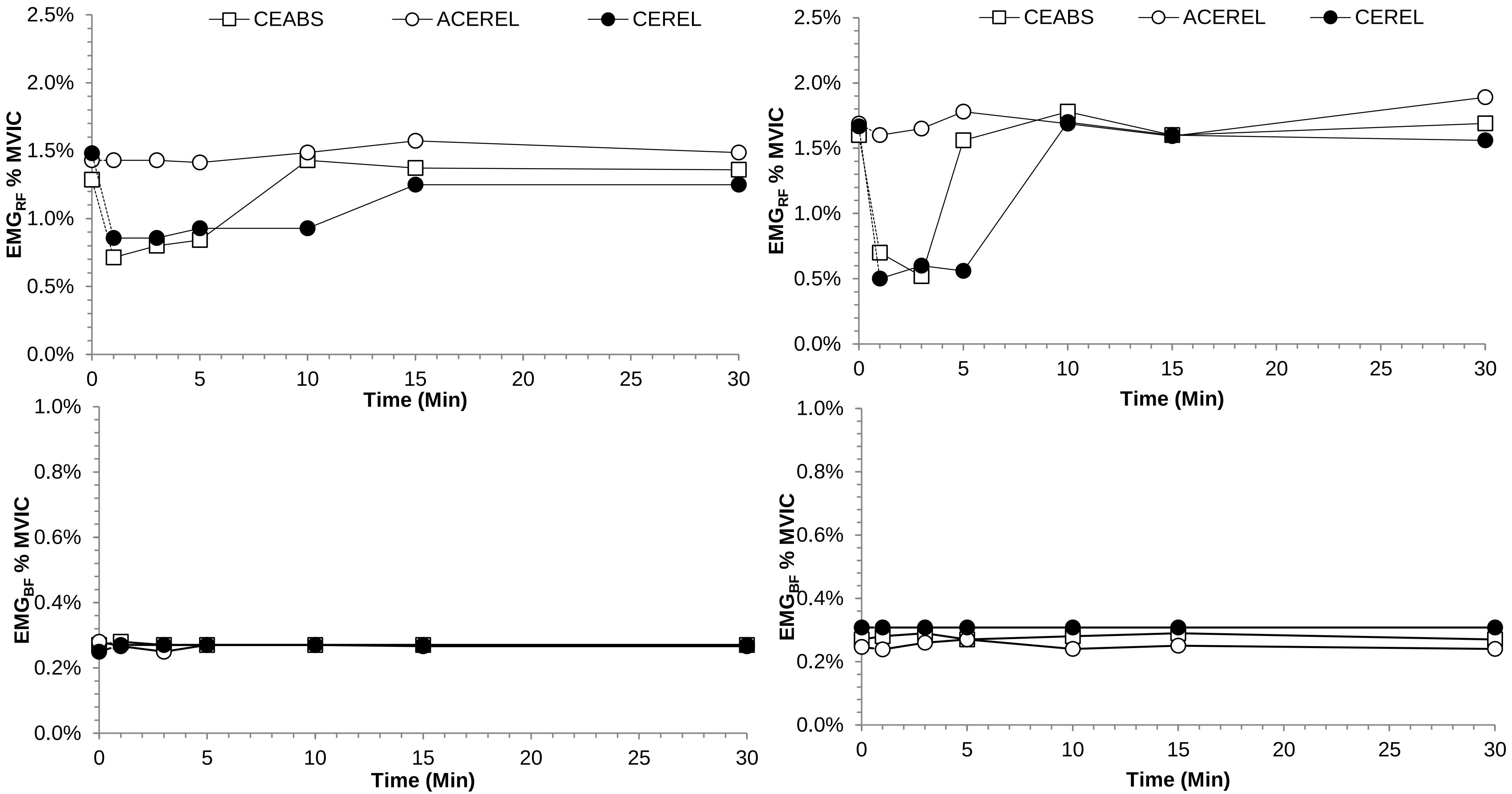


Figure 4.8: Rectus femoris (top) and biceps femoris (bottom) muscle activation before and during a 30 min recovery from submaximal exercise. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

4.4 Discussion

The present study investigated the effects of upper and lower body exercise on centre of pressure (COP) measures of postural sway during bipedal stance. Upper body exercise did not elicit any increases in COP measures of postural sway when compared to pre-exercise conditions, suggesting that ACE does not disturb bipedal postural sway following short term maximal exercise and longer duration submaximal exercise. On the contrary, CE resulted in an increase in COP measures of postural sway post exercise when performed maximally and at the same relative and absolute intensity as ACE. Therefore, changes in the control of postural sway post exercise appear to be specific to the recruited muscle mass engaging the lower body musculature involved more directly in balance control. This study provides novel findings in the context that we have identified a mode of exercise which does not induce any post exercise balance impairment in young healthy adults.

4.4.1 Physiological responses to upper and lower body exercise

4.4.1 i Peak physiological responses

Mean $\dot{V}O_{2PEAK}$, HR_{MAX} and W_{MAX} were similar to previous studies using arm crank ergometry (Price *et al.* 2011; Pimental *et al.* 1984; Smith *et al.* 2001). For example, in a similar cohort of participants, Price et al. (2011) observed $\dot{V}O_{2PEAK}$, HR_{MAX} and W_{MAX} values of 2.47 ± 0.42 $L{\cdot}min^{-1}$, 184 ± 7 $beats{\cdot}min^{-1}$ and 137 ± 38 W, respectively. In well trained men upper body $\dot{V}O_{2PEAK}$ can exceed 3.5 $L{\cdot}min^{-1}$, while W_{MAX} generally ranges between 170 – 270 W (Smith and Price, 2007). In a cohort of well-trained rowers, upper body $\dot{V}O_{2PEAK}$, HR_{MAX} and W_{MAX} values of 4.15 ± 0.24m $L{\cdot}min^{-1}$, 176 ± 8 $beats{\cdot}min^{-1}$ and 262 ± 13 W have been reported (Voliantis *et al.* 2003). Peak lower body exercise values were also similar with previous research (Fairbarn *et al.* 1994). In a group of healthy, but non-specifically trained individuals, Fairbarn *et al.* (1994) reported $\dot{V}O_{2PEAK}$, HR_{MAX} and W_{MAX} values of 3.58 ± 0.77 $L{\cdot}min^{-1}$, 185 ±

14 beats·min^{-1} and 255 ± 50 W respectively. In contrast, in a group of well-trained cyclists $\dot{V}O_{2PEAK}$, HR_{MAX} and W_{MAX} values of 4.48 ± 0.48 L·min^{-1}, 193 ± 5.89 beats·min^{-1} and 401 ± 45 W have been reported (Palmer *et al.* 1999). These peak physiological responses show that all data in the current study lie within the normative range reported in the literature for healthy untrained individuals.

It should be noted that the criterion for a true maximal exercise test using the lower body is an RER of >1.15 (Issekutz, Birkhead and Rodahl 1962). The observed values for RER at volitional exhaustion in study 1 were below the values expected to be achieved following CE (1.00 ± 0.09) and ACE (0.98 ± 0.06). However, since other physiological and perceptual parameters were monitored, there is strong evidence that both exercise protocols were able to elicit a maximal effort; (1) $\dot{V}O_{2peak}$ values for ACE (2.62 L-min^{-1}) and CE (3.23 L-min^{-1}) both reached typically expected values for the upper (Smith and Price 2007) and lower body (Hermansen and Andersen 1965) in untrained individuals; (2) HR attained a mean value of 173 (beats·min^{-1}) for ACE and 182 (beats·min^{-1}) for CE (corresponding to ~88% and ~93% of theoretical maximal HR (Karvonen, Kentala and Mustala 1957); (3) all subjects perceived their effort to be of maximal exertion (Mean RPE_L & RPE_C= 20), with exception of upper body exercise RPE_C (17) which is likely due to peripheral fatigue; and (4) the achievement of the recommended secondary measure of blood lactate, exceeding the maximal exercise criterion of >8-9 mmol·L^{-1} (Duncan, Howley and Johnson 1997). Therefore, all indicators of maximal effort provide strong evidence that individuals did indeed exercise to their maximum.

4.4.1 ii Comparison of maximal upper and lower body exercise

As expected, peak cardiorespiratory responses were greater during CE than ACE (Pimental, Sawka, Billings, & Trad, 1984). Peak power output, $\dot{V}O_2$ and $\dot{V}_E$ values reported during maximal ACE were 49.2 ± 8.0, 82.2 ± 11.2 and 82.5 ± 11.4 % of values reported during maximal CE. These values were generally in agreement with previous studies which have found that peak oxygen uptake for ACE is 60-80% of that during CE in untrained individuals, although at the top end of range (Astrand *et al.* 1965; Marais *et al.* 2002; Reybrouck, Heigenhauser and Faulkner 1975). The lower peak responses observed during maximal ACE were most likely a result of peripheral factors limiting exercise and greater levels of localised muscle fatigue when compared to CE (Sawka 1986). These limiting peripheral factors include the utilisation of a relatively small muscle mass during ACE in comparison to CE (Olivier *et al.* 2008) and restrictions to muscle perfusion (Sawka *et al.* 1983). Peripheral fatigue during ACE can be demonstrated by local and central RPE. During ACE, the difference between central and local RPE (~3 points) indicates that exercise cessation was attributable to localised muscular factors rather than cardiorespiratory stress, whereas for CE local and central RPE were the same (Borg, Hassmen and Lagerstrom 1987; Price et al. 2007). While peak cardiorespiratory responses were different between the protocols, both exercise modes elicited maximal effort based upon similar blood lactate concentrations ($\geq$ 8 mmol·L^{-1}), the attainment of as expected maximal heart rate and maximal local rating of perceived exertion (RPE_L) (Table 4.1).

4.4.1 iii Physiological responses to submaximal upper and lower body exercise

In accordance with previous research (Astrand *et al.* 1964; Marais *et al.* 2003; Pimental *et al.* 1984; Sawka *et al.* 1982) physiological responses to upper and lower body exercise were different at the same relative intensity. Greater absolute levels of $\dot{V}O_2$, $\dot{V}_E$, BLa and HR during CE_{REL} compared to ACE_{REL} were observed due to differences in maximal exercise capacity (Astrand *et al.* 1964; Eston and Brodie 1986; Reybrouck, Heigenhauser and Faulkner 1975; Sawka *et al.* 1983; Sawka 1986; Stenberg *et al.* 1967; Toner *et al.* 1983; Vokac *et al.* 1975). As a result, postural sway responses to submaximal exercise could not be directly compared due to greater physiological responses elicited during CE_{REL} compared to ACE_{REL}. For example, increases in postural sway following CE may have been due to greater absolute exercise intensity, rather than the modality of exercise, highlighting the need for an absolute exercise intensity comparison. All physiological and perceptual variables, with the exception of RER, were greater during CE_{REL} compared to CE_{ABS} which was due to differences in the absolute power output. While ACE_{REL} and CE_{ABS} were matched for the same absolute power output (53 ±8 W), all variables were statistically greater during upper compared to lower body exercise. However, respiratory differences were small (i.e., $\dot{V}O_2$ ~ 0.1 $L{\cdot}min^{-1}$ difference). Therefore, in general, the exercise modes were well matched. Interestingly, no significant differences were observed between exercise trials for RPE_L suggesting that trials were well matched with respect to work muscle perceived exertion. At comparable exercise intensities local and central RPE are typically greater in the arms compared to the legs (Marais et al., 2001).

4.4.2 Postural sway responses to cycle ergometry

The results from the present study confirm previous studies reporting increased anteroposterior postural sway following CE (Derave et al. 1998; Gouchard et al., 2002; Lepers et al., 1997; Mello, de Oliveira and Nadal, 2010; Nardone et al., 1997; Vuillerme and Hintzy 2007). When considering that COP_{AP} sway is controlled by sagittal plane movers (i.e., the ankle and knee) and these same muscles are engaged during cycling exercise, an increase in the COP displacement along this axis is not surprising (Vuillerme and Hintzy 2007). These findings support the evidence of directional sensitivity of postural muscles (Winter et al. 1996).

The present study shows that following maximal CE an increase in COP_{ML} of ~ 117 % (EO) and ~ 92 % (EC) was observed, but remained unchanged following submaximal CE. These results suggest that control of ML balance is not affected by moderate intensity cycling exercise up to 50% W_{MAX}. An increase in COP_{ML} following maximal CE in the present study indicates a potentially unique adaptation in balance strategy post exercise. This increase could represent a safety strategy adopted by participants to reduce fall risk following such exhaustive exercise (Burdet & Rougier 2004). While the author does not advocate that participants were at an increased risk of falling post exercise, previous research has suggested that COP_{ML} sway measures are strong predictors of imbalance and future fall risk in older adults (Piirtola and Era, 2006). It should be noted that while significant, the increase in COP_{ML} following CE was small (~0.25 cm) and therefore these findings are probably not clinically significant. However an increase in postural sway in this direction may be important in older adults where the ability to minimise postural sway is already limited.

Exercise resulting in fatigue is known to alter the regulation of sensory proprioceptive information and also decreases muscular system efficiency and force generating capacity (Windhurt 2007). Therefore, exercise affects both the sensory and motor contributions of the postural process (Vuillerme, Anziani and Rougier 2007). Proprioceptive impairment as a result of muscle fatigue could be caused by changes in discharge patterns of muscle afferents due to metabolite build up resulted in altered muscle spindle information (Hiemstra, Lo and Flower 2001). The present findings corroborate previous literature (Gouchard et al. 2002; Mello, de Oliveira, & Nadal, 2010) which reported a greater disturbance to postural sway following a maximal incremental exercise test on a cycle ergometer compared to a submaximal cycling trial. The present study found a proportional effect of exercise on postural sway, with the greatest negative effects observed after CE_{MAX}, while the least prominent effects were observed after CE_{ABS}. From a physiological perspective, ventilatory responses may provide an explanation of mechanisms which might be responsible for the proportional effect. For example, minute ventilation increased proportionally with exercise intensity. Increased activity of respiratory muscles negatively affects balance (Hodges et al. 2002).

As blood lactate increases, it causes a decrease in muscular action potential, changes calcium deployment from the sarcoplasmic reticulum and decreases muscle enzyme activity (Gribble and Hertel 2004; Ruzic et al. 2014). Indeed Ruzic et al. (2014) recently reported that quiet standing was affected progressively according to the following exercise intensities; below aerobic threshold (HR 118 beats·min^{-1}, BLa 2.0 mmol·L^{-1}), between aerobic and anaerobic threshold (HR 137 beats·min^{-1}, BLa 2.4 mmol·L^{-1}), anaerobic threshold (HR 158 beats·min^{-1}, BLa 5.5 mmol·L^{-1}), above lactate threshold (HR 168 beats·min^{-1}, 8.0 mmol·L^{-1}) and maximal effort (HR 187 beats·min^{-1}, 12.0 mmol·L^{-1}). The change in postural sway with each increase in exercise intensity was not linear. Instead, the authors reported that three distinct balance thresholds were evident; the initiation of exercise, anaerobic threshold and maximal effort. Ruzic et al. (2014) specified that the proportional negative effects according to exercise intensity were likely a result of increased ventilatory responses and elevated blood lactate responses.

The present study also provides additional data in the context of the time course effects of CE on postural sway. Previous research has shown that the negative consequences of treadmill exercise are transient and last no more than 20 min post exercise in young healthy adults (Fox et al. 2008; Yaggie & Armstrong 2004; Bove et al., 2007). While our findings are difficult to compare due to differences in exercise protocols, we show that the negative effects of CE are similar and diminish within 15 and 5 min for maximal and submaximal CE, respectively. The initial 15-min after exercise may suggest a window where the effects of exercise on postural stability are most exacerbated. This is particularly important because most athletic injuries occur in the lower body at the end of an activity when the musculature is fatigued (Yaggie and Armstrong 2004). Therefore, an appropriate time of post exercise recovery should be endorsed in order to avoid any fatigue induced risk of injuries.

4.4.3 Postural sway responses to upper body exercise

More recent investigations have considered the effects of exercise using the upper body musculature on balance control, but have demonstrated equivocal findings (Douris et al., 2011; Smith et al., 2010). In the present study, maximal and submaximal ACE had no effects on COP measures of postural sway (Fig 4.2 & 4.3) and are in accordance with those reported for healthy older adults (Smith et al., 2010). The absence of changes in postural sway post

ACE in the present study may be explained by the relative ease of standing on two legs, since this task does not provide a major challenge to the balance control system (Clifford & Holder-Powell, 2010). In line with the present study, Smith et al., (2010) showed that in healthy older adults, postural sway was not disturbed following a bout of ACE. However, Douris et al., (2011) reported that maximal aerobic ACE ($\dot{V}O_{2peak}$ test) disturbed postural sway to a greater extent than CE when standing on a single limb. According to Douris et al. (2011) following the $\dot{V}O_{2peak}$ test the trunk muscles (not alluded as to which specific muscles) were more fatigued after ACE compared to CE, and therefore these muscles were less able to assist in postural adjustments. While not obvious, the increase in postural sway following ACE may be a result of adopted stance in the sway test. During bipedal stance, postural sway is predominantly controlled by the triceps surae muscles (Loram & Lakie, 2002). However, when stance is more challenging, such as standing on a single limb, postural adjustments are made at the hip and trunk (Tropp & Odenrick, 1988). Indeed, competition between trunk musculature for balance and respiration may explain the compromise in control of ML balance. Indeed, Smith et al., (2010) showed that adults with chronic obstructive pulmonary disease demonstrated increased COP_{ML} sway after ACE. Since frontal plane balance is primarily controlled by hip and trunk movement, if the musculature in these regions are fatigued in addition to an increase in respiratory demand, balance may be compromised. The present study provides novel data in that maximal and submaximal ACE does not elicit post exercise balance impairments during quiet standing.

4.4.4 Electromyography responses

Previous research typically agrees that acute exercise deteriorates the quality of sensory proprioceptive information and/or integration (Paillard, 2012). In addition, exercise decreases muscular system efficiency (Nardone et al., 1997), thus disturbing postural sway post exercise. While limited to a single participant, this study presents novel findings in the context of neuromuscular responses to upper and lower body exercise during quiet standing.

Following ACE an immediate increase in the amplitude of the EMG_{GS} was observed. According to Paillard (2012) fatigue of the distal musculature (e.g., gastrocnemius) induces the recruitment of proximal muscles (e.g., hip musculature) to counteract its destabilising effects on postural control. Therefore, it is possible that fatigue of the proximal muscles, in particular to the trunk and shoulders, increases the activation of unaffected lower limb muscles, in this case the posterior calf musculature. The central nervous system may develop strategies to compensate for muscular fatigue. For example, several studies have observed a greater EMG amplitude and earlier onset of activation (anticipatory postural adjustments) for unfatigued muscles and weaker activation of fatigued muscles (Morris and Allison 2006; Strang and Berg 2007; Strang et al. 2009). Kanekar et al., (2008) selectively fatigued the deltoid muscles and reported that increased activation of lower limb muscles (e.g., soleus, gastrocnemius, semitendinosus and biceps femoris) were able to compensate for the fatigue of upper body muscles to maintain postural stability in bipedal stance. The earlier onset of anticipatory postural activity reported by Kanekar et al. (2008) may represent a functional adaptation by the central nervous system to preserve postural stability in the presence of fatigue. The present study is the first to report a potential compensatory strategy after exercise engaging the whole of the upper body muscle mass. These novel findings might indicate that the CNS may develop a strategy of increasing lower limb muscle activation to compensate for upper body muscular fatigue.

Furthermore, following CE at the same relative and absolute intensity as ACE, there was an immediate reduction in the amplitude of the EMG_{RF} which was accompanied by an increase in the amplitude of the EMG_{GS}. Since quiet standing does not maximally activate muscles involved in postural regulation, the CNS can increase the activation level of different muscles (Paillard 2012). The present data might indicate that the CNS purposely reduced the activation of the rectus femoris musculature and increased activation of the gastrocnemius.

4.4.5 Summary

In summary, this study contributes to the existing knowledge on post exercise postural sway by demonstrating novel findings that arm crank exercise does not elicit post exercise balance impairment which has applications to those at risk of falling such as the elderly. Furthermore, CE at both maximal and submaximal intensities resulted in a disturbance to postural sway, suggesting that exercise effects were specific to lower extremity muscles involved in balance control during quiet bipedal standing. The impact of this work is important as the current findings may enable potential risks of injury during fatiguing exercise to be anticipated and prevented. In the context of sensorimotor impairments after acute lower limb exercise, the absence of any changes in postural sway after ACE is a favourable response in that this mode of exercise may offer a safe alternative for fall risk populations. It is acknowledged that the generalisability of these results to an at risk population are limited due to the young healthy cohort used in the present study. Whether the present effects would be observed in individuals with impaired postural control (i.e., older adults) where the negative consequences of exercise may be more exacerbated remains to be investigated. The impact of upper body exercise on postural sway in older adults may be more pronounced than young adults since their proprioceptive and neuromuscular systems are less efficient (Bisson et al. 2011). Future research should therefore consider the effects of upper and lower body exercise on postural sway in a group of older but otherwise healthy adults.

Chapter 5

The Effects of Upper, Lower and Whole body Exercise on Postural Sway in Healthy Older Adults

5.1 Introduction

Maintaining balance is a complex task where the central nervous system must integrate visual, vestibular and proprioceptive information while modulating commands to the neuromuscular system (Bisson et al. 2011). Several studies have reported that older adults show increased centre of pressure (COP) displacement during quiet stance following cycle ergometry (CE) (Stemplewski et al. 2012; 2013) and treadmill walking (TM) (Donath et al. 2013) which is interpreted as a reduction in postural stability and may be predictive of falling (Egerton et al. 2009). Physical activity can elicit a number of acute effects such as muscle fatigue leading to a decreased ability to generate muscular force (Kent-Braun 1999) and deterioration in proprioceptive information and/or its integration within the central nervous system (Windhurst 2007). Therefore, acute exercise can affect both the sensory and motor contributions of the sensorimotor postural process (Horak 2006; Vuillerme, Anziani and Rougier 2007). At present, studies in older adults have considered either males (Stemplewski et al. 2012; 2013) or both males and females (Donath et al. 2013; Egerton et al. 2009). Young females are more adversely affected by acute general muscular exercise than males (Springer and Pincivero 2009) and therefore, it is important to identify whether older females are adversely affected by different forms of endurance exercise.

Presently, the relationship between upper body muscular fatigue and postural sway remains equivocal (Douris et al. 2011; Hill et al. 2014; Smith et al. 2010). The field of arm crank ergometry (ACE) is relatively unexplored, despite this mode of exercise possessing important practical applications. For example, ACE training elicits transferable training effects by improving lower body exercise capacity in both healthy older (Pogliaghi et al. 2006) and clinical groups (Tew et al. 2009). There are also indications that afferent input from the upper body plays an important role in the control of upright stance as evidenced by an increase in postural sway following localised fatigue of the neck (Schieppati, Nardone and Schmid 2003), deltoids (Nussbaum 2003) and trunk extensors (Davidson, Madigan and Nussbaum 2004). These findings are of particular interest because fatigue was induced in muscles which are not considered to a play major role in the control of quiet standing balance (Davidson, Madigan and Nussbaum 2004).

During ACE, isometric work of the abdominal musculature provides trunk stability to maintain balance and posture in the seated position (Di Blasio et al. 2009). These isometric contractions may become a limiting factor following ACE because the abdominal and intercostal muscles may compete for ventilation and balance control (Smith et al. 2010). Recently, Douris et al. (2011) found that anaerobic ACE disturbed single limb balance to a greater extent than anaerobic cycling (CE) in young adults. In contrast, opposite findings were reported for aerobic exercise. However, the Dynamic Stability Index adopted by Douris et al. (2011) is dissimilar to conventional COP measures of postural sway measured using a force platform. Therefore, it is difficult to compare their findings with previous investigations. On the other hand, Hill et al. (2014) (Chapter 4) reported that maximal and submaximal ACE does not have any destabilising effects on COP measures of postural sway during quiet bipedal standing in healthy young adults, while marked alterations in postural stability were observed after cycling. It has also been shown that the effects of submaximal ACE on postural stability might be minimal in healthy older adults (Smith et al. 2010) confirming findings in young adults (Hill et al. 2014). The impact of upper body exercise on postural sway in older adults may be more pronounced than young adults since their proprioceptive and neuromuscular systems are less efficient (Bisson et al. 2011).

Within this context, no studies have ascertained whether postural sway is affected differently by exercise modes involving postural and non-postural musculature in elderly females. The lack of knowledge in this area is surprising, considering that such exercises are prescribed for prevention of chronic disease at intensities which represent the demands experienced in daily life (Donath et al. 2013). Moreover, whilst young healthy adults need ~ 5 - 20 min recovery post-exercise to fully return to baseline performance of postural sway (Bove et al. 2007; Fox et al. 2008; Hill et al. 2014; Chapter 4; Nardone et al. 1997) it is not yet clear how long older adults require. Addressing the time course effects of exercise on postural sway are needed to estimate the open window of an exercise induced increased risk of falling. Addressing the time course effects of exercise on postural sway can provide data allowing us to estimate the window of an exercise induced increased risk of falling.

Research Hypothesis (H_2): Upper body exercise will elicit post exercise balance impairments, but to a lesser extent than cycling and walking in healthy older females.

Null Hypothesis ($H0_2$): There will be no significant changes in postural sway following arm cranking, cycling or walking in healthy older females.

5.2 Methods

5.2.1 Participants

Nine healthy older female adults (mean ± SD age, 70.2 ± 7.8y; height, 1.56 ± 0.10m; mass, 64.8 ± 6.4kg; BMI 26.8 ± 3.3) volunteered to take part in this study. All procedures were reviewed and ethical approval was received by the University Ethics Committee. After being informed of the nature and requirements of the study each participant provided written informed consent prior to any involvement. Based on the physical activity readiness questionnaire (PAR-Q), medical history was recorded by the principal investigator. All participants were sedentary and not meeting exercising guidelines (e.g., exercise less than three times per week for less than 20-min each time at an intensity below 50% HR_{MAX}). None of the participants presented any history of cardiovascular or pulmonary diseases, neurological and vestibular disorders, orthopaedic pathology or musculoskeletal problems. Participants were considered eligible if they were over 60 years of age, community dwelling and living independently. All participants considered for testing were then required to score $\geq$52/56 on the Berg Balance Scale (BBS) (Berg et al. 1992). None of the participants reported to have experienced any falls due to loss of balance in the past 12 months. However, two participants did report to falling in the past year due to tripping. Those participants were not excluded from this study as individuals who report tripping do not exhibit greater levels of postural sway compared to those who fall due to loss of balance.

5.2.2 Study design

All participants completed six separate exercise tests, each separated by a total of 72 hours. The tests were performed at the same time of day to avoid any circadian rhythm effects on balance. Participants initially randomly completed three incremental exercise tests for ACE, CE and TM to predict HR_{MAX} and to determine the effects of high intensity exercise on postural sway. Following these preliminary trials, all participants completed three subsequent tests of 20-min duration for ACE, CE and TM at a relative workload corresponding to 50% of the ergometer-specific heart rate reserve (HR_E). Trials were performed in a randomised order.

5.2.3 Preliminary tests

Each individual performed an incremental step test on an arm crank ergometer (Lode Angio BV, Groningen, Netherlands), cycle ergometer (Monark, 824E, Ergomedic, Sweden) and treadmill (Powerjog GX100, Birmingham, UK). Due to concerns regarding participant safety due to age, the exercise test was terminated by the principal investigator when participants attained a HR of 85 % of the age-predicted maximum (Tanaka 2001). Specific protocols for

ACE and CE are outlined in section 3.7.1. For the treadmill walking (TM) trial participants started with a 5 minute warm up at 1.5 km·h^{-1} and a 0% grade. Following a warm up, walking speed was increased to 3.2 km·h^{-1}, with an initial grade of 0%. The treadmill grade was then increased by 1% every minute until the desired HR was reached.

For all trials and modes, once participants had reached 85 % HR_{MAX} they were asked to maintain the intensity until the end of the current exercise stage. These protocols have previously been successfully used to elicit peak responses in older adults with reduced lower body exercise capacity (Tew et al., 2009). All participants met at least one of the following criteria for exercise cessation; 1) achievement of 85% HR_{MAX} 2) local and/or central RPE of 15-17 or 3) or an RER of >1.1 (Borg 1982; Cress et al. 1991).

5.2.4 Experimental exercise tests

At least 72 hours after the preliminary trials participants visited the laboratory on three further occasions to perform a 20-min submaximal, mode specific exercise test. Tests were performed in a randomised order separated by 72 hours. For both ACE and CE participants began with a 5-min warm up at 5 and 10 W, respectively, at a cadence of 60 rev·min^{-1}. For the TM trial participants started with a 5 minute warm up at 1.5 km/h and a 0% grade. The warm up was consistent with that used in the preliminary trials. Each experimental trial was performed at a power output corresponding to 50% of each individuals heart rate reserve (HR_E). Heart rate reserve was calculated using Karvonen's formula (Karvonen and Vuorimaa 1988) which takes into account resting heart rate (RHR), the participant's age and the required exercise intensity (i.e., 50%), for example;

$$HR_E = (208 – 0.7 \text{ x age} – \text{RHR}) \text{ x \% intensity (fraction of preliminary trial)} + \text{RHR}.$$

In accordance with section 3.8.4, 20 beats·min^{-1} were subtracted to predict HR_{MAX} for ACE to account for the smaller muscle mass. Expired gas was continually recorded at rest and during exercise. Values were averaged in the final 30 s of each 5 minute interval. Heart rate was also closely monitored and recorded along with ratings of RPE_L and RPE_C every 5 min.

5.2.5 Physiological Measurements

Expired gas was continuously analysed using a breath-by-breath gas analyser system (MetaMax, Cortex Biophsik, Borsdorf, Germany) for oxygen uptake ($\dot{V}O2$), minute ventilation (V_E), respiratory exchange ratio (RER), breath frequency (Bf) and tidal volume (Vt). Heart rate was continuously monitored using a heart rate monitor (Polar Electro, Oy, Finland). A rating of both local (RPE_L; working muscles) and central (RPE_C; cardiorespiratory stress) perceived exertion (RPE) using the 6-20 point Borg scale (Borg 1973) was used to record subjective perceptions of exertion during exercise. To determine HR_E, resting heart rate was closely monitored in each participant prior to exercise trials by requesting participants to sit quietly for 15 – 20 minutes in a room with only the experimenter present. Heart rate was monitored every 2 minutes, with the consistently lowest heart rate used to calculate 50% heart rate reserve (HR_E). Power output on the treadmill was calculated using the following equation; W = 0.1634 · Speed (m·min^{-1}) · (Grade/100) · Body mass (Cooper and Storer 2001).

5.2.6 Postural sway

Posturographic trials are reported in section 3.9.3. Analysis of COP data lead to the computation of main sway parameters; COP path length (COP_L) and COP displacement in the anteroposterior (COP_{AP}) and mediolateral (COP_{ML}) directions. These parameters of postural sway were used for comparative proposes with previous investigations. The validity and reliability of these parameters computed with a force platform have been accepted (Pinsault and Vuillerme 2009).

5.2.7 Berg Balance Scale

The 14-item Berg Balance Scale (BBS) (Berg et al. 1992) was administered to each participant by the principal investigator who scored each item in keeping with the operational definitions of each item. The final score is interpreted as follows; high fall risk 0 – 20, medium fall risk 21 – 40 and low fall risk 41 – 56. However, it has previously been shown that scoring $\leq$ 52/56 on the Berg Balance Scale (BBS) can classify older adults as balance impaired (Brauer, Woollacott and Shumway-Cook 2001). The BBS was administered three times in total, prior to each experimental trial. None of the participants scored $\leq$ 52 during any of the assessments (trial 1, 54 ± 2; trial 2, 54 ± 1; trial 3, 54 ± 1).

5.2.8 Statistical analysis

Data were tested for normality (Shapiro-Wilk) and homogeneity of variance (Levene's Test). Peak physiological and perceptual responses to the preliminary exercise tests were analysed using a One-way ANOVA. Additionally, a two-way analysis of variance (ANOVA) with repeated measures on both factors (time ^{x} mode) was conducted to examine changes induced by exercise on each sway measure (e.g., mode; ACE, CE and TM x time; pre (0), 1, 3, 5, 10, 15 and 30 min post exercise) (Figure 5.1). Two-way ANOVA was also conducted to examine differences in physiological variables between protocols over time for each of the submaximal exercise protocols. Where the result of the ANOVA was statistically significant, Tukey's Honestly Significant Difference (HSD) post hoc analysis was conducted to determine differences. Data was analysed using IBM version 20.0 (SPSS Inc., Chicago, IL). Statistical significance was set at $P \leq 0.05$. Percentage changes between pre and post exercise postural sway (baseline and immediately post) were also calculated and analysed. Comparison of postural sway responses will also be reported between young adults (Chapter 4) and older adults in the present study.

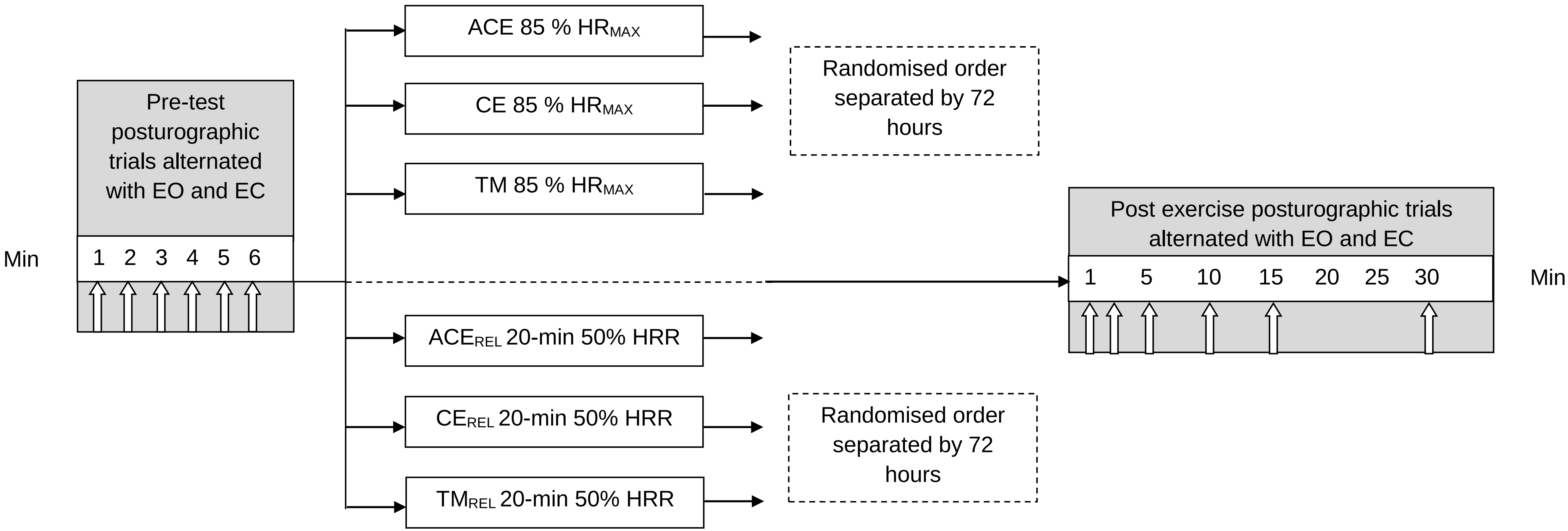


Figure 5.1: Experimental setup indicating posturographic trials before and after incremental and submaximal exercise trials.

5.3 Results

5.3.1 Physiological responses to incremental exercise

Table 5.1: Mean ±SD physiological responses and perceptions of exertion during incremental ACE, CE and TM trials to 85 % HR_{MAX}

	ACE	CE	TM
Power Output (W)	30 ± 8	65 ± 18	59 ± 25
Absolute $\dot{V}O_2$ ($L \cdot min^{-1}$)	0.83 ± 0.12 $^{a, c}$	1.11 ± 0.09 ^{b}	1.25 ± 0.10
Predicted absolute $\dot{V}O_{2MAX}$ ($L \cdot min^{-1}$)	1.11 ± 0.24 $^{a, c}$	1.33 ± 0.11 ^{b}	1.54 ± 0.27
Relative $\dot{V}O_{2MAX}$ (ml/min/kg)	12.9 ± 2.1 $^{a, c}$	17.3 ± 2.1 ^{b}	19.6 ± 0.9
Predicted relative $\dot{V}O_{2MAX}$ (ml/min/kg)	17.2 ± 3.9 $^{a, c}$	20.6 ± 2.0 ^{b}	24.1 ± 5.3
$\dot{V}_E$ ($L \cdot min^{-1}$)	28.7 ± 7.1	35.2 ± 6.3	32.8 ± 3.3
RER	1.07 ± 0.11	1.09 ± 0.06	1.08 ± 0.03
HR ($beats \cdot min^{-1}$)	112 ± 9 $^{a, c}$	131 ± 6	131 ± 7
Target HR (85% HR_{MAX})	110 ± 7	127 ± 7	127 ± 7
Time to THR (min)	7.05 ± 1.42	7.44 ± 1.51	8.02 ± 3.1
RPE_L	17 ± 1	17 ± 1	16 ± 1
RPE_C	15 ± 1	15 ± 1	16 ± 1

^{a} Significantly different between ACE and CE ($P \leq 0.05$)

^{b} Significantly different between CE and TM ($P \leq 0.05$)

^{c} Significantly different between ACE and TM ($P \leq 0.05$)

Summary data for incremental exercise are presented In Table 5.1. Oxygen uptake ($\dot{V}O_2$) (P = 0.001) and HR (P = 0.001) differed significantly across the three exercise modes, while approached significance for $\dot{V}_E$ (P = 0.055). $\dot{V}O_2$ was greater during TM compared to CE (P =

0.007) and ACE (P = 0.001). Likewise, $\dot{V}O_2$ was greater for CE compared to ACE (P = 0.002). Heart rate was the same at the end of CE and TM (P = 0.898) but was significantly lower during ACE compared to CE and TM (both P = 0.001). The lower peak HR was primarily due to an adjusted HR_{MAX} prediction for the arms. However, trials were matched well for both RPE_L and RPE_C where there was a 1 point different between exercise modes ($P \geq 0.05$). Moreover, RER was not different between any of the three trials ($P \geq 0.05$).

5.3.2 Physiological responses to submaximal exercise

Table 5.2: Mean ± SD physiological responses and perceptions of exertion during ACE, CE and TM trials at 50% HR_E

	ACE	**CE**	**TM**
Power Output (W)	18 ± 12	40 ± 20	29 ± 12
$\dot{V}O_2$ (L·min^{-1})	0.63 ± 0.13 $^{a, c}$	0.85 ± 0.18 ^{b}	0.98 ± 0.20
$\dot{V}_E$ (L·min^{-1})	20.3 ± 3.5	25.2 ± 4.6	25.9 ± 8.7
RER	0.92 ± 0.03	0.90 ± 0.05	0.91 ± 0.04
Breath frequency (breaths·min^{-1})	28 ± 5	29 ± 5	28 ± 5
Tidal Volume (L)	0.92 ± 0.29	0.92 ± 0.28	0.95 ± 0.22
HR_E (beats·min^{-1})	98 ± 8	108 ± 8	108 ± 8
RPE_L	16±1 ^{c}	15±1	15±1
RPE_C	14±1 ^{c}	14±1	15±1

* Data presented are based on mean values during final 20 seconds of the exercise bout, ^{a} Significantly different between ACE and CE ($P \leq 0.05$), ^{b} Significantly different between CE and TM ($P \leq 0.05$), ^{c} Significantly different between ACE and TM ($P \leq 0.05$)

An interaction of mode ^{x} time was found to be significant for $\dot{V}O_2$ (P = 0.003), RPE_L (P = 0.004) and RPE_C (P = 0.001). A main mode effect was observed for RPE_L (P = 0.006). $\dot{V}O_2$ was proportional to the engaged muscle mass. Local RPE was greatest for ACE, and lowest for TM however, this was only a 1 point difference. In contrast, central RPE was lowest for ACE and CE and greatest for TM, which likely reflects the recruited muscle mass. Actual HR was ~ 10 beats·min^{-1} greater than HR_E for ACE and CE, but was only ~ 5 beats·min^{-1} greater for TM, which again likely reflects individuals habituation for walking. Exercise trials were closely matched for RER, with just a 0.01 difference between the three exercise modes ($P \geq 0.05$).

5.3.3 Postural sway responses following incremental exercise

No significant mode ^{x} time interactions were observed for COP_L (EO; P = 0.99, EC; P = 0.147), COP_{AP} (EO; P = 0.442, EC; P = 0.251) and COP_{ML} (EO; P = 0.709, EC; P = 0.187). Significant main effects of time were observed for COP path length (EO; P = 0.007, EC; P = 0.049) and COP_{ML} (EO; P = 0.002, EC; P = 0.001) (Figure 5.2). Post hoc comparisons revealed that COP_L and COP_{ML} values returned to baseline conditions within 3 and 5 min for CE and TM trials, respectively. Post hoc analyses revealed that COP measures were not affected by ACE ($P \geq$ 0.05). Although Figure 5.2 shows that a CE and TM elicited a disturbance to COP variables, the lack of significance might be explained by the large standard deviations.

5.3.4 Postural sway responses to submaximal exercise trials

Significant mode ^{x} time interactions were observed for COP_L (EO; P = 0.001, EC; P = 0.001), COP_{AP} (EO; P = 0.001, EC; P = 0.001) and COP_{ML} (EO; P = 0.001, EC; P = 0.001) (Figure 5.3). Post hoc analyses revealed that following cycling and walking, ll COP measures increased from baseline levels until 5-min post exercise under EO conditions, and 10-min post exercise for EC conditions ($P \leq 0.05$). Arm crank ergometry had no significant effects on any of the COP measures ($P \geq 0.05$). Percentage increases in COP_{AP} and COP_{ML} from baseline

(0 min) to the first post exercise trial (1 min post) were greater following CE and TM for 20-min at 50 % HRE compared to incremental exercise to 85 % HR_{MAX} ($P \leq 0.05$) (Table 3). The COP_{AP} and COP_{ML} were similar after CE and TM. However, the TM trial disturbed for COP_L to a greater extent than CE for EO ($P = 0.035$) and EC ($P = 0.019$) conditions. Individual data plots are also reported to illustrate changes in postural sway within and between subjects.

Table 5.3: Percentage change (% Δ) in postural sway measures from pre (0) to immediately post CE and TM. Note; ACE values not included due to non-signifciant changes

COP variable	Visual condition	CE 85 % HR_{MAX} (% Δ)	TM 85 % HR_{MAX} (% Δ)
COP_L	EO	7.98 ± 6.93	5.67 ± 4.22
	EC	4.96 ± 8.30	4.80 ± 6.95
COP_{AP}	EO	19.83 ± 14.46	45.31 ± 47.91
	EC	17.99 ± 19.24	31.96 ± 23.68
COP_{ML}	EO	31.17 ± 30.30	67.17 ± 110.25
	EC	22.82 ± 15.24	50.22 ± 37.09
		CE 50 % HR_E (% Δ)	**TM 50 % HR_E (% Δ)**
COP_L	EO	10.03 ± 9.52	12.02 ± 6.87
	EC	12.25 ± 9.90	7.63 ± 11.27
COP_{AP}	EO	47.96 ± 26.23	83.20 ± 36.78
	EC	50.20 ± 45.85	62.27 ± 24.17
COP_{ML}	EO	84.66 ± 55.16	121.01 ± 44.10
	EC	52.51 ± 25.45	102.08 ± 88.71

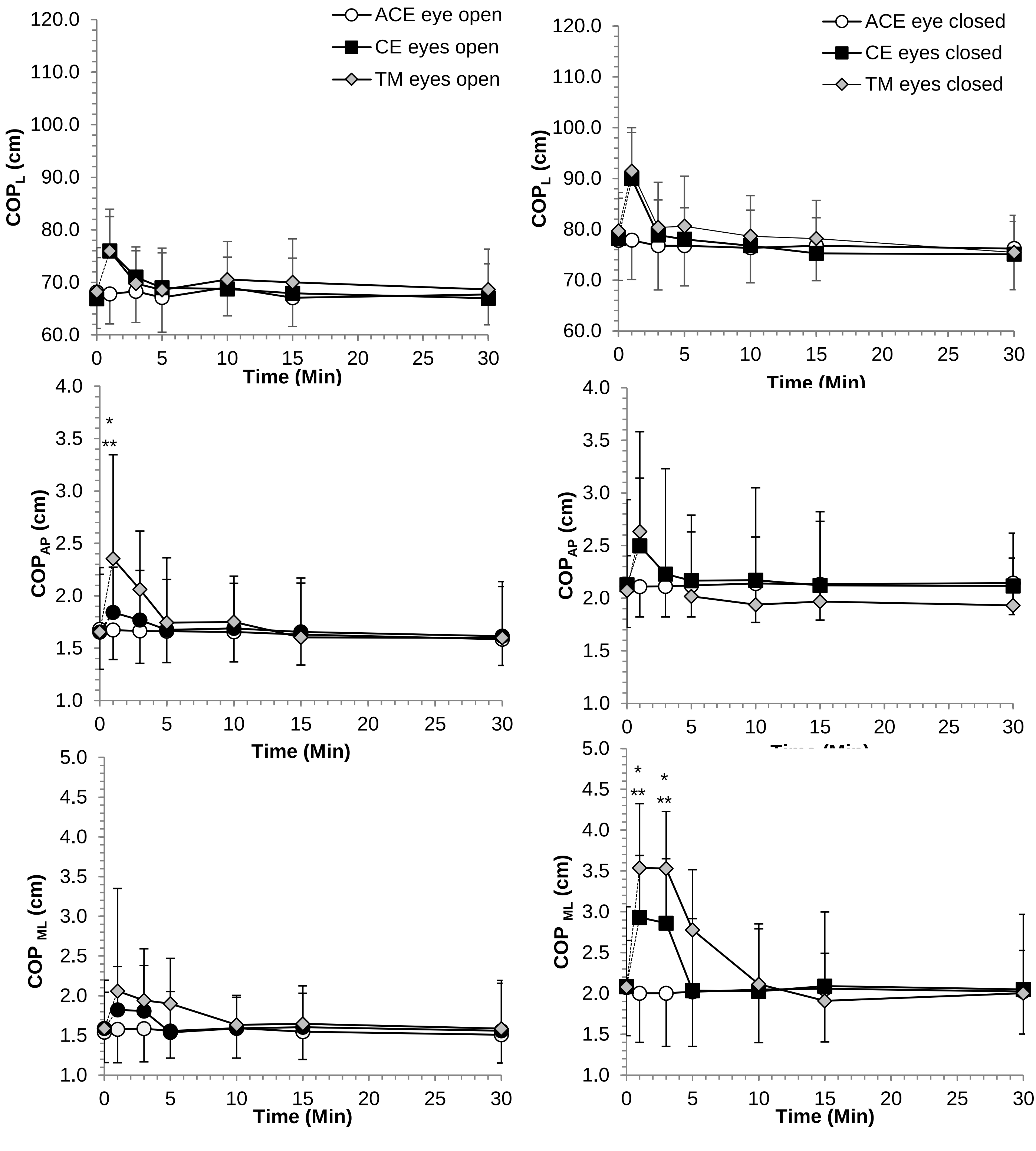


Figure 5.2: Mean ±SD change in COP measures of postural sway from baseline (0) to post incremental exercise for ACE, CE and TM. * Significantly different to baseline trial for TM. ** Significantly different to baseline trial for CE. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

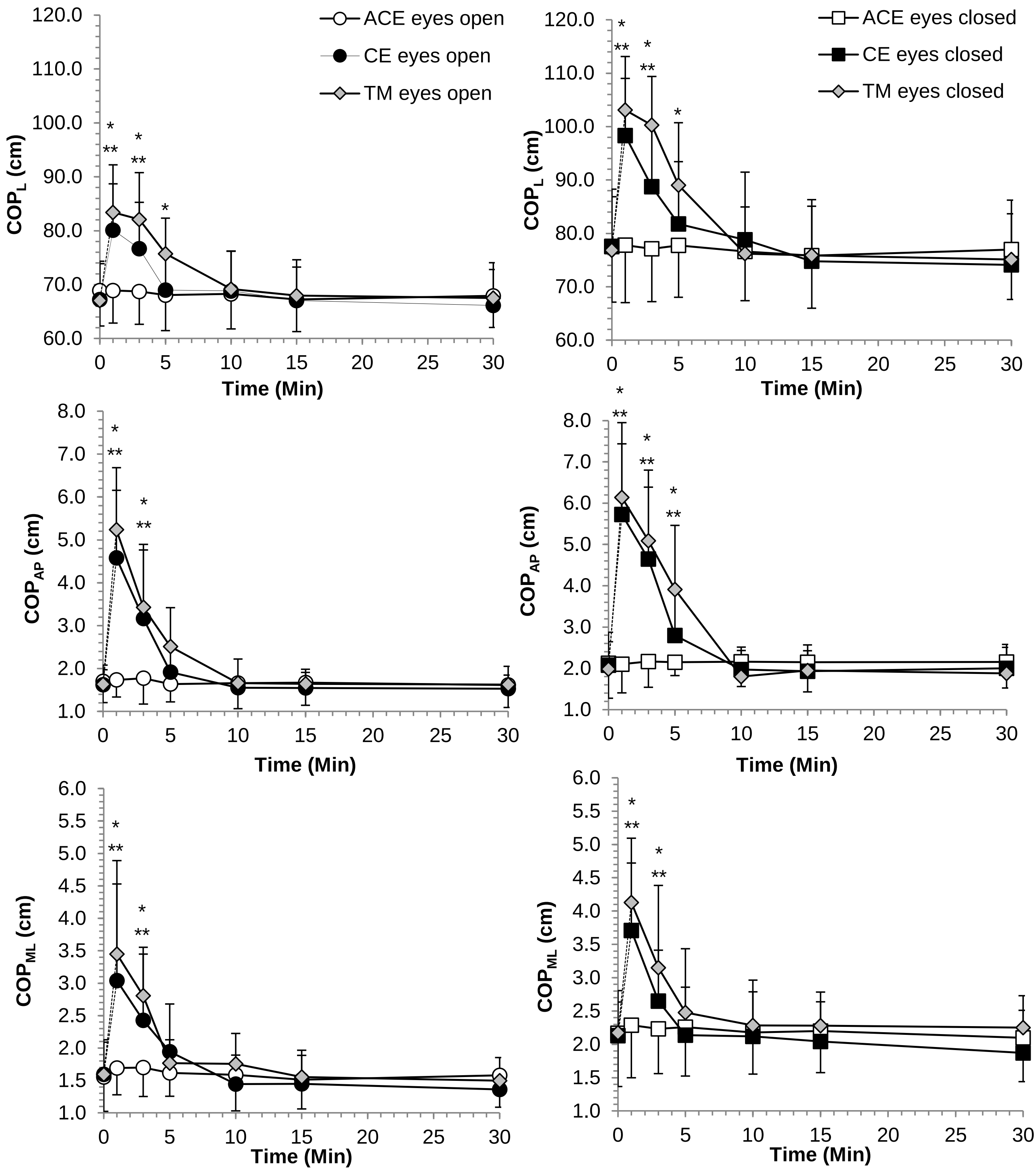


Figure 5.3: Mean ±SD change in COP measures of postural sway from baseline (0) to post submaximal exercise for ACE, CE and TM. * Significantly different to baseline trial for TM. ** Significantly different to baseline trial for CE. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

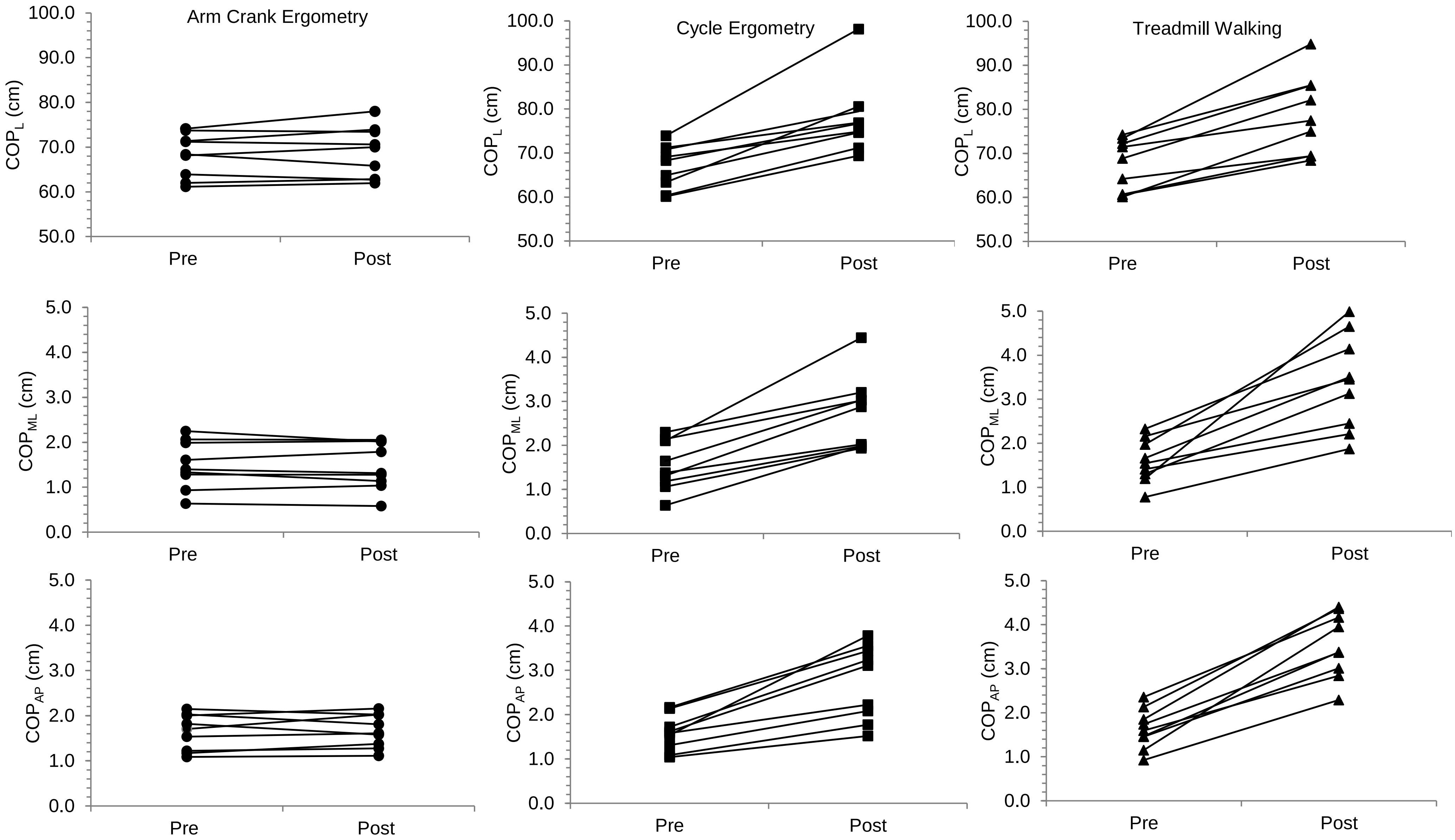


Figure 5.4: Single data plots between pre and immediately post exercise during EO conditions for ACE (left), CE (middle) and TM (right)

5.3.5 Relationship between physiological responses and postural sway

The percentage change in COP_{AP} from baseline (0 min) to immediately after exercise (1 min) was significantly and positively correlated with breathing frequency (Bf) and tidal volume (Vt) at the end of CE (Bf; r = 0.759, *P* = 0.017, Vt; r = 0.819, *P* = 0.001) and TM (Bf; r = 0.701, *P* = 0.035, Vt; r = 0.707, *P* = 0.033) trials. Similarly, there were also significant positive correlations between the percentage changes in COP_{ML} with Bf and Vt following CE (Bf; r = 0.747, *P* = 0.020, Vt; r = 0.936, *P* = 0.002) and TM (Bf; r = 0.805, *P* = 0.008, Vt; r = 0.890, *P* = 0.001) trials. In contrast, no significant correlation as observed between the percentage change in COP_{AP} (Bf; r = 0.188, *P* = 0.626, Vt; r = - 0.125, P = 0.748) or COP_{ML} (Bf; r = -0.124 P = 0.758, Vt; r = -0.292, *P* = 0.444) with respiratory responses. Correlations ae illustrated in Figure 5.5 – 5.7.

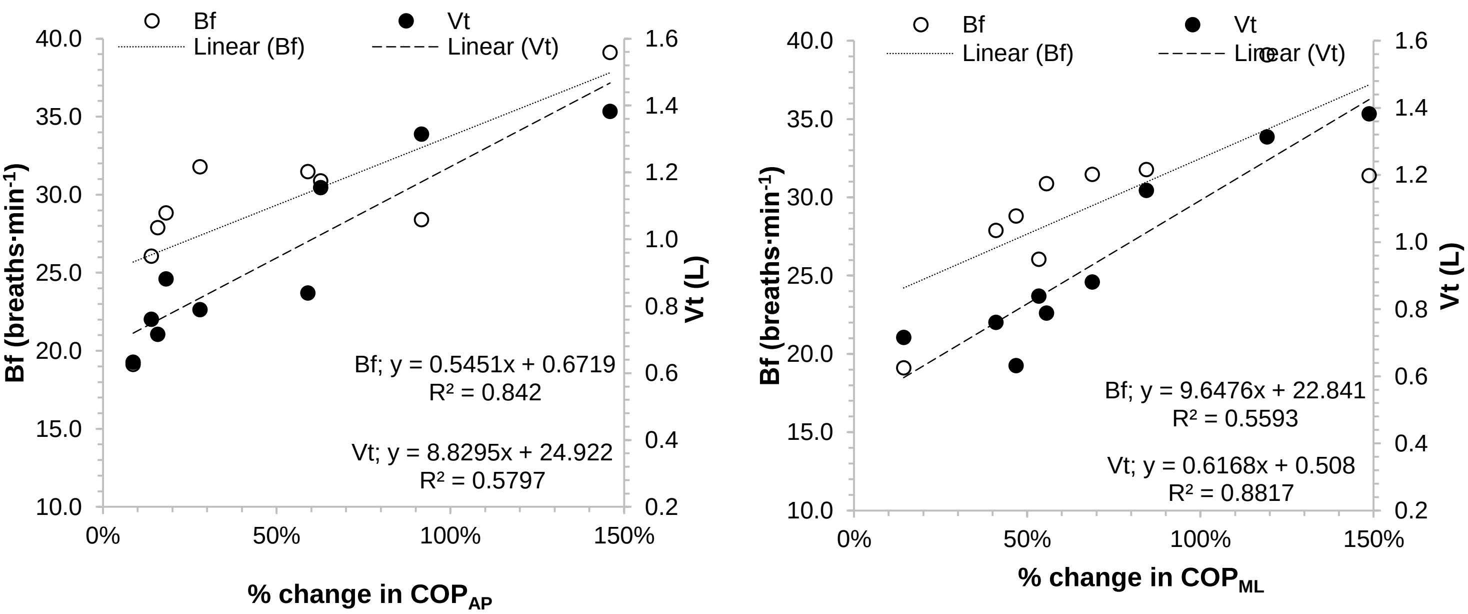


Figure 5.5: The relationship between the percentage change in COP_{AP} (left) and COP_{ML} (right) from baseline (0) to immedatley following exercise (1 min) with respiratory responses averaged in the final 20 s of submaximal CE trials.

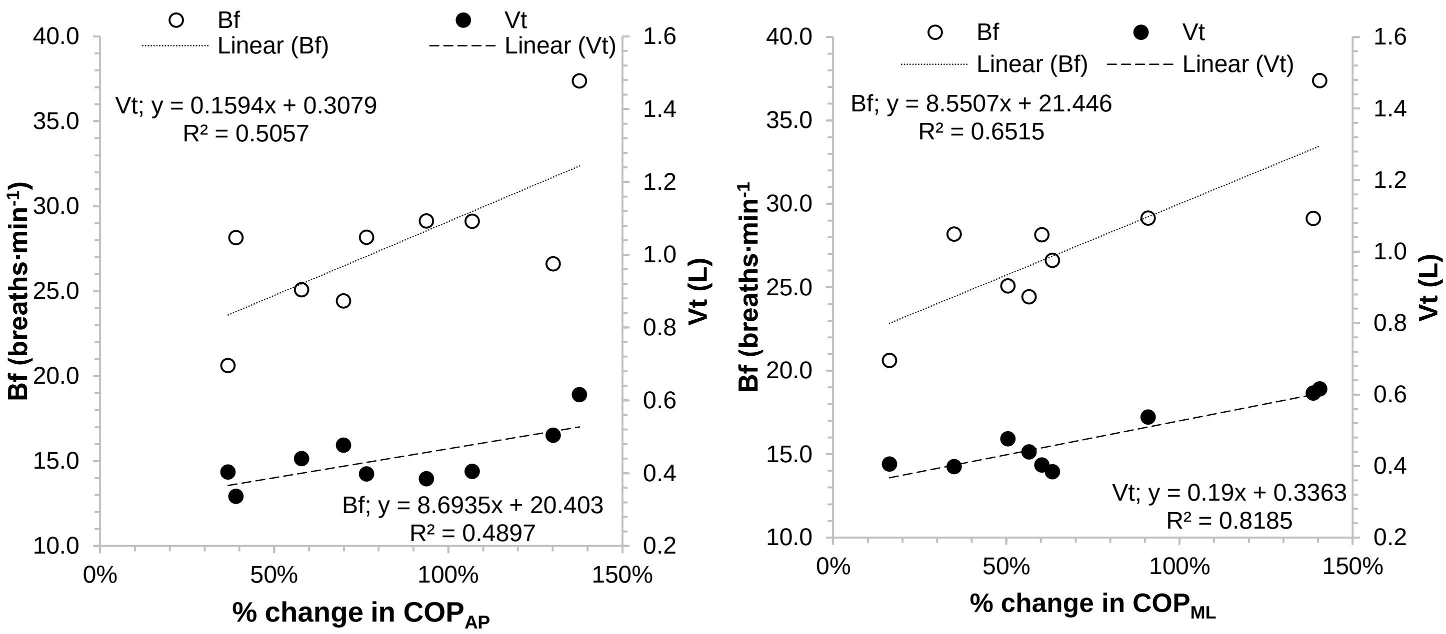


Figure 5.6: The relationship between the percentage change in COP_{AP} (left) and COP_{ML} (right) from baseline (0) to immedatley following exercise (1 min) with respiratory responses averaged in the final 20 s of submaximal TM trials

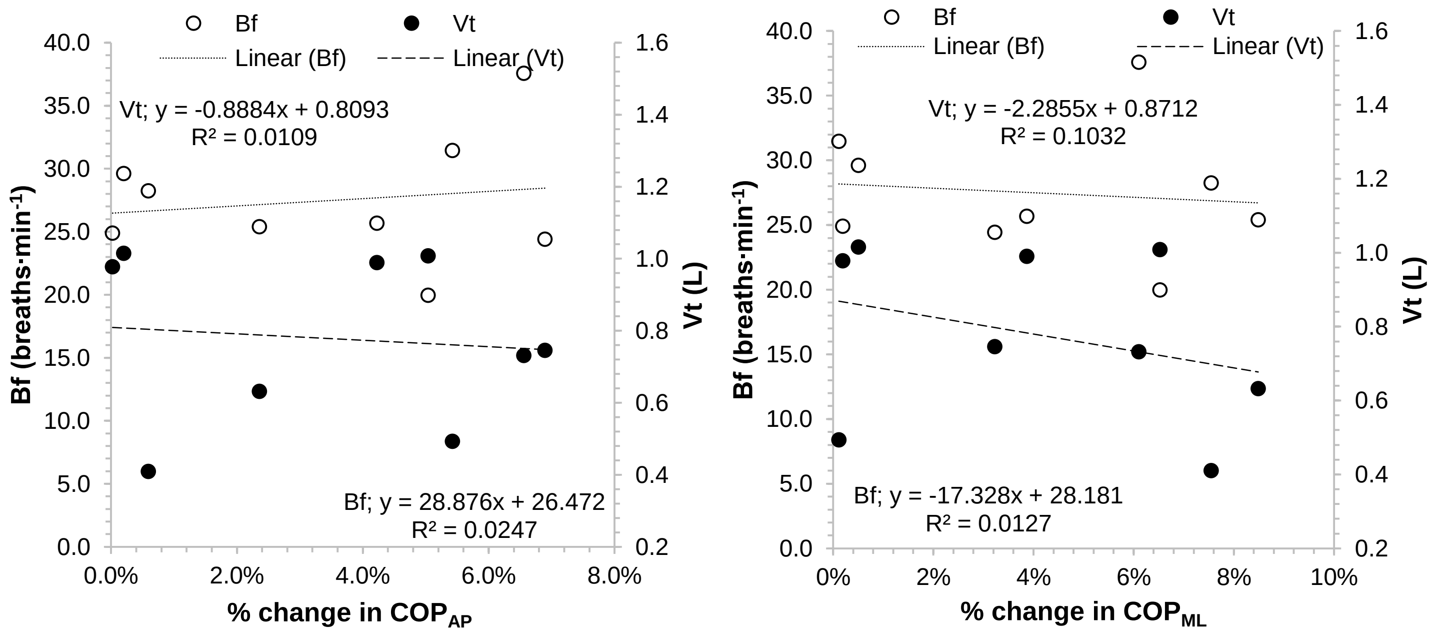


Figure 5.7: The relationship between the percentage change in COP_{AP} (left) and COP_{ML} (right) from baseline (0) to immedatley following exercise (1 min) with respiratory responses averaged in the final 20 s of submaximal ACE trials. Note difference in scale with CE and TM

5.3.6 Comparison between old and young adults

5.3.6 i Arm Crank Ergometry

Baseline COP_{AP} and COP_{ML} displacement were significantly greater in older compared to younger adults for both EO and EC conditions ($P \leq 0.05$) (Figure 5.7). The greatest difference between age groups was observed for COP_{ML} displacement with EC ($P = 0.001$).

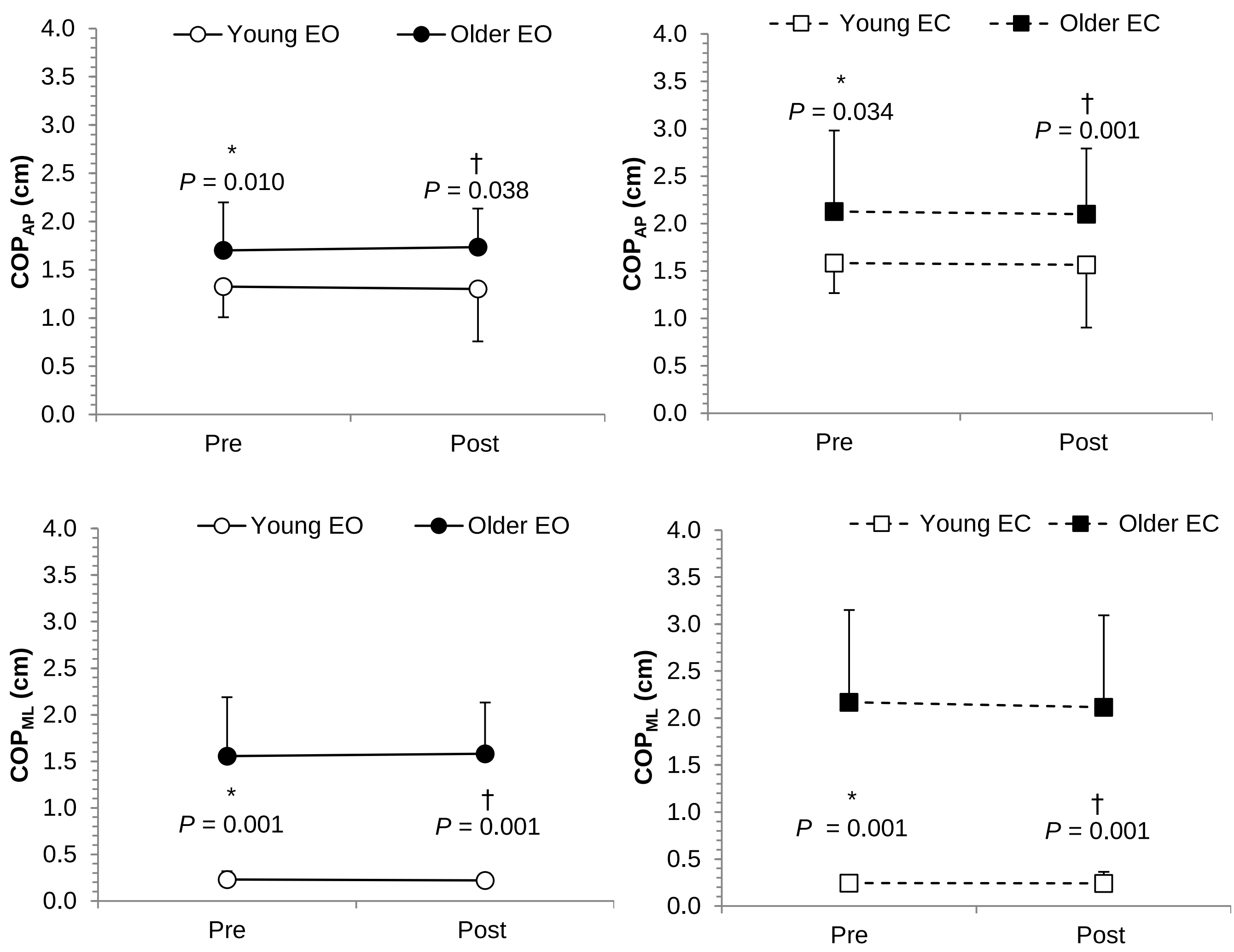


Figure 5.8: Comparisons of the effects of ACE on postural sway in young (30 min, 50 % W_{MAX}) and older (20 min, 50 % HR_E) adults. * $P < 0.05$ between young and old pre exercise. † $P < 0.05$ between young and old post exercise. Post exercise refer to the first measurement recorded (1-min)

5.3.6 ii Cycle Ergometry

Baseline sway measured during CE conditions were different between young and older adults ($P \leq 0.05$) (Figure 5.8). An increase in COP_{AP} from pre to post exercise was observed in both age cohorts, but only COP_{ML} in the older group. Older adults showed larger increases in postural sway compared to their younger counterparts.

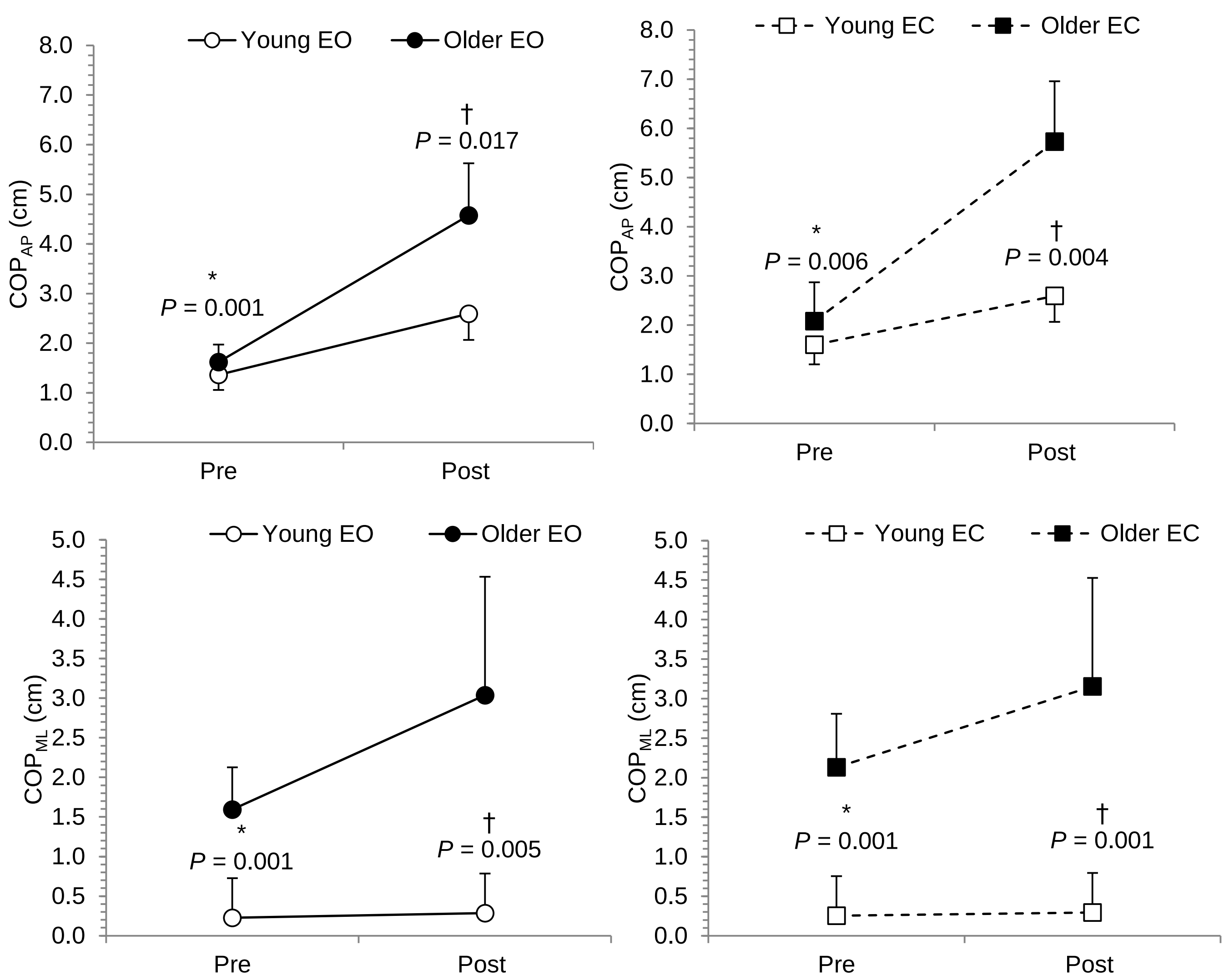


Figure 5.9: Comparisons of the effects of CE on postural sway in young (30 min, 50 % W_{MAX}) and older (20 min, 50 % HR_E) adults. * P < 0.05 between young and old pre exercise. † P < 0.05 between young and old post exercise. Post exercise refer to the first measurement recorded (1-min)

5.4 Discussion

Given the contradictory effects of ACE on postural sway recently cited in the literature (Douris et al. 2011; Hill et al. 2014; Smith et al. 2010) the present study sought to clarify whether arm exercise affected postural stability to the same extent as conventional walking and cycling in older female adults. The present findings show that ACE does not directly disturb postural sway during quiet bipedal standing, which corroborates previous findings in healthy young adults (Hill et al. 2014; Chapter 4). In contrast, postural sway was greater following CE and TM, with the latter appearing to elicit a greater destabilising effect on postural sway. A key finding was an increase in COP_{ML} following CE and TM which might suggest a temporary increase in fall risk post exercise (Egerton et al. 2009; Piirtola and Era 2006). Moreover, this study is the first to report the time course effects of exercise on balance impairment in older adults. Older adults should exercise caution for ~ 10 min following CE or TM, where postural sway values are elevated compared to baseline.

5.4.1 Physiological responses to incremental exercise

Mean $\dot{V}O_2$, HR and power outputs were generally lower than values reported in the literature. For example, Pogliaghi et al. (2006) reported $\dot{V}O_{2MAX}$ values for ACE and CE of 22 $ml \cdot min^{-1} \cdot kg^{-1}$ and 31 $ml \cdot min^{-1} \cdot kg^{-1}$ respectively, in healthy older males (67 ± 5 years). The greater values reported by Pogliaghi et al. (2006) are due to their cohort exercising to volitional exhaustion and being composed of males. The gender difference in $\dot{V}O_{2MAX}$ can largely be explained by differences in body composition (i.e., body fat greater in females) and heart size (larger in males) (Hutchinson, Cureton, Outz & Wilson, 1991).

Oxygen uptake was significantly different between each mode of exercise. The $\dot{V}O_2$ attained upon reaching 85 % HR_{MAX} was proportional to the recruited muscle mass and is in accordance with $\dot{V}O_{2MAX}$ values in young healthy adults (Davis et al., 1976). In contrast, peak HR was the same for CE and TM (~ 130 beats·min^{-1}) but significantly lower for ACE (112 beats·min^{-1}). This is not surprising since HR_{MAX} prediction was adjusted by minus 20 beats·min^{-1} for ACE. The same HR achieved during CE and TM was expected due to the same prediction equation being used (Tanaka 2001). Furthermore, RER was similar between all three exercise modes. As a result, exercise intensity was well matched in terms of the fuel metabolised. Of further interest is the attainment of RER values approaching 1.1, which is widely used as a marker of exhaustion in older adults. As such, while not volitionally exhausted, this data suggests that the older adults were close maximal exercise intensity. Peak RPE_L and RPE_C were not different between exercise modes.

5.4.2 Physiological responses to submaximal exercise

This is the first study to report physiological responses to submaximal ACE, CE and TM in healthy older adults. The present study used an exercise duration of 20-min at 50% HR_E which complies with physical activity recommendations for inactive older adults (Nelson et al. 2007) and is consistent with the only studies that have examined the acute effects of lower body exercise of postural sway in healthy older adults (Egerton, Brauer and Cresswell 2009; Maciaszek, Stemplewski and Osinski 2011; Stemplewski et al. 2012; 2013). With the exception of $\dot{V}O_2$, all physiological variables and RPE showed similar responses between exercise modes, suggesting that trials were closely matched. As for oxygen uptake during ACE, CE and TM was generally proportional to the recruited muscle mass. Furthermore, respiratory responses were also similar between each mode of exercise. As a result, the ventilatory effects of exercise on postural sway were controlled for (e.g., differences between exercise modes on postural sway were not due to differences in ventilatiory responses).

5.4.3 Postural sway following cycling and treadmill walking

Our results confirm previous investigations reporting increased COP displacement after cycling (Maciaszek, Stemplewski & Osinski 2010; Stemplewski et al. 2012; 2013) and treadmill exercise (Donath et al. 2013) in older adults. In general, the percentage changes in postural sway were greater following TM than CE (Table 5.3). Lower body exercise which involves muscles directly involved in balance control deteriorates the quality of sensory proprioceptive information and/or integration (Paillard 2012) and also decreases muscular system efficiency and force generating capacity (Nardone et al. 1997). Walking, which induces eccentric muscle damage of the calf musculature (Vissing et al. 2008) likely disrupts proprioceptive information to a greater extent thaN cycling, which mainly induces concentric contractions of the same muscles (Nardone et al. 1997). Moreover, it has previously been shown that treadmill walking is likely to elicit greater vertical displacements and linear accelerations of the head than cycling (Derave et al. 2001, 2002). Horizontal accelerations of the head induced by walking decreases the sensitivity and increases the detection threshold of the otholitic organs which affects the integration of vestibular information (Lepers et al. 1997; Paillard 2012). Walking exercise activates the otholitic organs thus disrupting vestibular afferents, while accelerations of the head induced by cycling are not significant enough to stimulate them (Paillard 2012). Overall, walking probably creates greater mechanical constraints at the level of active muscle, tendons and cutaneous receptors than cycling does (Lepers et al. 1997; Paillard et al. 2012). The consequences of changes in sensorimotor control may result in improper corrective actions resulting in increased postural sway.

Comparative changes in postural sway reveal similar percentage changes in COP measures previously reported (Donath et al. 2013; Egerton et al. 2009; Stemplewski et al. 2012; 2013). For example, Stemplewski et al. (2012) reported an increase in mean velocity of the COP of ~34 % following 10-min of cycling at 60% HR_E from ~ 1.2 $cm \cdot sec^{-1}$ 9~36 cm COP_L) to ~ 1.6 $cm \cdot sec^{-1}$ (~48 cm COP_L). In a more recent study, Stemplewski et al. (2013) reported an increase from ~ 1.4 cm·sec^{-1} (~42 cm COP_L) to 2.2 cm·sec^{-1} (~66 cm COP_L) using the same protocols as the former study. In the present study, COP variables increased by ~ 10 – 85 % depending on the component of the COP and the visual condition. Similarly, Donath et al. (2013) reported a ~ 13 – 15 % increase in the COP_L before (~ 64 cm) and after (~75 cm) a 2-km treadmill walk (76 ± 8 % $\dot{V}O_{2MAX}$). Although the absolute postural sway values are different to those reported by Stemplewski and colleagues, presumably due to differences in protocols, the relative changes in the motion of the COP displacement are consistent across the present study and those reported by others (Donath et al. 2013; Stemplewski et al. 2012 Stemplewski et al. 2013).

According to Donath et al. (2013) changes of less than 20 % in postural sway following exercise appear to be of minor practical relevance. In the present study the effects of CE and TM on COP_L were not greater than 20 %. In contrast, CE and TM both elicited significant increases in COP_{AP} (CE; 20 – 50 %, TM; 30 – 85 %) and COP_{ML} (CE; 22 – 85 %, TM; 50 – 120 %). It is currently difficult to ascertain whether these percentage changes in balance are pertinent to fall risk. However, the magnitude of sway after exercise up to 4 cm for CE and 6 cm for TM suggest that the centre of mass was moving close to the limits of the base of support and therefore might be clinically relevant to falls.

The increase in COP_{ML} has previously been interpreted as temporary increase in the risk of falling (Egerton et al. 2009), based upon prospective studies showing a link between fall risk and mediolateral balance deficits (Piirtola and Era 2006). An explanation for the increased mediolateral sway in the present study after lower body exercise could be associated with the postural control strategy. For example, during quiet standing, the ankle strategy predominates (Horak and Nashner 1986). However, when lower body muscles are fatigued postural hip and trunk movements dominate the control of balance (Horak and Nashner 1986). This is important since mediolateral balance is dependent on the musculature of the hip and trunk (Winter et al. 1996).

Recently, Bove et al. (2007) reported that the COP_L following running was positively correlated with the quick recovery of oxygen uptake, suggesting that the increased cardiac and respiratory muscular contractions causes an internal disturbance to postural sway. In the present study respiratory responses (Bf and Vt) were significantly correlated with the increase in both COP_{ML} and COP_{AP} sway following CE and TM. In contrast, no significant relationships were observed for the ACE trial. This suggests that increased breath frequency and depth are likely to have produced some of the disturbance to postural sway after CE and TM however, the lack of change in sway after ACE suggests that other factors may have contributed to the increase in sway after lower body exercise.

5.4.3 i Gender effects

The present study provides further evidence that postural sway is also adversely affected in elderly females. In the present study, a ~ 12 % increase in COP path length was observed following CE for 20-min at 50 % HR_E (117 beats·min^{-1}). In contrast, Stemplewski et al. (2012) reported an increase of ~ 34 % COP path length following 10-min cycling at 60% HR_E (118 beats·min^{-1}). While these protocols were not the same intensity or duration, both studies report a similar final heart rate. Therefore, the present findings findings provide some support of a gender effect of exercise on postural sway. Indeed, previous findings have noted an apparent gender difference in muscle fatigability, in favour of females, attributed to the lower absolute force produced performed at the same relative workload as males (Pincivero and Gandaio 2003). This may be explained by the presumption that females possess a greater capacity for utilising oxidative metabolism, therefore reducing the reliance of glycolytic pathways (Russ et al. 2005).

5.4.3 ii Time course of sway disturbance following CE and TM

Further information is provided in the present study with regard to time course of exercise effects on postural sway. In general, all postural sway measures returned to baseline levels within 10 minutes of exercise completion for EO and EC visual conditions. Bove et al. (2007) reported a linear relationship between oxygen uptake and COP_L following running, suggesting that the short lasting effects of exercise on postural sway (~6 min) is related to the recovery of oxygen uptake. The acute effects of localised ankle plantar and dorsi flexor fatigue last up to between 1 – 20 min young adults (Harkins et al. 2005; Lin et al. 2009; Yaggie and McGregor 2002). However, recovery from fatigue is substantially quicker (2 min) in older adults (Lin et al. 2009). In these studies, it is likely that other leg musculature were able to compensate for local fatigue of the ankles (Harkins et al. 2005; Lin et al. 2009; Yaggie and McGregor 2002) while the cycling and walking protocols in the present study involve several large muscles groups of the legs, suggesting compensation may not have been possible in the 10 min following exercise. The post exercise disturbances in balance after CE and TM are consistent with those reported in young healthy adults (Hill et al. 2004; Chapter 4). The transient disturbance to standing balance after lower body exercise in the present might suggest a window where the effects of cycling and walking on balance are most exacerbated and pose the greatest risk for falling or injury in elderly females.

5.4.4 Postural sway responses to arm crank ergometry

Postural sway was not affected by ACE when performed at the same intensity as leg exercise, confirming our previous findings in young adults (Hill et al. 2014; Chapter 4). In the present study, any changes in postural sway following ACE were within the normal measurement error / daily variation expected during quiet standing (Section 3.9.5). While exercise was performed at the same relative intensity, ACE had no effects on postural sway. It is possible that other lower extremity postural muscles that were not fatigued (i.e., triceps surae) were able to compensate for fatigue of the upper extremity musculature and increased respiratory demand to maintain postural stability (Kanekar et al. 2008; Case study, Chapter 4). The non-significant changes in postural sway suggest that older females are able to compensate for the effects of ACE just as well as younger adults (Hill et al. 2014; Chapter 4). Such a favourable response might suggest that older individuals can exercise the upper body safely without increasing the risk of falling. It is also possible that participants did not achieve a level of fatigue which is detrimental to balance.

It remains possible that fatigue of the upper body musculature during other activities (Davidson, Madigan and Nussbaum 2004; Nussbaum 2003; Schieppati, Nardone and Schmid 2003) may disturb postural stability because afferent input from the proprioceptive system forms a chain from the eyes to the feet (Roll and Roll 1988). Furthermore, increased muscle activity of the trunk musculature elicited by ACE may increase trunk stiffness which might reduce the contribution of the trunk to control balance (Smith et al. 2010) especially when postural adjustments are primarily made at the hip, for example during single limb stance (Douris et al. 2011). Additionally, increased muscle activity of the trunk musculature elicited by ACE may increase trunk stiffness which might reduce the contribution of the trunk to control balance (Smith et al. 2010). Therefore, ACE may impair balance when postural adjustments are primarily made at the hip, for example during single limb stance (Douris et al. 2013). Single limb stance was not adopted in the present study due to concerns that older adults would not be able to complete the required task and limiting the ability to compare data with previous findings which have adopted bipedal stance (e.g., Stemplewski et al. 2012; 2013).

Smith et al., (2010) examined the alterations in postural sway following ACE by comparing responses between healthy older adults and those with chronic obstructive pulmonary disease (COPD) patients. The study reported disturbances to balance in the COPD group but not in healthy controls. While the authors did not elucidate their findings in healthy controls (as this was not the novel aspect of the study) their data provide initial indications that the effects of

ACE on postural sway in healthy older adults may be minimal. Arm crank ergometry decreases the ventilatory contribution of some of the inspiratory muscles of the rib cage as they have to contribute in non-ventilatory functions (e.g., upper torso and arm positioning) and there is a shift of ventilatory work to abdominal muscles and the diaphragm (Celli et al. 1988; Couser, Martinez and Celli 1992). Therefore, ACE may alter the contribution of the trunk musculature to maintain balance and respiration, thus comprising the ability to minimise sway. Ultimately, increased activity of superficial abdominal muscles increases trunk stiffness and is likely to reduce the contribution of the trunk musculature movements for postural control (Douris et al. 2011; Smith et al., 2010).

5.4.5 Comparison between young and older adults

Surprisingly few studies have investigated the effects of exercise in relation to age on postural sway (Bisson et al., 2014; Parreira et al., 2013). It might be expected that the impact of muscle fatigue on postural sway may be more pronounced in older adults than in young adults because older adults have less efficient proprioceptive and neuromuscular systems than younger adults (Bisson et al., 2014). Both cohorts in this thesis showed an increase in COP sway following lower body exercise, which is consistent with numerous studies in young (Gouchard et al., 2002; Mello, de Oliveira, & Nadal, 2010) and elderly (Stemplewski et al. 2012; 2013) individuals. When comparing sway responses to CE in younger adults in Chapter 4 (30 min, 50 % W_{MAX}) and older adults in the present study (20 min, 50 % HR_E) the percentage changes from pre to post exercise were dependent on the sway measure. For example, COP path length increased more in young adults (~ 40 – 60 %) compared to older adults in the present study (~10 – 12 %) following CE. This might be due to a much smaller baseline COP path length (e.g., 20 cm vs 70 cm in young and old, respectively) and therefore greater room for disturbance in younger adults. Mediolateral COP displacement increased more following CE in older compared to younger adults. In young adults, the effects of CE_{REL} on COP_{ML} displacement were small (14 – 22 %) compared to relatively greater changes in older adults

(52 – 85 %). In contrast, COP_{AP} displacement showed greater relative changes in young adults (~98 %) compared to older adults (~ 50 %) following submaximal CE.

While relative changes may have been different for COP_{AP} and COP_{ML} displacement, the absolute increase in all sway variables was greater in older compared to younger adults (Figure 5.5). According to Bellew et al. (2009), older adults use compensatory strategies as effectively as younger adults to maintain stability after fatiguing exercise. While, young adults preferentially adopt an ankle strategy during quiet standing (Horak and Nashner 1986), older adults tend to adopt a hip strategy (Woollacott et al. 1986), which may explain the different changes in COP_{AP} and COP_{ML} displacement after cycling in young and older adults. Therefore, it seems that older females have a greater difficulty in maintaining postural stability compared with young males following submaximal CE.

5.4.6 Summary

In conclusion, quiet standing balance is markedly affected following moderate intensity cycling and walking, and remains for approximately 5 - 10 min after exercise in older females. In contrast, upper body exercise performed at the same relative intensity as CE and TM did not elicit post exercise balance impairments. This work contributes to a better understanding of fall risk following different exercise regimens in the older population. In particular, seated exercise using the arms may offer a novel approach to exercise training in older populations without acutely disturbing balance.This thesis has thus far determined the effects of exercise effects on postural sway in a healthy young male and elderly female cohorts. However, it is acknowledged that to thoroughly assess all aspects of balance more dynamic tests are required. It would also be of interest to determine whether there is a training adaptations with repeated sessions of ACE. The present study only evaluate a single ACE session. It might be reasonable to assume that repeated upper body exercise may lead to adapted responses of balance performance.

Chapter 6

A Comparison of Conventional Postural sway, Balance and Functional Balance Tests in Relation to Age

6.1 Introduction

Retrospective studies have reported that almost half of all adults over the age of 65 years suffer a fall at least once per year (Goodwin et al. 2014). Physiological factors that have been shown to be strongly associated with fall risk include increased postural sway (Piirtola and Era 2006), reduced dynamic stability (Gribble et al. 2012), reduced walking speed and impaired lower body and trunk strength (Granata and Lockhart 2008). There is now increasing evidence from meta-analyses for the effectiveness of exercise interventions to mitigate some of these aging processes associated with fall risk (Goodwin et al. 2014). While aging is broadly considered to reduce postural control, the outcome measures to assess the effects of interventions such as training on postural control vary from study to study. Furthermore, while a range of valid and reliable balance measures are currently in use, there is limited evidence regarding the most appropriate measure to assess any changes in performance in older adults (Pardasaney et al. 2011). This lack of standardisation for the measurement of postural control reduces the ability to compare data between different interventions (Du Pasquier et al. 2003).

Selection of a balance measure should be based on how well the specific training intervention matches the purpose of the balance assessment (Steffen, Hacker and Mollinger 2002) and should ideally be dependent on the appropriateness and practicality for the target population (Pardasaney et al. 2011). For example, some studies have reported improved static and/or dynamic balance following endurance training (Paillard et al., 2004; Rissel et al. 2013) while others report reduced fall risk in the absence of changes in balance (Buchner et al. 1997). For example, Buchner et al. (1997) reported a retrospective reduction in fall risk without any improvements in single limb stance time or gait speeds following cycle training. Buchner and colleagues suggested that exercise might affect fall risk by mechanisms which were not ascertained in their study (i.e., quantitative posturography or indirect measures such as functional reach) (Howe et al. 2008). For any balance assessment tool to be a useful outcome measure after any intervention it must be sensitive to change in the older population. Therefore, it is essential to determine which posturographic and functional measures of balance are able to distinguish postural control impairment which is attributable to aging.

Training interventions in the elderly have adopted direct (e.g., quantitative posturography) (Buchner et al. 1997; Ramsbottom et al. 2004), indirect (e.g., functional reach, timed gait, single leg stance time) (Rissel et al. 2013) and observational (Berg Balance Scale [BBS]) measures of balance. The ability to maintain stability and complete functional activities is multifaceted involving not only balance control but also other factors such as strength and proprioception (Howe et al. 2008). Some previous investigations have sought to explore the relationship between functional measures of performance (i.e., functional reach distance, timed gait) with measures of postural sway to establish the clinical relevance of postural sway measures among elderly adults (Hughes et al. 1995; Ringsberg et al. 1999). These studies reported that COP measures of postural sway demonstrate weak associations with other measures at the functional level (e.g., gait speed, functional reach, muscular strength etc). As postural sway and functional tests might not furnish the same information regarding balance mechanisms, measuring postural stability arguably requires a multidisciplinary approach.

The first two studies within this thesis (Chapters 4 and 5) have focused on quantitative assessment of posturography during quiet standing. However, most falls among older adults are reported during ambulation or dynamic movements (Shumway-Cook and Woollacott 2011) and standing as still as possible might not translate to movements during physical activity (Gribble et al. 2012). The Star Excursion Balance Test (SEBT) is a dynamic single limb balance test which has been used in recreationally active young (Olmsted et al. 2002) and middle aged adults (Boullion and Baker 2011). The goal of this test is to establish a stable base of support on the stance limb while performing a maximal reach with the other limb in a prescribed direction (Gribble 2003). A greater reach distance indicates better dynamic postural control. While this test has been advocated for use with training interventions (Gribble et al. 2012) no study has determined the usefulness of this measure in older adults. Indeed, in a recent review, Gribble et al. (2012) highlighted the need to collect normative data for elderly adults to determine whether the SEBT can establish appropriate ‘risk-threshold’ reaching distances for fall risk. This test may overcome the low ceiling effects in other clinical balance assessments such as the Timed up and Go Test.

A classical test of dynamic balance used by many physical therapists is the functional reach test (FRT) (Duncan et al. 1990). The FRT measures the distance an individual can reach in the forward direction from a comfortable standing posture without a loss of balance (Duncan et al. 1992). The Functional Reach Test (FR) was validated and later modified to include Multi-Directional Functional Reach (MDFR) (Newton 2001). Many activities of daily living require individuals to performance dynamic standing balance while reaching is performed (e.g., opening a door, picking up an object from the floor or reaching for an item of food from the cupboard). While the forward functional reach distance is not related to postural sway (Hughes et al. 1996) it remains unknown whether reach in lateral and posterior directions (i.e., MDFR) are related to postural sway. The assessment of lateral and posterior balance is important as older adults typically fall laterally or backwards (Cummings and Nevitt 2004). Since increased mediolateral postural sway is a strong predictor of future falls (Maki 1994; Piirtola and Era 2006; Stel et al. 2003) examining the relationship between mediolateral COP displacement during quiet standing and lateral reach direction in more functional tests might help validate a measure for more clinical use.

While predictive of falls, results obtained from clinical assessment of balance (e.g., Berg Balance Scale, Functional Reach Test, Timed Up and Go Test, etc) can show ceiling effects and might not always be sensitive enough to measure small magnitude of progression (Blum and Korner-Bitensky 2008). Conversely, postural sway assessment is helpful when determining any underlying causes of balance deficit in order to treat it effectively (Horak 1997). Posturography overcomes the subjective nature of the scoring systems and sensitivity to small changes associated with functional assessment (Visser et al. 2008). Although quiet standing is considered a relatively easy task, several manipulations can be introduced to provide more of a postural challenge. For example, reducing the size of the base of support (single limb stance), lack of visual feedback (eyes open / closed) and decreasing proprioceptive feedback (standing on a compliant surface) (Mancini and Horak 2011) are often used in the literature.

Postural sway during quiet standing is robust enough to detect changes in stability (Doyle et al. 2007) however, standing lacks functional relevance as this task rarely occurs in isolation but is rather integrated within activities of daily living (Van Emmerik and Van Wegan 2002). Therefore, there is a need to improve the practical relevance of postural sway assessment. For example, carrying bags during upright tasks is important for many recreational activities (i.e., shopping, socialising etc) (Rugelj and Sevsek 2011). Since shopping for one's own groceries is considered an essential daily activity it is important to investigate the effects of aging on postural sway when undertaking such tasks. While the effects of carrying loads on the back have been addressed (Rugelj and Sevsek 2011) the author is not aware of any data which reports postural sway when bags are held in the hands. Furthermore, standing on a compliant surface has been shown to be more sensitive to balance problems among older adults (Horak 1989). Standing on foam increases reliance of the trunk musculature to maintain balance in both anterioposterior and mediolateral directions (Smith et al. 2010). This might be a useful test to determine the efficacy of upper limb training given the potential contribution of the trunk during arm cranking (Pendergast, Cerretelli and Rennie 1979). To date, only two studies (Era et al. 2006; Illing et al. 2010) have examined postural control under more challenging surface and visual conditions, such as standing on a compliant surface or balance on one leg across decades of ageing, but demonstrating inconsistent findings. For example, in a large cohort (n = 7,979) Era et al. (2006) showed that centre of pressure (COP) measures of postural sway (e.g., path length and displacements) during quiet standing show a systematic and significant increase from 30 – 39 years up to 80+ years of age (Era et al. 2006). However, in a relatively smaller sample size (n = 106), Illing et al. (2010) reported that postural control was not deteriorated until the 60s or 70s. Further evidence is needed across decades of both males and females which may reveal when reduced postural control first emerges, therefore indicating when intervention strategies need to be implemented (Illing et al. 2010).

The purpose of this study was therefore twofold. Firstly, to determine whether postural sway could predict functional performance in a range of tests (i.e., SEBT and MDFR). Secondly, to determine the posturographic and functional balance measures that would yield the best discrimination between age groups. The data obtained will enable determination of tests which may be suitable for inclusion in the assessment of the effectiveness of a training intervention (Study 4) and aid in determining the effects of exercise training on fall risk. This study will also allow us to provide norm reference data for such studies.

Research Hypothesis (H_3): Postural sway will show weak associations with walking performance, dynamic balance and functional abilities. Outcome measures will get proportionally worse with advancing age.

Null Hypothesis ($H0_3$): Postural sway will not correlate with any measures of functional ability or walking performance and no age related differences in outcome measures will be observed.

6.2 Methods

6.2.1 Participants

Participation was open to University staff, students and members of the public. The study population comprised 60 healthy adults between the ages of 22 – 85 years of age (n = 25 Males, n = 35 Females) (Table 6.1). The cohort was divided into six, 10-year age groups (Era et al. 2006) (20 – 29 years, 30 – 39 years, 40 – 49 years, 50 – 59 years, 60 – 69 years and 70 - 85 years). There was a 92% compliance, with a total of five adults withdrawing themselves from the study. Each participant provided details of past medical history, surgical history (if any), prescribed medications, fall history, activity levels and limb dominance as these variables may influence balance (Lord, Menz and Tiedemann 2003).

Table 6.1: General participant characteristics

Decade	Sample	Age (Years) (Range)	Height (cm)	Mass (Kg)	Gender (M/F)
20-29	13	25.5 ± 2.3 (22 – 28)[1]	170.6 ± 8.7	72.1 ± 15.1	5 / 7
30-39	7	33.8 ± 2.7 (31 – 39)[2]	173.8 ± 4.1	73.2 ± 6.7	4 / 3
40-49	11	43.4 ± 2.9 (40-49)[3]	172.4 ± 7.4	72.7 ± 13.0	5 / 6
50-59	9	53.0±2.0 (51 – 56)[4]	170.3 ± 7.6 [4]	79.4 ± 16.2 [4]	5 / 4
60-69	9	63.8±1.9 (61 – 67)[5]	163.1 ± 7.5 [5]	71.3 ± 22.3 [5]	4 / 5
70+	11	74.7±4.3 (70 – 84)	152.6 ± 9.4	80.9 ± 28.5	5 / 6

[1] Significant difference with 30-39; [2] Significant difference with 40-49; [3] Significant difference with 50-59; [4] Significant difference with 60-69; [5] Significant difference with $\geq$70.

6.2.2 Experimental Procedure

Each participant visited the laboratory on three separate occasions separated by a minimum of 24 hours and maximum of 72 hours (Gil et al. 2011). Balance assessments were administered as follows; session one: postural sway (fixed and compliant surface, holding loads and single limb stance); session two dynamic balance (functional reach test and star excursion balance test); session three: walking tests (timed gait and gait ground reaction forces). The order of tests was counterbalanced both within and between sessions.

6.2.3 Postural sway

Section 3.9.3 details the specific sampling procedures for the assessment of postural sway. All participants completed postural sway tests under the following conditions; (a) standing in a bipedal position on a fixed surface, (b) standing in a single limb position (both limbs) on a fixed surface, (c) standing in a bipedal position on a fixed surface while holding loads and (d) standing in a bipedal position on a compliant surface. Specific procedures for these tests are detailed in 3.9.4.

6.2.4 Walking Tests

Each participant was instructed to walk at either a comfortable or fast gait speed on rubber tiled laboratory flooring. Gait velocity was recorded using two photoelectric timing gates (SmartSpeed, Fusion Sports, Australia) at a distance of 8 m apart (Montero-Odasso et al., 2005). The height of each measuring device was standardised at 0.5 m for all trials to avoid the timing gaits being incorrectly set off by swinging arms. Two portable force platforms (AMTI, AccuGait, Watertown, MA) embedded in the laboratory floor were used to record ground reaction forces of each foot during a single gait cycle during the comfortable walking speed trials only. Consecutive force platform strikes of the right and left foot were subsequently acquired. To ensure valid data acquisition each participant’s starting position was adjusted until the right foot contacted the platform first followed by the left foot without any visible or self-reported alteration in normal gait. All participants were asked to remove footwear. In the case of the participant missing the force platform, partially or completely, or if both feet come into contact with the same platform, the trial was discarded. If trials were repeatedly unsuccessful participants were instructed to start the gait initiation from a different location along the walkway. This procedure was repeated until three valid trials were recorded. Ground reaction force data were recorded at 200 Hz using real time data acquisition software package (AMTI, Netforce®, Watertown, MA) and subsequently analysed using the accompanying gait analysis software package (AMTI, BioAnalysis, Version 2.2, Watertown, MA). Data were filtered with a 4^{th} order low-pass Butterworth filter with a cut off frequency of 6 Hz. Analysis of COP data lead to the computation of main sway parameters; vertical maximum deceleration, vertical maximum acceleration, COP excursion along anteroposterior and mediolateral axis, COP path length, maximum and average velocity of the COP of each foot strike. Ground reaction force data were only recorded for comfortable gait speed trials due to difficulties in recording a clean foot strike during fast walking speeds. Spatio-temporal data included single limb stance duration, double limb stance duration, gait velocity, cadence (subjectively counting the number of steps taken) and stride length (distance / cadence). It is acknowledged the latter two variables are susceptible to less precision.

6.2.5 Star Excursion Balance Test (SEBT)

The SEBT was performed with each participant standing barefoot at the centre of a grid marked out on the laboratory floor using highly visual adhesive tape. The first two lines formed the horizontal and vertical axes, and a further two lines were positioned perpendicular to each other at 45° increments from the centre of the grid (Figure 6.1). Participants were asked to place their metatarsophalangeal joints on the mediolateral line so that the centre of the foot was aligned with the intersection of the anterioposterior and mediolateral lines (Figure 6.2). A crosshair was positioned at the centre of the grid for visual aid. Participants maintained a single leg stance while reaching with the contralateral limb to touch as far as possible along each of the eight lines. Each participant was instructed to make a light touch contact with the ground with the great toe of the reach leg and to return to a double leg stance. Following familiarisation (one practice attempt in each direction), participants performed three reaches in each direction. Each trial began with the anterior direction and progressed clockwise and counter-clockwise for the left and right leg, respectively. All participants began with a left leg stance. In accordance with the test protocol, maximal excursion distance was measured visually and to within a centimetre. The investigator was positioned along each directional line and a dry wipe marker was used to mark the excursion distance. The trial was completed when the participant returned to the starting position. After each trial, participants were given sufficient time to recover in the standing position before starting the next trial. A standard metal tape measure was used to quantify excursion distance in each direction from the centre point of the grid to a point of maximum reach distance by the contralateral leg using the most distal part of the foot. If the investigator felt that the participant used the reach leg to provide substantial mechanical support or did not return to a sufficient double leg stance position the trial was discarded and repeated. Data presented were normalised for each individual's height (Gribble et al. 2003).

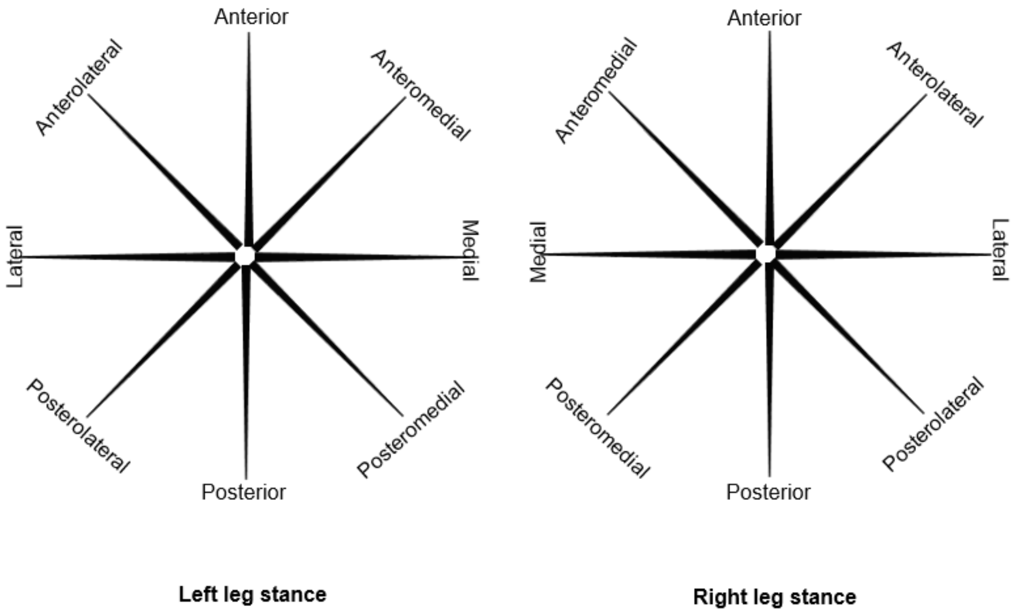


Figure 6.1: Reaching directions for the star excursion balance test

6.2.6 Multi-Directional Functional Reach Test (MDFR)

Participants stood barefoot with feet shoulder width apart and parallel to each other. Particpants were asked to flex the shoulder of the dominant arm, with the elbow joint fully extended. Instructions were similar to those of Duncan et al. (1990) where particpants were instructed to reach as far as possible without taking a step and ensuring heels remained in contact with the ground.

Figure 6.2: Performance of the Star Excursion Blance Test with left leg stance leg reaching in the posterolateral direction

A meter stick was mounted horizontally to the wall at the same height of the acromion. Reach distance was measured as the displacement of the most distal part of the hand from the initial to the end position. In accordance with Duncan et al. (1990) the reaching strategy was not controlled for. In addition to the standard forward reach proposed by Duncan et al. (1990) this test also incorporated posterior reach and both right and left lateral reaches. The same instructions and criteria were consistent for each test direction. For clarity, data are presented as anteroposterior (total forward and backward reach combined) and mediolateral (total right and left reach combined) distance.

6.2.7 Grip Strength

Grip strength was measured using a standard adjustable hand dynamometer (Lafayette Instrument Company, USA) during upright standing. The shoulder was flexed to 180°, the elbow was fully extended and the wrist was pronated (Robertson et al. 2011). The following standardised instructions were verbally communicated to each participant 'I want you to hold the handle like this and squeeze as hard as you can'. The principal investigator demonstrates and then gives the dynamometer to the participant. After the participant is positioned appropriately, the investigator says, 'Are you ready? Squeeze as hard as you can'. As the participant begins to squeeze, the examiner says, 'Harder! Harder! Relax' (Mathiowetz, Rennells and Donahoe 1985). A total of three measurements were recorded for the dominant and non-dominant hand, with the best measurement recorded for analysis.

6.2.8 Statistical analysis

Differences between all variables with respect to the six age groups were assessed using a one-way ANOVA. In the case of statistical significance, Scheffe's post-hoc analysis was undertaken to calculate the difference required between means. Data was analysed using SPSS (IBM v17 and 20, Chicago, USA). Statistical significance was accepted with an alpha level of P = 0.05. Data were initially tested for normality (Shapiro-Wilk Test) and homogeneity of variance (Levenes Test). Pearson's correlation coefficient was used to estimate the direction and strength of the relationships between all variables. Coefficients were calculated using a correlation matrix in Microsoft Excel ®.

6.3 Results

6.3.1 Age related increases in postural sway

Results for bipedal standing with eyes open (EO) and eyes closed (EC) when standing on a fixed and compliant surface are shown in Figure 6.3. The results indicate a significant effect of age related increases in the amount (COP_L) and size (COP_{AP} and COP_{ML} displacement) of sway during both EO and EC conditions when standing on compliant and fixed surface (all P = 0.001). Scheffe's post-hoc comparisons indicated no differences between young (20 – 29, 30 – 39 years) and middle-aged (40 – 49, 50 – 59 years) groups for all sway measures for both compliant and fixed surfaces (all $P \geq 0.05$). Young and middle aged groups showed significantly less COP_L, COP_{AP} and COP_{ML} than 60 – 69 years ($P = 0.001$) and 70+ years ($P = 0.001$) for all sway measures, under both visual conditions and surfaces. Comparisons between 60 – 69 years and 70+ year groups were not found to be statistically significant between visual or surface conditions. The COP_{AP} and COP_{ML} were approximately double when standing on foam with EO compared to on a fixed surface with EO. However, when the eyes were closed these variables increased approximately six fold compared to standing on a fixed surface. The ability to minimise sway with eyes closed was more adversely affected when standing on foam as demonstrated by larger differences in sway measures between visual conditions. Figure 6.4 contextualises the differences in postural sway between age groups with illustration of the COP migration. No gender differences were observed for any postural sway variables when standing on a fixed or compliant surface (all $P \geq 0.05$).

In the single limb stance test, only COP_L showed a significant effect of age on both right and left limb stance (both $P = 0.001$) (Figure 6.5). COP_L was only different between young (20 – 29 years) and oldest (70+ years) cohorts ($P = 0.001$). COP_{AP} and COP_{ML} were not found to be significant between any age groups ($P \geq 0.05$). It is noteworthy that COP_{AP} and COP_{ML} for single limb stance were approximately four times greater than bipedal standing with eyes open (all $P \leq 0.05$). When holding a load with both hands, as with single limb stance, an effect of age was observed only for COP_L for both hands ($P = 0.006$), the left hand ($P = 0.049$) and the right hand ($P = 0.001$) (Figure 6.6). Scheffe's post-hoc comparisons indicated that the youngest group (20 – 29 years) showed significantly shorter COP_L compared to the oldest group (70+ years) ($P \leq 0.05$). For COP_{AP} and COP_{ML} while holding a load, no differences were observed between any of the age groups ($P \geq 0.05$).

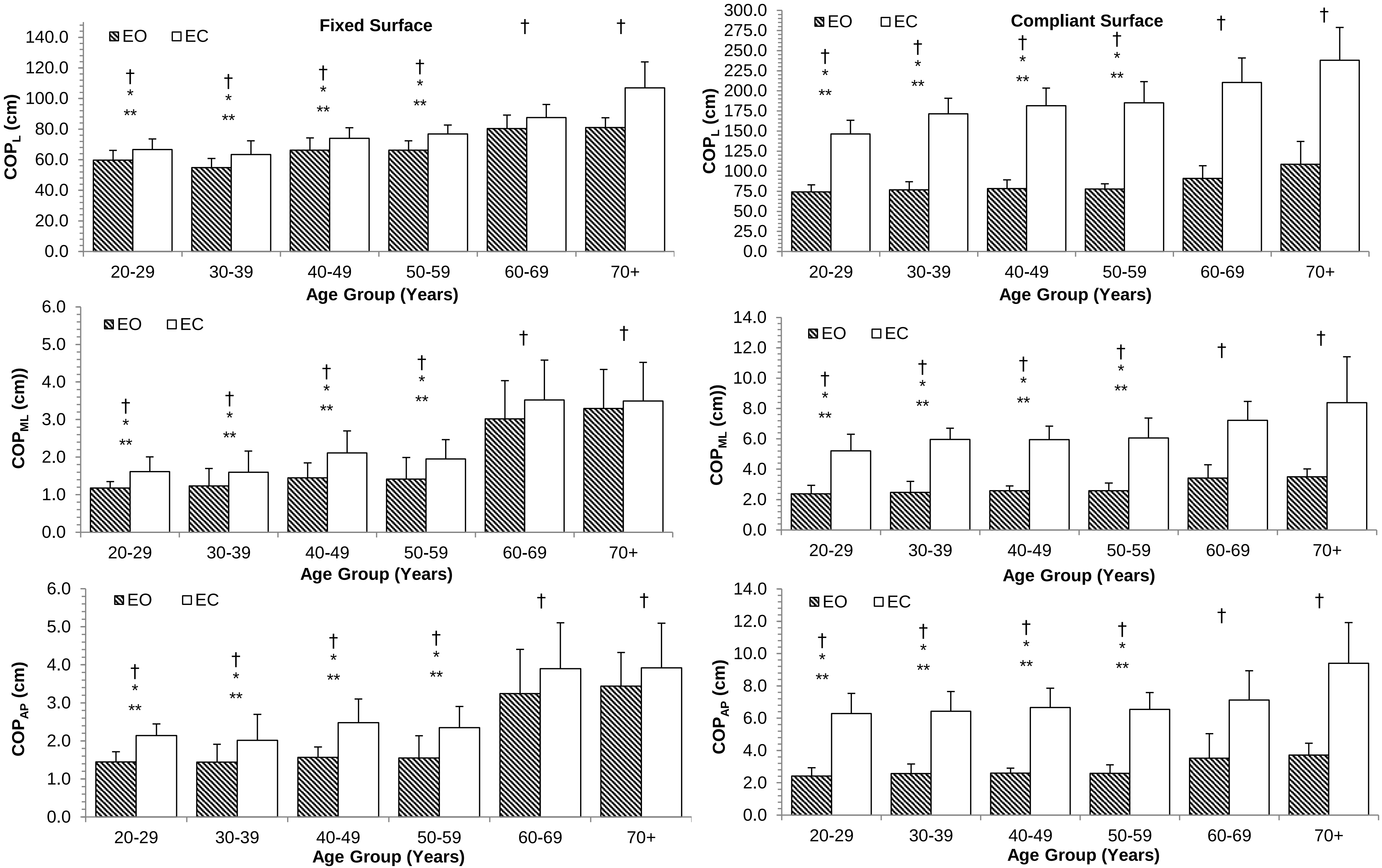


Figure 6.3: Mean ± SD for COP path length and mediolateral and anteroposterior COP displacement when standing on a fixed (left) and compliant (right) surface. † Signifciant between visual conditions (*P* ≤ 0.05). * Signifciant with 60 – 69 years (*P* ≤ 0.05). ** Signifciant with 70 – 85 years (*P* ≤ 0.05).

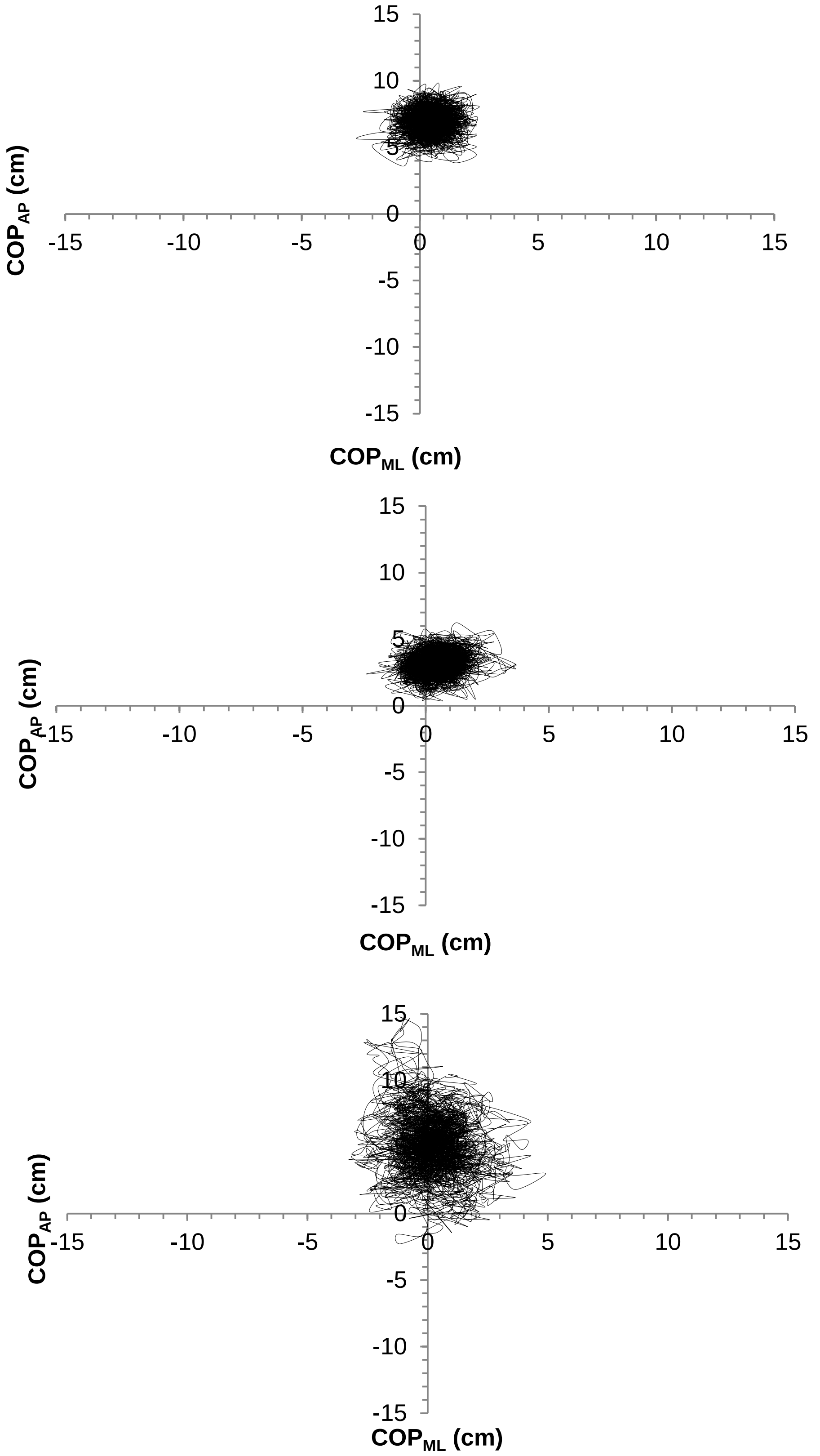


Figure 6.4: Representative figure illustrating the migration of the COP_{AP} and COP_{ML} during bipedal stance with EO in a 20-29 year (top) 40 – 49 year (middle) and $\geq$ 70 year (bottom) adult

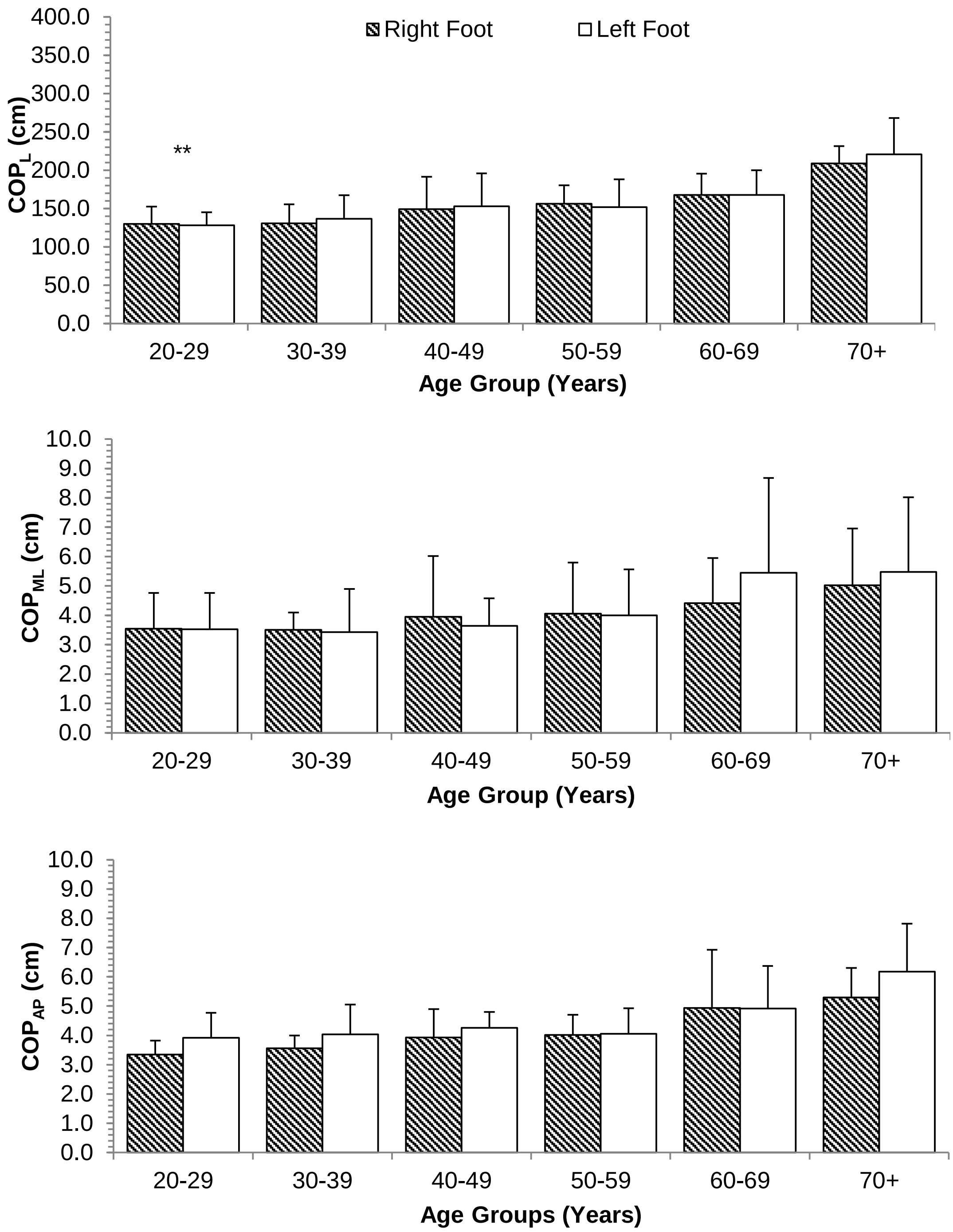


Figure 6.5: Mean ± SD for COP_L, COP_{ML} and COP_{AP} when standing in a single limb stance on a fixed surface. ** Significant with 70 – 85 years ($P \leq 0.05$).

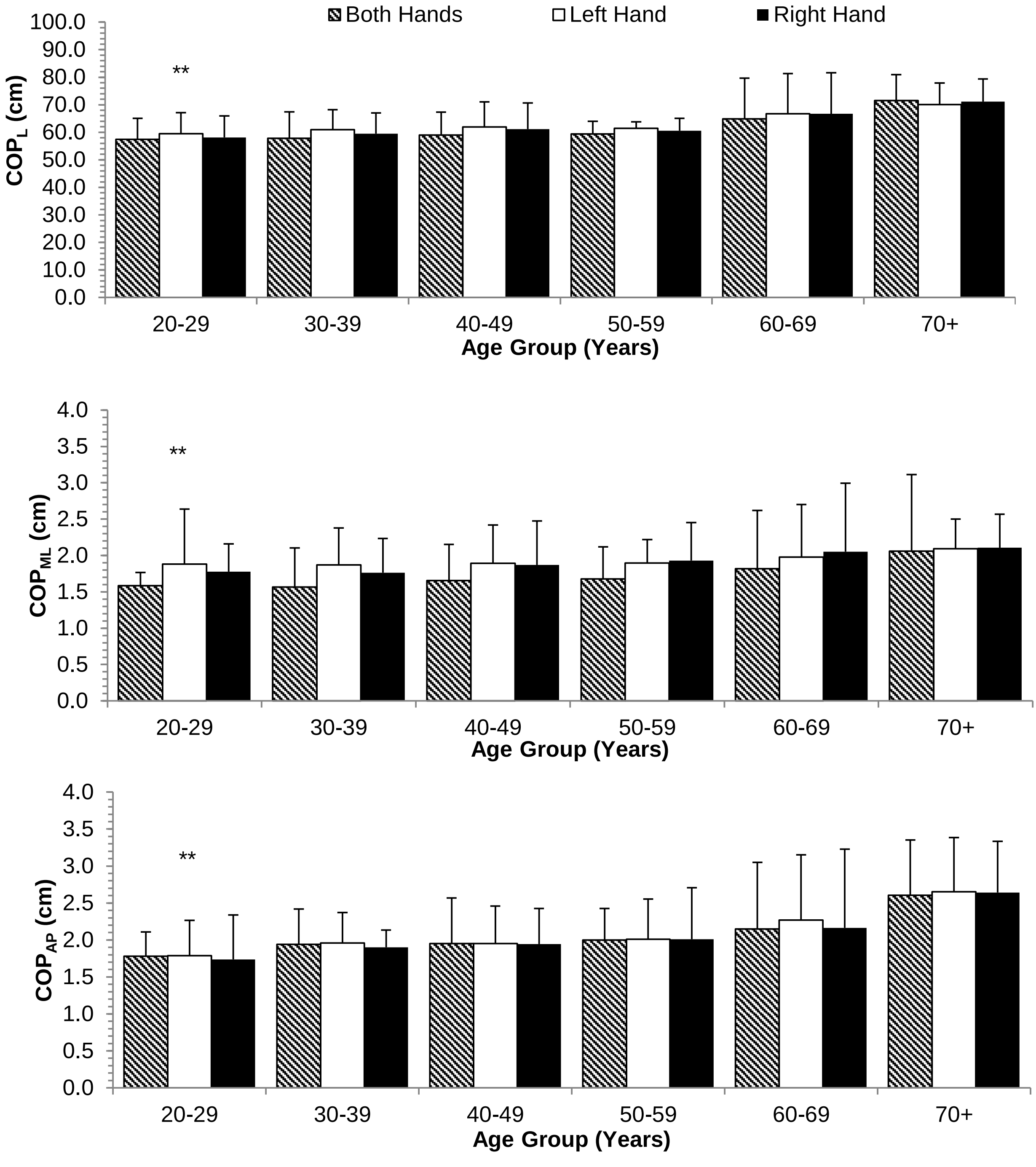


Figure 6.6: Mean ± SD for COP$_L$, COP$_{ML}$ and COP$_{AP}$ when standing holding a load with a left, right or both hands. ** Significant with 70 – 85 years ($P \leq 0.05$).

Compared to bipedal standing a significant reduction in COP_L was observed when holding a bags symmetrically or asymmetrically (all $P \leq 0.05$) (Figure 6.7). The COP_{AP} and COP_{ML} were not significantly different when holding loads compared to bipedal standing without holding loads.

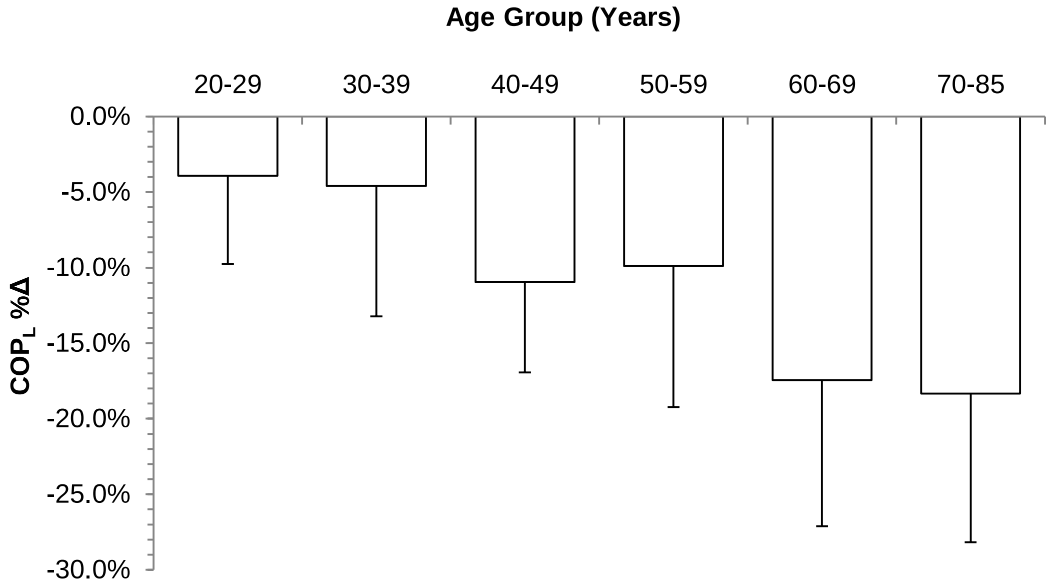


Figure 6.7: Reduction in COP_L when holding symmetrical and asymmetrical loads. Note that percentage change is expressed as the mean change for holding the load with both hands and the right and left hand.

6.3.1 i Relationship between postural sway and age

All postural sway variables showed significant relationships with age (all $P \leq 0.05$). For example, age was strongly correlated with path length EO when standing on a fixed (P = 0.001, r = 0.78) and compliant surface (P = 0.001, r = 0.64). When the eyes were closed there was also a strong relationship between age and path length when standing on a fixed (P = 0.001, r = 0.71) and compliant surface (P = 0.001, r = 0.70) (Figure 6.8).

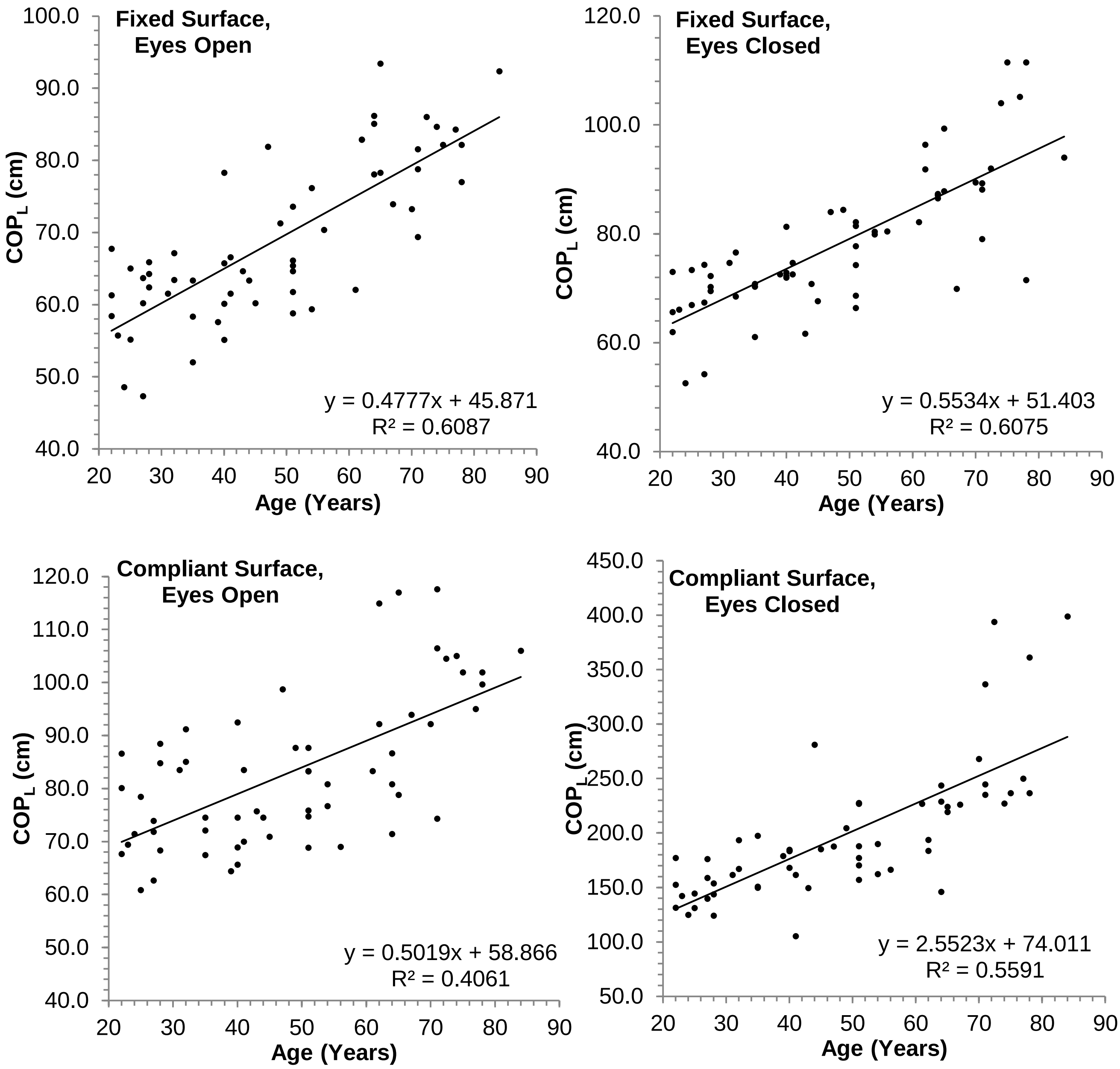


Figure 6.8: Relationship between COP path length when standing on a fixed (top) and foam (bottom) surface with age.

6.3.2 Age related reductions in functional reach distance

An effect of age was observed for forwards, backwards, right and left functional reach distance (all P = 0.001) (Figure 6.9). The distance reached in each of the directions was greatest in the 20 – 29 years ($P \leq 0.01$). A significant reduction in each reach direction was apparent by 50 –

59 years (all P = 0.001). For all directions the trend was a reduction in reach distance at 50-59 with a plateau until 60-69 and a further significant reduction at $\geq$ 70 years (all $P \leq 0.05$). For all age groups, there were no differences between right and left reach distance (all $P \geq$ 0.05). Backward reach distance was significantly poorer than forward reach distance for all age groups (all P = 0.01). For all reach distances there was a reduction of ~ 5 cm from the 40 – 49 group to the 50 – 59 groups, with a further ~ 4 cm reduction between the 60 – 69 group and $\geq$ 70 year group.

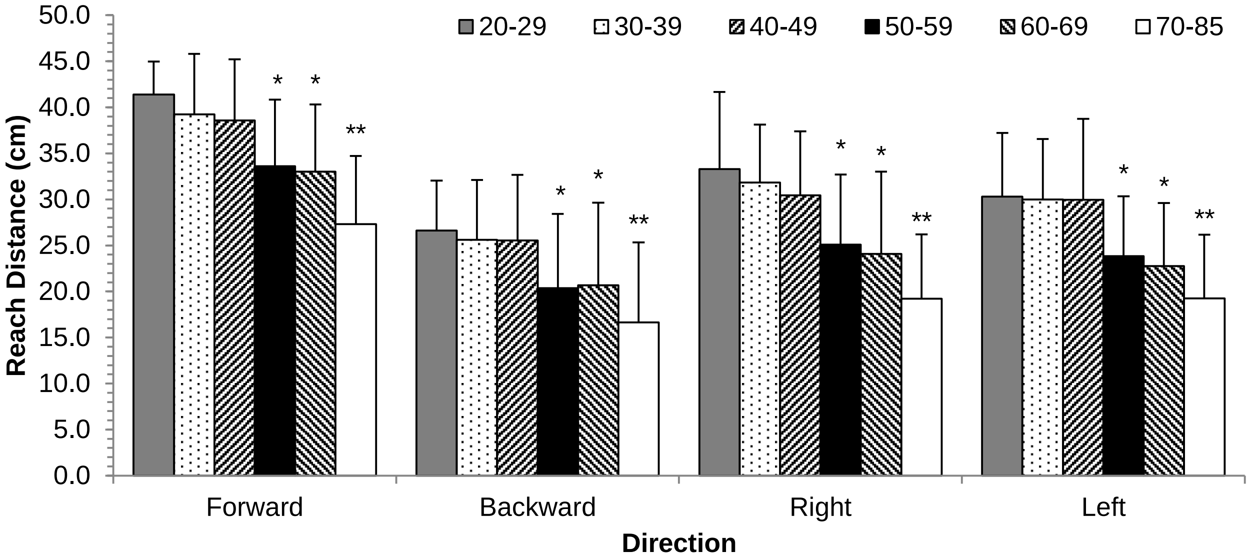


Figure 6.9: Reach distance for each age group. * Significant with $\leq$ 59 years ($P \leq 0.05$). ** Significant with 50 – 59 and 60 - 69 years.

6.3.3 Age related reductions in Star Excursion Balance performance

An effect of age was observed for all SEBT reach directions on both feet (all P = 0.001) (Figure 6.10). For all reach directions the younger cohorts (20 - 29 years and 30 – 39 years) and the 40 – 49 year group demonstrated significantly greater reach distance compared to the 50 + age groups when reaching with both right and left limbs (50 – 59 years, 60 – 69 years and 70+ years all P = 0.01). When standing on the right and left feet lateral and posterolateral reach distances were approximately two fold greater in the younger adults (20 – 29 and 30 – 39 years) compared to oldest cohorts ($P \leq 0.01$) (Figure 6.10)

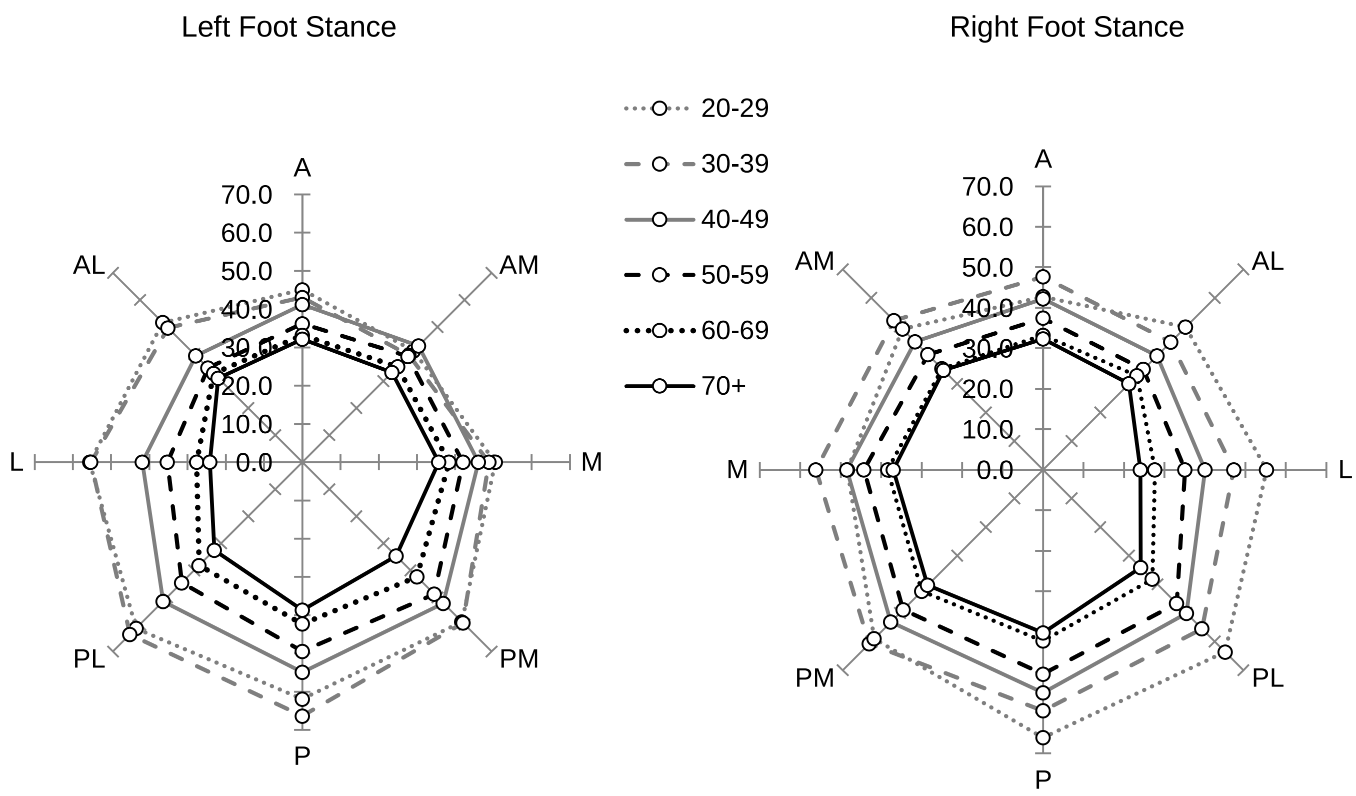


Figure 6.10: Mean data for normalised SEBT for both left foot and right foot stance among each age group

6.3.4 Age related reductions in gait speed

When walking at fast speeds a significant effect of age was observed for gait velocity (P = 0.001) and stride length (P = 0.001). Scheffe's post hoc comparisons indicated that the youngest cohorts (20 – 29 years and 30 – 39 years) walked significantly quicker and exhibited a greater stride length than all cohorts over the age of 50 years ($P \leq 0.05$). The 40 – 49 years group showed greater velocity and stride length than the two oldest cohorts (60 – 69 years and 71+ years) ($P \leq 0.05$) only. No age effects were observed for cadence (P = 0.689). These data suggest that fast gait velocity and stride length systematically decreases with age however, a marked decrease is observed at 50 – 59 years of age. A significant reduction in velocity of ~ 0.40 ms^{-1} was observed after 40 years, with a systematic decrease in speed of ~ 0.10 ms^{-1} thereafter for every decade.

There were no differences between age groups with respect to comfortable walking speed or cadence (Table 6.2) ($P \geq 0.05$). However, there was a significant effect of age on stride length ($P = 0.003$) when walking at a comfortable speed. Scheffe's post hoc comparisons indicate that young and middle age groups demonstrated a significantly greater stride length compared to the 60 – 69 years and 70 + years groups (both $P \leq 0.05$). No main effects of age were observed for ground reaction forces (COP_{AP} and COP_{ML}, maximal or average velocity of the COP and double or single limb stance duration) (all $P \geq 0.05$).

Table 6.2: Gait parameters during comfortable walking speeds among age groups

	Young		**Middle**		**Older**	
	18-29	**30-39**	**40-49**	**50-59**	**60-70**	**70 - 85**
Comfortable						
Cadence (s/min)	117 ± 6	121 ± 9	120 ± 10	116 ± 18	113 ± 14	120 ± 10
Velocity (m/s)	1.24 ± 0.11	1.25 ± 0.12	1.21 ± 0.17	1.19 ± 0.23	1.07 ± 0.18	1.13 ± 0.13
Stride Length (m)	0.63 ± 0.04**	0.61 ± 0.03**	0.61 ± 0.05**	0.61 ± 0.06**	0.56 ± 0.03	0.56 ± 0.06
Fast						
Cadence (s/min)	164 ± 23	166 ± 26	153 ± 23	144 ± 16	147 ± 17	145 ± 18
Velocity (m/s)	2.35 ± 0.25¥	2.34±0.36¥	1.94 ± 0.28¥	1.74 ± 0.34**	1.60 ± 0.20	1.51 ± 0.16
Stride Length (m)	0.87 ± 0.08¥	0.85 ± 0.05¥	0.77 ± 0.06¥	0.72 ± 0.08**	0.65 ± 0.04	0.63 ± 0.05

* Significantly different to 60 – 69 years. ** Significantly different to ≥ 70 years. ¥ Significantly different to 50 – 59 years

6.3.5 Relationship between postural sway and functional performance

6.3.6 i Postural sway and functional performance

None of the COP measures of postural sway measured in normal bipedal standing, standing on foam, single limb stance or bipedal stance while holding bags were correlated with any gait characteristics during comfortable or fast walking speeds ($P \geq 0.05$). Furthermore, none of the COP measures were associated with ground reaction forces recorded during comfortable walking speeds. All associations were typically weak and negatively directed. Interestingly, there was no correlation between any measures of postural sway functional reach distance or star excursion balance test results. Associations were again weak and negatively directed ($P \geq 0.05$).

6.3.6 ii Relationship between MDFR and SEBT

Functional reach distance showed moderate to strong correlations with star excursion balance test performance (all $P \leq 0.05$, r = 0.40 to r = 0.79) (Figure 6.11). Anterior functional reach was strongly related to SEBT in all directions and on both feet ($P \leq 0.05$, r = 0.69 to r = 0.79). Posterior functional reach showed moderate associations with all SEBT reach directions ($P < 0.05$, r = 0.42 to r = 0.54). Functional reach in the left direction was a stronger indicator of SEBT performance in all directions and when standing on both feet ($P < 0.05$, r = 0.63 to r = 71) compared to right functional reach ($P < 0.05$, r = 0.45 to r = 59). Correlations between all measures are reported in Table 6.4.

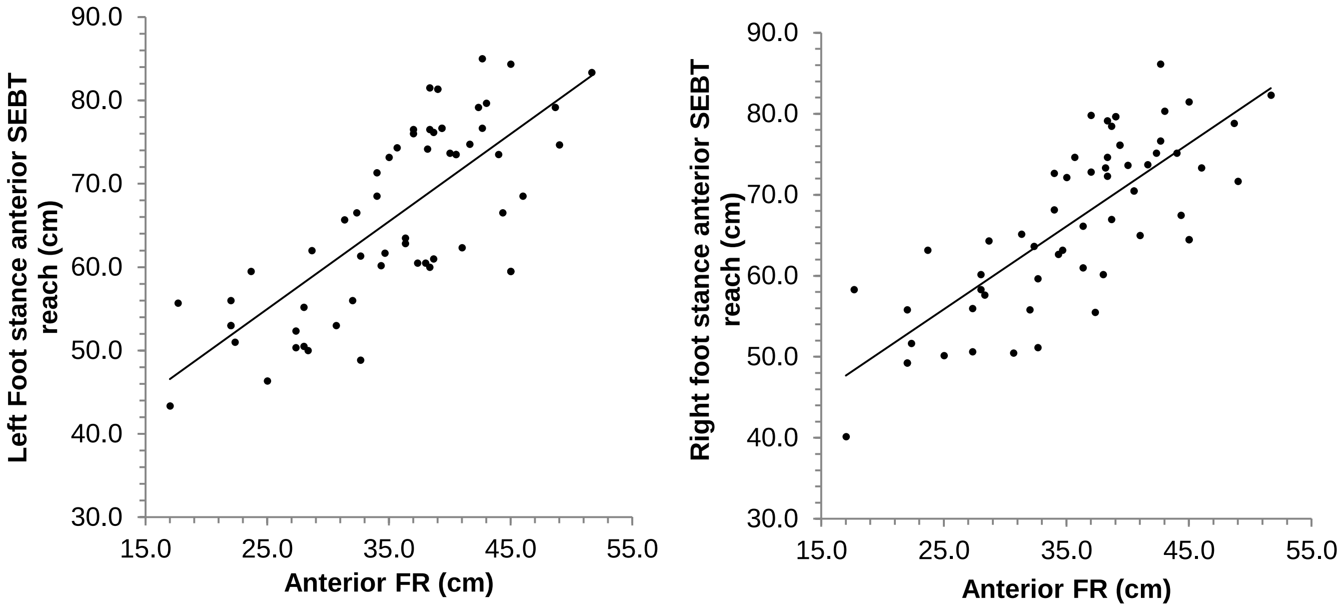


Figure 6.11: Relationship between anterior MDFR and anterior SEBT for left and right legs among all age cohorts

6.3.6 Age related reductions in grip strength

Significant effects of age were observed for right and left hand grip strength (both $P = 0.001$) (Table 6.3). Grip strength was signifcaintly reduced in adults over 60 years of age compared to adults under 59 years (all $P \leq 0.001$). There was a strong negative correlation between age and grip strength with the right ($P = 0.0001$, $r = -0.64$) and left ($P = 0.0001$, $r = -0.65$) hands.

Table 6.3: Mean ± SD for hand grip strength among each age cohort

Grip Strength	**Young**		**Middle**		**Older**	
(Kg)	**20-29**	**30-39**	**40-49**	**50-59**	**60-70**	**70+**
Right Hand	43 ± 14	51 ± 14	43 ± 9	42 ± 12	24 ± 6*	19 ± 3*
Left Hand	39 ± 13	46 ± 9	43 ± 9	39 ± 10	21 ± 6*	16 ± 4*

*Significantly different with < 60 years

Table 6.4: Correlations between postural sway conditions and functional measures of dynamic balance, reach distance, walking speed and gait

		Bipedal Standing Fixed Surface	Bipedal Standing Compliant Surface	Single limb Standing Fixed Surface		Load Carriage Bipedal Standing Fixed Surface		
				Right	Left	Right	Left	Double
Spatio-temporal gait	Stride length	-0.41 – -0.25	-0.53 – -0.22	-0.47 – 0.04	-0.49 – 0.11	-0.31 – 0.19	-0.24 – -0.07	-0.40 – -0.31
	Cadence	-0.19 – 0.04	-0.14 – -0.05	-0.20 – -0.07	-0.20 – 0.09	-0.06 – 0.11	-0.09 – 0.16	-0.11 – 0.08
	Velocity	-0.37 – -0.23	-0.42 – -0.19	-0.47 – -0.02	-0.47 – -0.04	-0.20 – -0.15	-0.22 – 0.01	-0.30 – -0.21
Gait mechanics	Right stance (s)	-0.09 – 0.38	-0.12 – 0.36	-0.09 – 0.28	-0.02 – 0.15	-0.12 – 0.30	-0.13 – 0.29	-0.13 – 0.41
	Left stance (s)	0.25 – 0.06	0.23 – 0.46	0.00 – 0.47	-0.02 – 0.40	0.12 – 0.30	0.06 – 0.29	0.16 – 0.41
	Double stance (s)	-0.18 – 0.06	-0.27 – -0.07	-0.07 – 0.29	-0.11 – 0.22	-0.18 – 0.01	-0.16 – 0.02	-0.09 – 0.04
SEBT	Right stance	-0.51 – -0.13	-0.62 – -0.23	-0.52 – 0.01	-0.51 – 0.01	-0.40 – -0.07	-0.33 – 0.06	-0.52 – -0.17
	Left stance	-0.53 – -0.11	-0.59 – -0.27	-0.53 – 0.04	-0.51 – -0.05	-0.38 – -0.02	-0.28 – 0.11	-0.47 – -0.23
MDFR	Anterior	-0.48 – -0.16	-0.47 – -0.33	-0.43 – -0.11	-0.40 – -0.18	-0.35 – -0.31	-0.27 – -0.16	-0.49 – -0.39
	Posterior	-0.38 – -0.07	-0.33 – -0.25	-0.31 – -0.07	-0.20 – -0.01	-0.28 – -0.05	-0.27 – 0.01	-0.34 – -0.12
	Right	-0.48 – -0.16	-0.44 – -0.18	-0.35 – -0.07	-0.30 – -0.09	-0.32 – -0.17	-0.19 – -0.09	-0.34 – -0.25
	Left	-0.31 – -0.12	-0.52 – -0.11	-0.35 – -0.03	-0.50 – -0.31	-0.25 – -0.11	-0.23 – -0.13	-0.38 – -0.19

6.4 Discussion

The findings from this study support the hypothesis that a decline in postural stability (COP_L, COP_{AP} and COP_{ML}) is present by 60 years of age, concurring with previous studies (Choy, Brauer and Nitz 2002; Illing et al. 2010). Furthermore, to the best of the author's knowledge this study provides novel data with regards to the age related reductions in both functional reach distance in four directions, and star excursion balance performance in eight directions. Specifically, this study provides evidence that earlier changes are observed for functional balance measures than for sway measures, emerging at 50 and 60 years of age, respectively. This study is also the first to report the relationship between dynamic balance tests (e.g., MDFR and SEBT) with more challenging postural tasks (e.g., standing on a compliant surface, single limb and holding a load). The main finding was that postural sway measures are not related with functional performance tests as evidenced by non-significant correlations. The finding that functional performance tests were unable to predict postural sway performance accentuates the need to employ both laboratory measures of balance (i.e., postural sway) and functional measures when determining the efficacy of interventions such as exercise training on balance performance and fall risk.

6.4.1 Age effects

6.4.1 i Postural Sway

The findings of this study support the hypothesis of an increase in postural sway with age during bipedal standing on a fixed and compliant surface as reflected by an increased COP_L, COP_{AP} and COP_{ML}, concurring with previous studies (Black et al., 1982; Era et al., 2006; Illing et al. 2010; Lord and Ward 1994). Specifically, in the current study, fixed and compliant surface conditions (EO and EC) only revealed changes in postural stability in adults aged 60 and over compared with younger and middle aged adults, which is consistent with the findings of Illing et al. (2010), who used similar protocols and age cohorts. The current understanding is that an increase in COP_{AP} and/or COP_{ML} is clinically undesirable, as it indicates that the centre of mass is moving closer to the limits of stability. In contrast, a smaller COP_{AP} and COP_{ML} sway is clinically beneficial which indicates that the centre of mass is moving over a shorter distance and therefore upright stance is more easily maintained, thus reducing the demands on the postural control systems. It could be hypothesised that increased COP_{AP} and COP_{ML} may only be problematic and relevant to fall risk if the COP moves beyond the limits of stability. Therefore, while the magnitude of change in COP_{AP} (~ 1.70 cm) and COP_{ML} (~ 1.60 cm) from 50 – 59 to 60 – 69 years while standing on a fixed surface may not be clinically relevant to fall risk, it is indicative of reduced sensorimotor function and thus poorer balance control. When compared to daily variation data, it is very likely that these increases are related to increased age and are not a result of normal between trial variations.

Most previous studies (Laughton et al. 2006; Maki, Holliday and Topper 1994; Nagy et al. 2007; Prieto et al. 2005) determined age-related reductions in postural control using single postural task assessments and considered only two age groups (i.e., young [20-30 years] and elderly [60-80 years]), thus limiting the generalisability of their findings. The present study provides novel data by assessing postural sway in six distinct groups and using a more diverse range of postural assessments, therefore potentially allowing for a better characterisation of changes that occur during the ageing process.

The present study showed that COP measures of postural sway increased at 60 – 69 years of age, with further reductions in adults aged 70 – 85 years, which has not be detected in previous investigations. Illing et al. (2010) reported that dynamic proprioception, measured by vibration sensitivity which declined at 60 years of age with further reductions observed in adults over 70 years. Era et al. (1986) also reported a significant difference in vibration sensitivity between young (31 – 35 years), middle age (51 – 55 years) and elderly (71 – 75 years). Several other mechanisms have been discussed in the literature to explain reduced postural control with age, such as a loss of receptor cells in the vestibular organs (Serrador et al. 2009), a reduction in peripheral sensation (Era and Heikkinen 1985), changes in visual function (Lord and Menz 2000), increased reaction time (Era et al. 1996) and reduced muscular strength (Lord, Clark and Webster 1991). Overall, bipedal standing on a fixed and compliant surface with eyes open and eyes closed provide definitive age related decline in postural stability. The current finding that such standing tasks can readily reveal changes in postural stability suggest that these tests may be appropriate to explore the benefits of an exercise training intervention in adults over 60 years of age.

In accordance with the findings by Illing et al. (2010), for the single limb stance test changes in postural stability were not revealed until 70 years when standing on either the dominant or non-dominant leg. This differs from other studies, where a reduction in postural stability was observed by 60 – 69 years (Choy, Brauer and Nitz 2003). The finding that the challenge of standing on one leg did not reveal age related impairments in balance may suggest that this test is subject to floor and ceiling effects. For example, Parreira et al. (2013) argue that single limb stance provides a more challenging task than bipedal stance. However, the current study suggests that for some groups, single limb stance may simply be too difficult and may result in floor effects, as evidenced by the limited sensitivity of the test to differentiate balance performance between age groups.

For bipedal standing while holding bags, changes in postural stability were not revealed until 70 years of age. To the best of the author's knowledge no previous data exists which documents the effects of holding bags in the hands during quiet standing. The reason why only the oldest (70 – 85 years) and youngest (20 – 29 years) cohorts were different is an interesting finding in the context that grocery bags are traditionally carried to the side of an individual and thus has the potential to increase postural sway (Sutton et al., 2010). Age related increases in postural sway in the mediolateral direction is correlated to falls in older adults (Maki 1994; Piirtola and Era 2006; Stel et al. 2003). It might be expected that carrying grocery bags either unilaterally or bilaterally may decrease stability in the mediolateral direction. However, no differences in COP_{AP} or COP_{ML} were observed when compared to standing quietly without loads. These findings suggest that holding loads in the hands up to 5 % of individual's body which equates to a relatively small amount of groceries (e.g., typically a bag of sugar and carton of milk) is not a meaningful factor for inducing postural instability. It remains possible that holding heavier bags may have a more profound effect on COP displacements. In contrast, the COP_L decreased systematically with age compared to bipedal standing. This suggests that while the size of sway with and without holding bags was consistent, the amount and velocity of sway reduced (Van Emmerik and Van Wegen 2002). Therefore, holding a load in the hands may reduce the activity required to regulate postural staility (Maki et al., 1990). While the underlying mechanisms for the reduction in COP_L remains unclear, it is possible that the bodies centre of mass is lowered, which will result in an inherently more stable position. The greater relative reduction in COP_L with each age group may suggest lowering the centre of gravity becomes more important as we get older.

6.4.1 ii Functional balance performance

The outcomes of the MDFR test of the present study reinforced the findings of Duncan et al. (1990) and Nolan et al. (2010) by demonstrating an age-related decline in forward, backward and lateral reach distance in older adults. The mean MDFR scores are the same as those reported by Nolan et al. (2010) across a similar age range. The findings of the functional reach test of our study demonstrated that there was a significant reduction in reach distance in all four directions by the fifth decade, with a further trending reduction by the seventh decade, concurring with the findings of Nolan et al. (2010). This study provides novel findings with regards to backwards reach distance, which showed a similar trend to other directions and was equally adept to detecting age related changes in reach distance. The findings from the current study suggest that reaching in all four directions may be an appropriate test to screen for balance impairments in older adults (> 50 years). While no studies have recorded neuromuscular responses to the MDFR test *per se*, volitional reaching tasks are coupled with stabilising postural activity of the leg and trunk muscles (Duncan et al. 1990). The contribution of the trunk and the upper extremity during reaching makes this an ideal test to assess pre and post training, particularly in the context of upper body exercise (Study 4). Pre-emptive exercise training interventions targeting upper limb muscles should consider the inclusion of the MDFR test.

This is the first study which has reported the performance of the SEBT in adults over 40 years of age. Reach distances in the present study were similar to previous studies reporting SEBT performance in young (Gribble and Hertel 2003) and middle aged adults (Bouillon and Baker 2011). This study provides novel data which demonstrates that the SEBT can differentiate balance deficits as early as the 50s. The SEBT has been used for the purpose of predicting lower limb injury (Bressel et al. 2007), chronic ankle instability (Olmsted et al. 2002), assessing deficits in dynamic postural control (Gribble et al. 2004) and differentiating performance between young and middle aged adults (Boullon and Baker 2011). Previous findings reported that healthy older adults show smaller voluntary centre of gravity excursions, reach maximal limits of stability more slowly and demonstrate poorer postural control once they reach maximal reach compared to younger adults (Blaszczyk, Lowe and Hanseb 1994). The finding that SEBT decreased with age suggests that this test discriminates balance impairments in older adults equally as well as other functional (i.e., MDFR) and quantitative (i.e., posturographic) assessments.

The present findings build on prior knowledge by providing promising indications that the SEBT can be successfully used to differentiate balance performance in older adults. In particular, the greatest differences in performance with age were evident in the lateral and posterior lateral directions, where reach distance was approximately half of that reported in the young cohorts (20 – 39 years). Future research should determine whether the SEBT can be used in the prediction of future fall risk.

6.4.1 iii Gait Tests

There is a wide range of comfortable and fast gait speeds reported in the literature for healthy older adults (Bohannon 2008; Nolan et al. 2010). The mean data reported in the present study typically fall in the centre of the reported age ranges within the literature. In agreement with the above studies there were no differences in gait velocity, stride length or cadence during comfortable walking speeds. Average gait speeds for healthy older adults range from 0.60 to 1.45 $m \cdot s^{-1}$ for comfortable walking speeds and from 0.84 – 2.1 $m \cdot s^{-1}$ for fast walking speeds which are comparable with the present study (Bohannon 2008). In contrast, when walking at fast speeds the current study shows significant age effect was observed for both velocity and stride length. The present study shows a systematic decrease in walking velocity from the 20 – 29 yeas to $\geq$ 70 years, concurring with previous findings (Steffan, Hacker and Mollinger 2002). Our older particpants in their 60s and 70s were able to increase there fast walking speed by 48 % and 34%, respectively, beyond a comfortable pace which is similar to those reported by Steffan, Hacker and Mollinger (2002) (38 % and 29% respectively) and Bohannon (1997) (56 % and 37 %, respectively) which tends to suggest our data fall in the centre of previous values. Comfortable gait speed has been recommended as a vital sign for functional performance in older adults and a decrease of 0.1 m/s in walking velocity has been associated with an increased risk of falls in older adults (Van Kan et al. 2009). The present study showed gait velocity was similar between adults aged 20 – 29 to 40 – 49 years (within ~ 0.04 m/s). However, while not statistically significant, the reduction in gait velocity after 50 – 59 years of 0.1 m/s (compared to $\leq$ 50 years) may be practically significant to fall risk among the older adults (60 – 69 years and $\geq$ 70 years).

6.4.2 Relationship between postural sway and functional performance

Postural sway measures recorded using a range of different tasks (e.g., fixed and compliant surface, single limb stance) are more likely to capture age related deficits rather than differentiate functional performance abilities, as evidenced by non-significant correlations between tests. None of the COP measures of postural sway were associated with any measures of functional reach distance, walking speed or star excursion reach distance. Despite differences in protocols, these findings are consistent with previous research (Gil et al. 2011; Hughes et al. 1995; Ringsberg et al. 1999), which reported weak or even no association between postural sway and other functional tests in elderly adults. However, the present study is the first to examine the relationship between postural sway measures and functional balance tests across a wide age spectrum. Previous literature specifies that quantitative posturograpghy and functional assessments may be measuring different aspects of balance control and are not comparable (Gil et al. 2011; Hughes et al. 1995; Ringsberg et al. 1999). These differences are discussed below. The lack of relationship found between functional balance measures and postural sway in the present study is likely explained by the different nature of static and dynamic balance tasks. For example, postural sway assessment involves maintaining an upright stance while standing on a fixed, firm and still base of support while minimising movements of the centre of gravity (Goldie, Bach and Evans 1989). In contrast, dynamic balance requires the maintenance of a stable base of support while completing prescribed movements, often displacing the centre of gravity close to the limits of the base of support (Gribble et al. 2004). Typically, dynamic postural stability tasks pose a greater challenge to musculoskeletal components and anticipatory and compensatory mechanisms for postural control (Sell et al. 2012). Therefore, static and dynamic balance appear to be measuring very different aspects of the postural control system.

While the FRT and SEBT both challenge dynamic balance the activation of muscles required for stabilisation and the strategy used to execute the reaching task while performing these tests is different (Wernick-Robinson, Krebs and Giorgetti 1999). For example, FR involves the trunk and arms moving in a pre-determined direction. In contrast, the SEBT involves the legs reaching in one direction, and the trunk and centre of gravity moving in the opposite direction. The present study shows for the first time that there is a strong relationship between FR and

SEBT, particularly in the anterior direction. This data provides further validation of the use of SEBT as a tool for dynamic balance assessment, particularly in older adults.

6.4.3 Conclusion and Recommendations

This study confirmed a decline in postural stability, dynamic balance and gait with age, which was typically definitive by 50 – 59 years and 60 – 69 years for dynamic and static tests, respectively. This study provides novel data which shows that functional tests get worse after 50 years of age and therefore appear to be better able to discriminate the effects of aging on balance. It is possible that functional tests may provide an earlier warning sign for balance impairment compared to postural sway assessment. The present results also provide normal performance levels to use as goals as well as identify factors associated with reduced balance in older adults in future studies (Chapter 7). In addition, the current findings suggest that postural sway is not a useful indicator of functional performance, but is a useful age discriminating assessment of balance. The different results obtained from posturography assessment and functional performance tests strengthen the evidence that these tests are measuring different aspects of postural control and are not comparable. Postural sway involves detecting disturbances and then responding with timely co-ordinated sensory strategies to stabilise the centre of mass over the base of support (Woollacott 2000). In contrast, functional tests typically measure how far an individual can reach or how quickly a tests can be completed (Podsiadlo and Richardson 1991).

While inherently different, the present data show that with advancing age an overall decline in balance is observed in both anteroposterior and mediolateral direction for both static and dynamic tests. The results of this study support the need for exercise training interventions to target muscles controlling mediolateral movements (e.g., hip abductor and adductor muscles and trunk extensors) and also those controlling anteroposterior plane movements (e.g., knee extensors and flexors and ankle plantar/dorsiflexors) (Choy, Brauer and Ntz 2002). It is expected that dynamic exercise such as arm crank ergometry and cycle ergometry will have different effects on static and dynamic balance and directional control of such tasks. Training interventions might elicit age reversal effects on balance performance as assessed using the abovementioned tests. Future training interventions can use the reference data from sources such as the current study for normative comparisons, in addition to published norm data, to quantify any reversal in fall risk and balance performance relative to age following a given intervention.

Chapter 7

The Effects of 6-weeks Upper and Lower Body Exercise Training on Measures of Postural Sway and Functional Balance Performance in Healthy Older Adults

7.1 Introduction

An increase in postural sway among older adults has been linked to an increased risk of falling (Maki, Hollida and Topper 1994; Piirtola and Era 2006). The increase in fall risk coincides with the increased risk of developing chronic disease associated with increased age due to sedentary living (Besson et al. 2008). Age related declines in strength, endurance and balance ability mainly account for severe falls (Lord 2007). Specific training to improve balance (e.g., Tai Chi) has been reported to improve postural control, injury and fall prevention in older adults (Bernett et al., 2003; Lord et al., 1995; 2003; Park et al., 2008; Ramsbottom 2004; Wolf 1997; 2001). However, such training interventions do not provide a significant cardiovascular training stimulus and therefore have limited benefits to health.

Few people perform endurance training to improve balance (Jakobsen et al. 2011). Although 30-min of balance training each day may be recommended to improve fall risk factors (Gschwind et al. 2013) it would be beneficial if 20 - 30 minutes of endurance training could simultaneously train cardiovascular fitness and balance (Donath et al. 2014; Gardner et al. 2000). Furthermore, it is likely that for older adults, adherence to physical activity interventions would be better if balance training is integrated into endurance training (Buchner et al. 1997b). Therefore, it is useful to identify which types of endurance training improve balance most. While cycling requires less balance capacity than other exercises such as walking there is growing evidence that cycling training is associated with increased leg strength, muscle endurance and postural stability which are important risk factors in falls reduction (Hassanlouei et al. 2014; Bouillon, Sklenka and Ver 2009; Rissel et al. 2013). From a strength and cardiorespiratory perspective, cycling training might provide an appropriate training stimulus and thus may be a feasible training modality for improving general fitness and balance among older adults.

Traditionally, lower limb training (e.g., balance, resistance and/or endurance) has been conducted for fall prevention purposes among the older population (Howe et al. 2008). However, muscle proprioceptive information from the upper extremity is also known to be important in the control of upright stance (Massion 1992; Roll and Roll 1988), as evidenced by an increase in postural sway following localised muscle fatigue of the neck (Schieppati, Mardone and Schmid 2003), deltoids (Nussbaum, 2003) and the trunk extensors (Vuillerme, Anziani and Rougier 2007). More recently it has been shown that increasing trunk muscle strength improves postural stability and may represent promising fall preventative strategies for older adults (Granacher et al. 2012). Nevertheless, the protocols proposed in the literature to improve trunk strength do not provide a cardiovascular challenge. As noted previously, in the interest of attrition it is increasingly important to develop multidimensional exercise modes which can concurrently improve cardiovascular fitness and balance. Arm crank ergometry (ACE) training offers a promising exercise modality for improving upper body stability in addition to cardiovascular health (Sawka 1986; Talbot 2012). For example, ACE training has been shown to provide an effective training stimulus in clinical populations (Saxton et al. 2011; Tew et al. 2009; Zwierska et al. 2007), and both healthy young (Bottoms and Price 2014; Lewis et al., 1980; Loftin et al., 1988) and older adults (Pogliaghi et al., 2006). Several studies have reported that training effects with the lower body can be transfered to another mode of exercise (Lewis et al., 1980; Loftin et al., 1988; Pogliaghi et al. 2006). This cross transfer effect is a result of central mechanisms, such as changes in cardiac output and stroke volume (Loftin et al., 1988) and thus transferable from one mode of exercise to another. Currently, little data is available in healthy older adults. From the perspective of improving upper body strength and general cardiovascular health, ACE training may subsequently provide an effective training stimulus without fatiguing the lower limbs and increasing the risk of falls immediately following exercise. It would be useful to examine the effects of ACE training on balance among older adults.

Since many recreational and daily activities require the use of the upper body in some manner with the arms often requiring sustained effort to a greater extent than leg work (Hellerstein 1977; Franklin 1989; Waller and Prettyman 2012), it seems reasonable to encourage older adults to train the arms as well as the legs. However, there is an absence of studies on postural control and upper body exercise training, probably because most upper body sports involve sitting (kayaking) or kneeling (canoeing) and therefore may not be considered to challenge the postural control system enough to elicit postural adaptations. In contrast, Stambolieva et al. (2012) suggested that the movement patterns observed during canoeing and kayaking, challenge frontal plane postural control. With respect to ACE, isometric work of the core is central to arm ergometry as it is used for upper body stabilisation (Sawka 1986), allowing the arms to generate propulsive forces to the cranks (Smith et al., 2008). In order to maintain an upright posture during ACE, the trunk musculature must continuously compensate for the displacement of the torso caused by the motion of the arms (Grigorenko et al. 2004). Rotational movements of the trunk musculature cause significant displacement of the core along the frontal plane during asynchronous arm cranking and might elicit favourable postural stability and balance adaptations in mediolatertal postural stability. This is important because mediolateral control of sway is dependent on movement of the hips and trunk to maintain the centre of mass within the base of support (Winter et al. 1996; Winter et al. 1993). Thus, increased mediolateral postural sway in older adults who fall (Piirtola and Era 2006) might be due to inadequate contribution of the trunk to balance (Smith et al. 2010). When considering that quiet standing sway in the anteroposterior direction is primarily controlled by the ankle plantar and dorsi flexors and mediolateral sway is maintained by hip and trunk frontal plane movers (Winter et al. 1996; Winter et al. 1993), adaptations in postural stability may be different following upper and lower body exercise training.

Currently, the author is aware of only one study (Donath et al. 2014) which has examined exercise induced changes of muscle activity after the completion of lower body exercise in older adults. However, these findings were limited to the acute responses to exercise. The authors reported that elevated muscle amplitudes after exercise led to an increase in muscle coativation (Donath et al. 2014), which is associated with poorer postural stability (Nagai et al. 2011). In this context, an increase in muscle amplitude would increase joint stiffness in order to minimise the destabilising effects of exercise and thus reflect a less efficient neuromuscular strategy (Donath et al. 2014). In contrast, a reduction in muscle amplitudes during standing with exercise training may suggest a more efficient neuromuscular-skeletal system. The authors are not aware of any studies which have explored acute and chronic exercise induced changes of muscle activity in older adults. Information of muscle amplitudes before and after acute and chronic exercise may help address the mechanisms that lead to increased fall risk with acute exercise and/or improved balance with chronic exercise.

The purpose of this study was to determine; (1) the effects of acute ACE and CE on postural sway before and after 6-weeks of upper (UBX) or lower body (LBX) exercise training, (2) the potential benefits of training on both specific exercise modes and cross transfer effects, (3) the effects of UBX and LBX training on functional outcome measures (i.e., walking speed, hand grip strength, reach distance).

Research Hypothesis (H_4): 6-weeks of upper and lower body exercise training will elicit improvements in specific and cross transfer exercise capacity

Research Hypothesis (H_5): 6-weeks of upper and lower body exercise training will elicit favourable changes in balance and functional ability.

Null Hypothesis ($H0_4$): No effect of 6-weeks of upper and lower body exercise n specific or cross transfer exercise capacity

Null Hypothesis ($H0_5$): No effect of 6-weeks of upper and lower body exercise on postural sway and functional ability.

7.2 Methods

7.2.1 Participants

Eighteen healthy older adults were randomly assigned to either an upper body exercise (UBX) training group or lower body exercise (LBX) training group. Participant demographics are reported in Table 7.1. There were no statistically signifcant differences between groups. All of the participants were considered as sedentary (i.e., were not meeting ACSM guidelines of exercising three times per week for 20-min at an intensity corresponding to at least 50 % HR_{MAX}) (Nelson et al. 2007). Participants confirmed they had not been involved in any scheduled exercise for at least two years. All volunteers underwent a preliminary examination to evaluate potential contraindications to exercise, which included a pulmonary function test (peak expiratory flow) (Mini Wright, Clement Clark international, UK) and measurement of resting systolic and diastolic blood pressure (Emron, M3, Japan). Those with a blood pressure over 140/90 mmHg were asked to provide clearance from their general practitioner. Cutoffs for peak expiratory flow were extrapolated from the manufacturers norm reference values based on age, gender and height. Additionally, all participants were screened using the departmental health screening questionnaire and those passed as healthy with no contraindications for exercise were approved to participate. A total of 20 adults were eligible for inclusion, but two older males withdrew themselves from the LBX group after three and four weeks of training, respectively.

7.2.2 Study Design

Participants initially visited the laboratory on five occasions prior to 6-weeks of moderate intensity LBX or UBX training. The first five visits included; (1) a maximal incremental exercise test on both an arm crank ergometer (ACE) and (2) a cycle ergometer (CE) to determine peak oxygen uptake (VO_{2PEAK}) and peak minute power (W_{PEAK}), (3) an experimental test battery consisting of balance and functional balance measures and two experimental trials which comprised of (4) CE and (5) ACE exercise for 20-min at 50% mode specific W_{PEAK} to determine the acute effects of exercise on postural sway. The training period consisted of 6-weeks of arm crank ergometry or cycle ergometry, three times per week, separated by at least one full day of rest. Upon completion of the training participants were re-tested on a further five occasions repeating the protocols outlined above. All tests were performed at the same time of day (±1h) to avoid any circadian rhythm effects (Reilly 1990) and time of day influences on balance (Gribble et al. 2007).

Table 7.1: Participant demographics for each exercise training group

Variable	UBX training	LBX training
Age (years)	66.2 ± 3.9	65.5 ± 7.8
Height (cm)	163 ± 10	163 ± 7
Mass (kg)	65.3 ± 13.6	65.5 ± 13.2
Male / Female	4 / 6	2 / 6

7.2.3 Maximal Exercise Tests

Pre training, all participants visited the laboratory on two separate occasions separated by at least 72 hours. All participants completed a graded incremental exercise test on an arm crank ergometer and cycle ergometer to determine each individual's ergometer specific $\dot{V}O_{2PEAK}$ and W_{PEAK} as described in section 3.7. The latter was used to prescribe exercise intensity for the experimental trials and training protocols. Post training, all participants completed a second graded incremental exercise test on an arm crank ergometer and cycle ergometer. All trials were completed in a counterbalanced order. Expired gas was analysed using a breath-by-breath online gas system (MetaMax, Cortex Biophsik, Borsdorf, Germany) for oxygen uptake ($\dot{V}O_2$) and minute ventilation ($\dot{V}_E$) and respiratory exchange ratio (RER). Heart rate was continually monitored (Polar Electro, Oy, Finland) and recorded in the final 10 s of each incremental stage and immediately upon reaching volitional exhaustion. A rating of perceived exertion for both local (working muscles; RPE_L) and central (cardiorespiratory; RPE_C) using the 6–20 point Borg scale (Borg, 1982) was obtained at the same time as HR and immediately upon reaching volitional exhaustion.

7.2.4 Experimental Trials

At least 72 hours after the pre-training maximal exercise tests, participants visited the laboratory on two further occasions to perform a 20-min ACE and CE submaximal exercise test at an intensity corresponding to 50% of each individual's mode specific W_{PEAK}. Each test was separated by a minimum of 48 hours. Posturographic assessment before and after exercise are outlined in section 3.9.3. Post exercise training, all participants completed a further two submaximal exercise tests for ACE and CE for 20-min at the same absolute intensity as the pre training mode specific trials. Surface EMG were recorded by a portable biomonitor (Mega Electronics Ltd, ME6000, Finland) at a sampling frequency of 1000 Hz at the same time of posturographic trials (section 3.11). The six muscles of interest were the tibilais anterior (TA), gastrocnemius medialis (GS), biceps femoris (BF), rectus femoris (RF), rectus abdominus (RA) and erector spinae (ES) muscles on the right side of the body.

7.2.5 Endurance training program

All participants underwent a supervised training program consisting of three exercise sessions per week for 6-weeks using a mode specific ergometer. The exercise was based on the basic principles of training including overload, progression, individualisation and specificity as outlined by Taylor and Johnson (2008). In line with recommendations, despite having a lower initial level of fitness, older adults should follow the same exercise prescription as their younger counterparts (Judge, Kenny and Kraemer 2003).

Each participant completed a total of 18 training sessions performed on a cycle ergometer (Monark 824E Ergomedic, Monark, Varberg, Sweden) or arm crank ergometer (Lode Angio BV, Groningen, Netherlands). Throughout training, participants were encouraged to continue their normal diet and to maintain habitual activity levels. All exercise sessions were performed at a similar time of day (±1h). Training sessions were separated by at least 24 hours (i.e., Monday, Wednesday, Friday). Each training session consisted of continuous exercise lasting 20-35 min at intensities based on the pre-determined W_{PEAK} (Table 7.2). Exercise intensity was set at 50, 60 or 70% of the W_{PEAK}, which has previously been used in participants who are unconditioned (Saxton et al. 2011; Tew et al. 2009; Zweirska et al. 2006) and otherwise healthy older adults (Pogliaghi et al. 2006) for both arm and leg training. Prior to each session, participants were asked to complete a 5-min warm up on the unloaded ergometer at a cadence of 60 rev.min^{-1}. This warm up was consistent with those used in the pre-training experimental trials and maximal exercise tests and also those used in study one and two (Chapter 4 & 5).

As recommended the training regimen was designed to cater for adults who were sedentary and were not accustomed to continuous exercise of a prolonged nature (Nelson et al. 2007). While exercise intensity was relative to each individual's maximal exercise tolerance exercise duration increased every two weeks throughout the 6-week training period to encourage progression (Table 7.2). For the first two weeks exercise duration was set at 20-min, followed by an increase in 5-min at weeks 3-4. It was anticipated that all participants would have a similar level of general fitness after this time, therefore allowing for exercise duration to be increased to 35-min per session in weeks 5-6. All participants tolerated the progression well and completed every session.

7.2.6 Outcome Measures

Postural control performance was measured by static and dynamic functional balance tests. The Timed Up and Go test (TUG), Multi-Directional Functional Reach test (MDFR), modified

Star Excursion Balance test (SEBT) and fast gait speed test were used to determine dynamic balance. Static balance was assessed by single limb stance time and posturographic assessment of bipedal standing balance on a fixed surface while holding a load and standing on a compliant surface. Upper body functional performance was assessed using the Seated Medicine Ball Throw, 30-sec arm curl test (Rikli and Jones 2013) and the hand grip strength test. All tests were performed in a randomised order during two separate visits to the laboratory (Figure 7.1). All participants completed outcome measures during their first two visits. The order of testing allowed the principal investigator to administer the Berg Balance Scale (BBS) which was always the first test to be completed. Participants who were balance impaired (< 52 / 56 on the BBS (Berg et al. 1992) were then excluded from the study. However, all participants were eligible for inclusion and scored $\geq$ 52 on the BBS.

Table 7.2 Summary of the 6-week ACE and CE aerobic training programme

Training Session	**Duration (min)**		
	Weeks 1-2	**Weeks 3-4**	**Weeks 5-6**
Session 1 & 3			
50 % W_{PEAK}	10	12	15
60 % W_{PEAK}	5	7	10
50 % W_{PEAK}	5	6	10
Total Duration	20	25	35
Session 2			
50 % W_{PEAK}	10	10	15
60 % W_{PEAK}	7	9	12
70 % W_{PEAK}	3	6	8
Total Duration	20	25	35

7.2.6 i Walking Tests

The TUG was administered by the principal investigator pre and post training. Participants were initially seated on a back supported chair (seat height 46 cm) with their arms resting on the lap. They were then asked to stand up from the chair without the use of the hands, walk a distance of 3 m away from the chair (marked out on the floor), turn around, walk back to the chair and sit back down as quickly and safely as possible. The time taken to complete the test was measured in seconds with a stopwatch. Participants were aware that each trial would be timed. Timing began as soon as the participants back left the chair and ended when the back returned to the same position. A practice trial was performed followed by three timed trials and the fastest trial was used for analysis (Shumway-Cook, Brauer and Woolacott 2000). In addition, three consecutive trials of FGS were recorded as each participant walked along an 8 m marked runway as outlined in section 6.2.4.

7.2.6 ii Upper body functional tests

Maximal hand grip strength was measured in kilograms using a hand held dynamometer (Lafayette Instrument Co., IN, USA). Specific protocols for grip strength are outlined in 6.2.7. As an indication of upper body power participants completed a seated medicine ball throw, as described by Harris et al. (2011). Participants were asked to sit on a chair with the back of the chair placed against a wall. Each participant was instructed to sit in the chair with their back against the back rest for support and also to keep their feet flat on the ground. To account for different arm lengths participants were asked to hold the ball with both hands with their arm fully extended and drop the ball onto the tape measure, which was then adjusted so that this point was the zero mark (Harris et al. 2011). Each participant was asked to perform three practice trials, pushing the ball away from the centre of the chest using a similar technique as a basketball chest pass. The principal investigator demonstrated this movement. The optimal angle of release was advised, as these instructions were shown to be useful in previous investigations (Harris et al. 2011). Participants then performed three throws. The maximal distance of each throw was marked on a measuring tape to the nearest 5 cm (it is acknowledged that this test may lack precision). Each participant performed three trials with both a 1.5 Kg ball and a 3.0 Kg ball in a randomised order. Approximately 1 min rest was allowed between each trial.

7.2.6 iii Functional Reach and Star Excursion Balance Test

Particpants completed the MDFR and a modified version of the SEBT (sections 6.2.5 and 6.2.6). Based on the results of study 3 (Chapter 6), the original SEBT was modified to the Y balance test (anterior, posteriolateral, posteromedial). Following familiarisation, participants performed three reaches in each direction for both the MDFR and SEBT. The reach directions were performed in a randomised order.

7.2.6 iv Timed bridge

Participants were asked to place their hands and elbows on the floor shoulder width apart, so that the elbow joint was flexed to 90°. The toes were aligned under the ankles and the forearms under the shoulders. Participants were encouraged to keep a straight line posture. The position was held for as long as possible. The test ended when the participants could no longer hold the position or the investigator terminated the test due to a change in position. A practice trials of 10 s was allowed followed by a single timed trial.

7.2.6 v Postural Sway

Postural sway was measured in two different conditions; (1) standing on a compliant surface and (2) standing on a firm surface while holding a bag. Section 3.9.4 describes the specific procedures of each test.

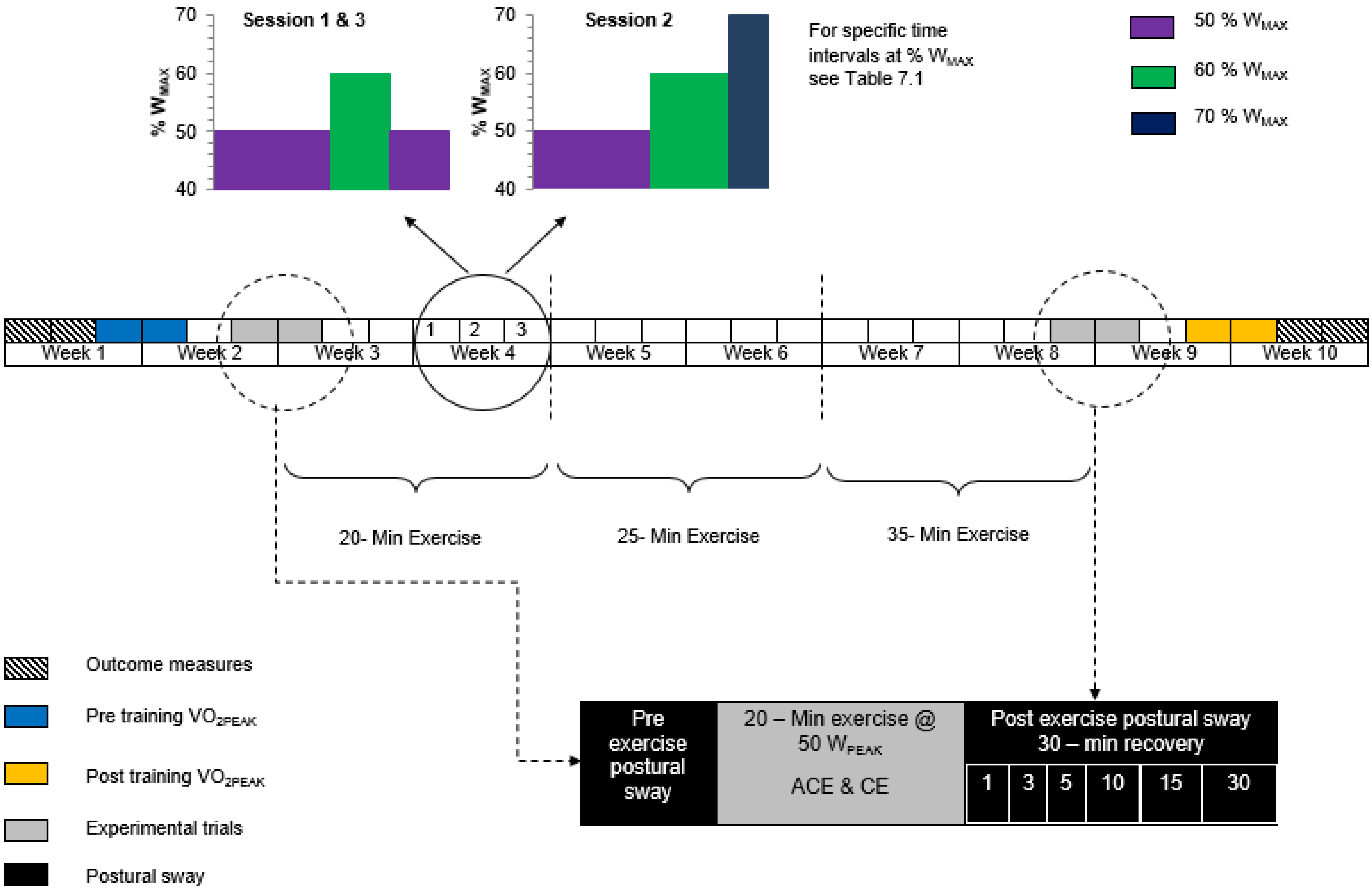


Figure 7.1: Schematic diagram of experimental setup for training intervention and experimental trials

7.2.7 Statistical Analysis

All cardiorespiratory and perceptual variables for submaximal exercise trials were analysed by a 3-way (time $\times$ mode $\times$ training status) repeated measures ANOVA (e.g., time; 0, 5, 10, 15, 20; $\times$ mode; ACE and CE $\times$ training status; pre and post). Additionally, a three-way analysis of variance (ANOVA) with repeated measures was conducted to examine changes induced by exercise on each sway measure (time; 0 [pre exercise], 1 [post exercise], 3, 5, 10, 15 and 30 min $\times$ mode; ACE and CE $\times$ training status; pre and post). For peak physiological responses and outcome measures (e.g., grip strength, functional reach etc) a two-way ANOVA was used (training status; pre and post $\times$ group; UBX and LBX). Where the result of the ANOVA was significant Scheffe's post hoc analysis was undertaken by calculating the difference required between means for significance at the level of $P \leq 0.05$ (Vincent 2005). Data was analysed using PASW version 17.0 (SPSS Inc., Chicago, IL). Statistical significance was set at $P < 0.05$. The above analyses enabled the following comparisons;

i. Maximal and submaximal physiological and perceptual responses to ACE and CE following upper body exercise training (UBX)
ii. Maximal and submaximal physiological and perceptual responses to CE and ACE following lower body exercise training (LBX)
iii. Postural sway responses to submaximal ACE and CE following UBX training
iv. Postural sway responses to submaximal ACE and CE following LBX training

7.3 Results

7.3.1 Incremental exercise tests

Data outlined in Table 7.3 demonstrates that all participants successfully reached the criteria for peak oxygen uptake for all exercise tests (Bird and Davison 1997). Peak values of W_{MAX}, $\dot{V}O_{2PEAK}$, $\dot{V}_{Emax}$ and HR_{MAX} were significantly greater during CE compared to ACE ($P \leq 0.05$). As expected, ACE $\dot{V}O_{2PEAK}$ represented as a fraction of CE was 72 ± 14 % and 71 ± 8 % for the UBX and LBX groups, respectively.

7.3.1i Specific training effect

When participants were tested on the same ergometer on which they trained, comparable improvements between groups were observed. For the UBX group significant effects of training were observed for W_{MAX} (P = 0.001), $\dot{V}O_{2PEAK}$ (P = 0.001), $\dot{V}_E$ (P =0.001), HR_{MAX} (P = 0.016), and RPE_C (P = 0.001) during ACE. For the LBX group, a significant increase in CE W_{MAX} (P = 0.001), $\dot{V}O_{2PEAK}$ (P = 0.001), $\dot{V}_E$ (P = 0.002) and HR_{MAX} (P = 0.008) was observed from pre to post-training. Absolute mode specific $\dot{V}O_{2PEAK}$ increased by ~ 26 % for both UBX and LBX groups ($P \leq 0.001$). Similarly, mode specific W_{MAX} increased by 26 and 28 % for LBX and UBX, respectively. Hand grip strength improved following ACE training (section 7.3.14) and the combined change of right and left hand rip strength was significantly correlated with the percentage increase in W_{MAX} (r = 0.886; P = 0.006). There was a positive correlation between the percentage increase in hand grip strength (r = 0.83; P = 0.002) and percentage increase in 30 s arm endurance test (r = 0.82; P = 0.003) with the increase in W_{MAX} in the UBX group.

Table 7.3: Summary of peak physiological responses obtained upon volitional exhaustion (peak values) during incremental cycle ergometry (CE) and arm crank ergometry (ACE) before (PRE) and after (POST) training for the UBX and LBX training groups

Group	Variable	CE			ACE		
		PRE	**POST**	**% Δ**	**PRE**	**POST**	**% Δ**
UBX	W_{MAX} (Watts)	98 ± 25	108 ± 23 *	11.7 ± 5.0	51 ± 13 * †	65 ± 16 * †	28.1 ± 16.1
	$\dot{V}O_{2PEAK}$ (L·min^{-1})	1.44 ± 0.43	1.74 ± 0.39 *	12.7 ± 8.4	1.10 ± 0.34 * †	1.50 ± 0.28 * †	25.6 ± 9.9
	$\dot{V}O_{2\,PEAK}$ (ml·min·kg^{-1})	23 ± 7	26 ± 7 *	12.7 ± 8.2	17 ± 4 * †	22 ± 5 * †	25.6 ± 9.4
	$\dot{V}_E$ (L·min^{-1})	54 ± 14	58 ± 12	10.2 ± 14.1	47 ± 11 * †	55 ± 13	17.5 ± 13.1
	RER	1.16 ± 0.03	1.15 ± 0.06	1.1 ± 5.0	1.14 ± 0.03	1.16 ± 0.06	1.95 ± 6.61
	HR_{MAX} (beats·min^{-1})	152 ± 31	155 ± 27	4.2 ± 5.0	143 ± 16	153 ± 11	7.4 ± 6.2
	RPE_L	20 ± 1	20 ± 1	1.7 ± 4.4	20 ± 1	20 ± 1	1.0 ± 3.0
	RPE_C	19 ± 2	19 ± 2	1.0 ± 3.4	18 ± 2	16 ± 2 * †	11.4 ± 9.8
LBX	W_{MAX} (Watts)	103 ± 56	129 ± 73 *	26.4 ± 12.0	57 ± 27 * †	62 ± 31 * †	11.6 ± 4.4
	$\dot{V}O_{2PEAK}$ (L·min^{-1})	1.55 ± 0.71	1.97 ± 0.56 *	26.3 ± 8.8	1.17 ± 0.49 * †	1.31 ± 0.58 * †	12.4 ± 7.9
	$\dot{V}O_{2PEAK}$ (ml·min·kg^{-1})	23 ± 8	30 ± 12 *	26.2 ± 8.6	18 ± 6 * †	20 ± 7 * †	12.4 ± 7.7
	$\dot{V}_E$ (L·min^{-1})	62 ± 29	71 ± 27	16.1 ± 12.9	43 ± 19 * †	59 ± 31 * †	21.5 ± 17.1
	RER	1.17 ± 0.02	1.15 ± 0.06	0.4 ± 1.6	1.18 ± 0.05	1.15 ± 0.01	-5.4 ± 9.6
	HR_{MAX} (beats·min^{-1})	153 ± 25	158 ± 16	5.0 ± 11.5	144 ± 16	150 ± 13	4.0 ± 5.7
	RPE_L	20 ± 1	19 ± 1	-1.2 ± 3.6	20 ± 1	18 ± 1	-1.0 ± 3.2
	RPE_C	19 ± 1	19 ± 2	- 2.4 ± 9.1	20 ± 1	16 ± 1 *	-6.4 ± 10.1

* Significant difference with pre-training values; † Significant difference with CE test

7.3.1ii General training effect (cross transfer effect)

When participants were tested on the non-specific training ergometer a similar improvement was observed for W_{MAX} and $\dot{V}O_{2PEAK}$ in both training groups of ~ 12 %. Increases in W_{MAX} and $\dot{V}O_{2PEAK}$ were observed for the UBX group during CE (P = 0.0015 and P = 0.004, respectively) and for the LBX during ACE (P = 0.006 and P = 0.009, respectively). These findings show that approximately half of the increase in W_{MAX} and $\dot{V}O_{2PEAK}$ was transferable from the training specific mode of exercise to the non-specific training exercise mode (Figure 7.2).

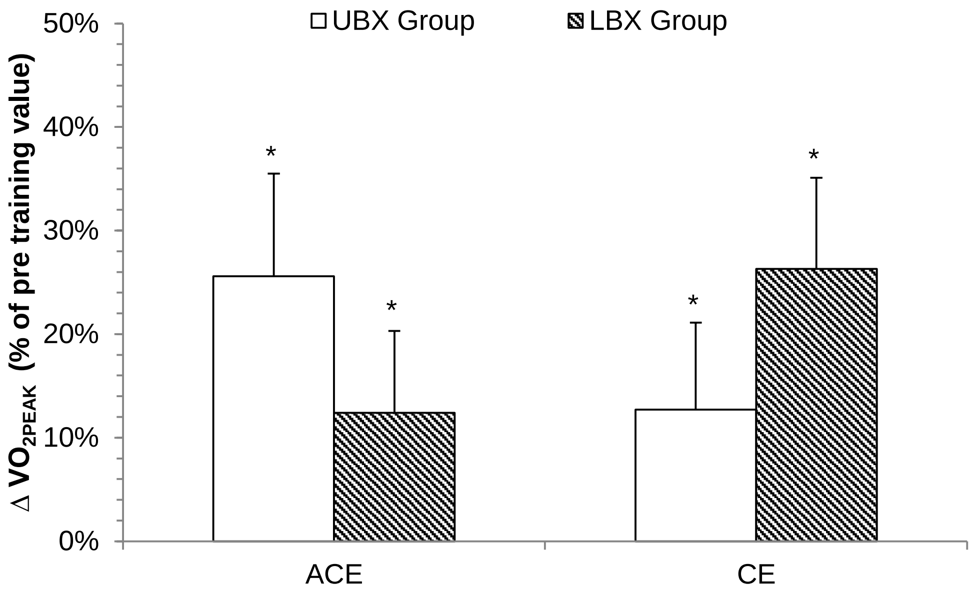


Figure 7.2: Percent change (mean ± SD) in peak oxygen uptake for mode specific training and non-specific training mode before and after training. * (P < 0.05) between pre and post training.

7.3.2 Submaximal exercise trials

7.3.2 i Upper body exercise vs. lower body exercise in the UBX training group

Main effects of mode were observed for $\dot{V}O_2$ (pre and post training; P = 0.001), $\dot{V}_E$ (pre and post training; P = 0.001) and HR (pre training; P = 0.057, post training; P = 0.035) (Table 7.4). Absolute $\dot{V}O_2$ (P = 0.001), HR (P = 0.001), $\dot{V}_E$ (P = 0.001), RPE_L ($P \leq 0.01$) and RPE_C ($P \leq$ 0.01) all significantly reduced from pre to post UBX training during ACE. Similarly, $\dot{V}O_2$ (P = 0.020), HR (P = 0.002), RPE_L ($P \leq 0.01$) and RPE_C ($P \leq 0.01$) were significantly reduced during CE following UBX training. The cross transfer effects were ~ 50 % of the generic mode specific effects (Figure 7.3) as noted for the incremental exercise tests.

7.3.2 ii Lower body exercise vs. upper body exercise in the LBX training group

Pre and post training, main effects for mode were observed for $\dot{V}O_2$, $\dot{V}_E$, HR and RER (all $P \leq$ 0.01). Absolute $\dot{V}O_2$, $\dot{V}_E$, HR and RER reduced significantly post training for CE (all $P \leq 0.01$). Similarly, reductions in $\dot{V}O_2$ (P = 0.002), RER (P = 0.006), HR (P = 0.015) and RPE_L (P = 0.004) were reported for ACE following LBX training. Similar to the UBX group, the cross transfer improvements were ~ 50 % of the general training effect, with the exception of HR (Figure 7.3)

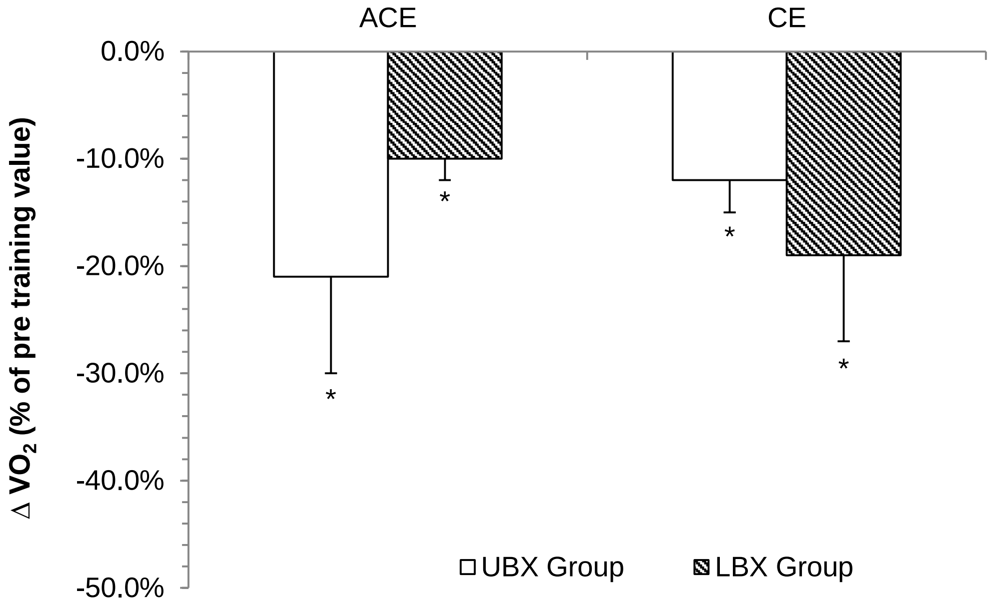


Figure 7.3: Percent change (mean ± SD) in oxygen uptake during submaximal experimental trials before and after training (20 min value). *($P \leq 0.05$) between pre and post training.

Table 7.4: Summary of physiological responses (mean ± SD) obtained at cessation of 20-min submaximal exercise during arm crank ergometry (ACE) and cycle ergometry (CE) for upper body training group (UBX) and lower body training group (LBX) before (Pre) and after (Post) training

* Significant difference with pre-training values; † Significant difference with CE test

Group	**Variable**	**ACE**			**CE**		
		Pre	**Post**	**% Δ**	**Pre**	**Post**	**% Δ**
UBX	$\dot{V}O_2$ (L·min^{-1})	0.80 ± 0.17	0.63 ± 0.17*	-21 ± 12	1.01 ± 0.35	0.87 ± 0.32*	-12 ± 8
	$\dot{V}_E$ (L·min^{-1})	24 ± 6	19 ± 5*	-20 ± 9	34 ± 7	30 ± 7	-11 ± 7
	RER	0.96 ± 0.09	0.87 ± 0.07	-10 ± 4	0.96 ± 0.08	0.92 ± 0.08	-4 ± 4
	HR (beats·min^{-1})	106 ± 24	90 ± 18*	-15 ± 5	116 ± 24	106 ± 19*	-7 ± 6
	RPE_L	14 ± 1	11 ± 1*	-5 ± 3	15 ± 1	13 ± 1*	-3 ± 2
	RPE_C	12 ± 1	10 ± 2*	-3 ± 2	14 ± 1	12 ± 1*	-2 ± 2
LBX	$\dot{V}O_2$ (L·min^{-1})	0.72 ± 0.17	0.62 ± 0.13	- 10 ± 4	1.13 ± 0.32	0.93 ± 0.29*	-19 ± 8
	$\dot{V}_E$ (L·min^{-1})	22 ± 4	19 ± 3	- 10 ± 10	34 ± 8	26 ± 7*	-22 ± 11
	RER	0.94 ± 0.05	0.88 ± 0.05	- 6 ± 3	0.99 ± 0.07	0.89 ± 0.05*	-13 ± 5
	HR (beats·min^{-1})	109 ± 18	95 ± 18*	- 13 ± 4	120 ± 18	100 ± 15*	-15 ± 4
	RPE_L	13 ± 0	12 ± 1*	-1 ± 1	14 ± 1	12 ± 2*	-4 ± 2
	RPE_C	12 ± 0	11 ± 1	-2 ± 2	13 ± 2	12 ± 2	-1 ± 2

* Significant difference with pre-training values;

7.3.2 iii Ratings of Perceived Exertion

Upper body exercise vs. lower body exercise in the UBX training group

There was a time $\times$ mode interaction for RPE_C (P = 0.008). Central RPE was lower for ACE compared to CE before and after UBX training after 10 and 15-min of exercise at the same relative intensity ($P \leq 0.05$). For RPE_L interactions were observed for time $\times$ training status (P = 0.016) which demonstrates that exercise training resulted in a lower RPE_L for both ACE and CE in the UBX training group at 10, 15, and 20-min of exercise ($P \leq 0.05$). There was also a training status $\times$ mode (P = 0.035) interaction, which demonstrates differences in RPE_L were observed between modes pre and post UBX training (P = 0.035). Following UBX training there was a 3-point reduction in RPE_L upon cessation of ACE, compared to a 2-point reduction upon cessation of CE.

Lower body exercise vs. upper body exercise in the LBX training group

No interactions were observed for RPE_C in the LBX training group (Figure 7.6). However, main effects for time (P = 0.001), mode (P = 0.003) and training status (P = 0.001) were observed. For RPE_L interactions were observed for time ^{x} training status (P = 0.047) and training status ^{x} mode (P = 0.030), which suggests that local RPE was significantly lower post training. The changes for RPE_L were small for CE (2 points) and ACE (1 point) post training. In general the change in RPE_L was similar between UBX and LBX training groups.

7.3.3 Physiological responses during training sessions

All participants completed the 6-week, 3 times per week training program. Indicators of absolute and relative training intensity are reported in Table 7.5. Cycling training was performed at a higher absolute power output compared to arm training (Section 7.7.6). As a result absolute HR was typically greater during LBX compared to UBX training, but similar when expressed as a relative percentage of mode specific HR_{MAX}. Local and central RPE were not different between groups throughout training period ($P \geq 0.05$). For relative exercise intensity, both UBX and LBX were performed at a similar % of HR_{MAX}. Both groups exercised above the minimum recommended by ACSM for maintaining and improving cardiorespiratory fitness in older adults (Nelson et al. 2007).

Table 7.5: Indicators of exercise intensity during training sessions, calculated as an average over three weekly sessions for week 1 and week 6

Group	**Variable**	**Week-1**	**Week-6**	***P* value**
UBX	HR beats·min^{-1}	109 ± 19	96 ± 18	0.0003
	% HR_{MAX}	70 ± 8	59 ± 11	0.0001
	RPE_L	13 ± 1	11 ± 1	0.0470
	RPE_C	12 ± 1	10 ± 1	0.0300
LBX	HR beats·min^{-1}	119 ± 17	107 ± 14	0.0001
	% HR_{MAX}	73 ± 11	63 ± 7	0.0020
	RPE_L	14 ± 2	12 ± 2	0.0460
	RPE_C	13 ± 2	12 ± 2	0.0690

Table 7.6: Absolute intensity of UBX and LBX training

	% W_{MAX}		
Training Group	50 %	60 %	70 %
UBX	25 ± 8 W	30 ± 10 W	35 ± 11 W
LBX	51 ± 28 W	62 ± 34 W	72 ± 39 W

7.3.4 Postural sway adaptations

7.3.4 i Upper body exercise training

Anteroposterior COP displacement

An interaction of time ^{x} mode for COP_{AP} was found to be significant (P = 0.011) (Figure 7.4). Before and after UBX training, CE induced an increase in COP_{AP}, returning to baseline values within 10 and 5-min of post exercise recovery, respectively. Baseline and post exercise COP_{AP} was not altered by UBX training ($P \geq 0.05$). Acute ACE had no effects of COP_{AP} ($P \geq 0.05$).

Mediolateral COP displacement

Pre UBX training, CE induced a significant increase in COP_{ML} (P = 0.004) returning to baseline levels within 10-min of exercise completion. After 6-weeks of UBX training, CE had no effects on COP_{ML} ($P \geq 0.05$) at the same absolute intensity. In addition, UBX training resulted in significant improvements in baseline and post exercise COP_{ML} from pre to post training ($P \leq$ 0.05). Acute ACE had no effects of COP_{ML} ($P \geq 0.05$).

COP mean velocity

Pre UBX training, CE induced an increase in COP_{V} (P = 0.001) returning to baseline levels within 5-min of exercise completion. After 6-weeks of UBX training, CE had no effects on COP_{V} ($P \geq 0.05$). In addition, UBX training resulted in significant improvements in baseline and post exercise COP_{V} from pre to post training ($P \leq 0.05$). Acute ACE had no effects of COP_{V} ($P \geq$ 0.05).

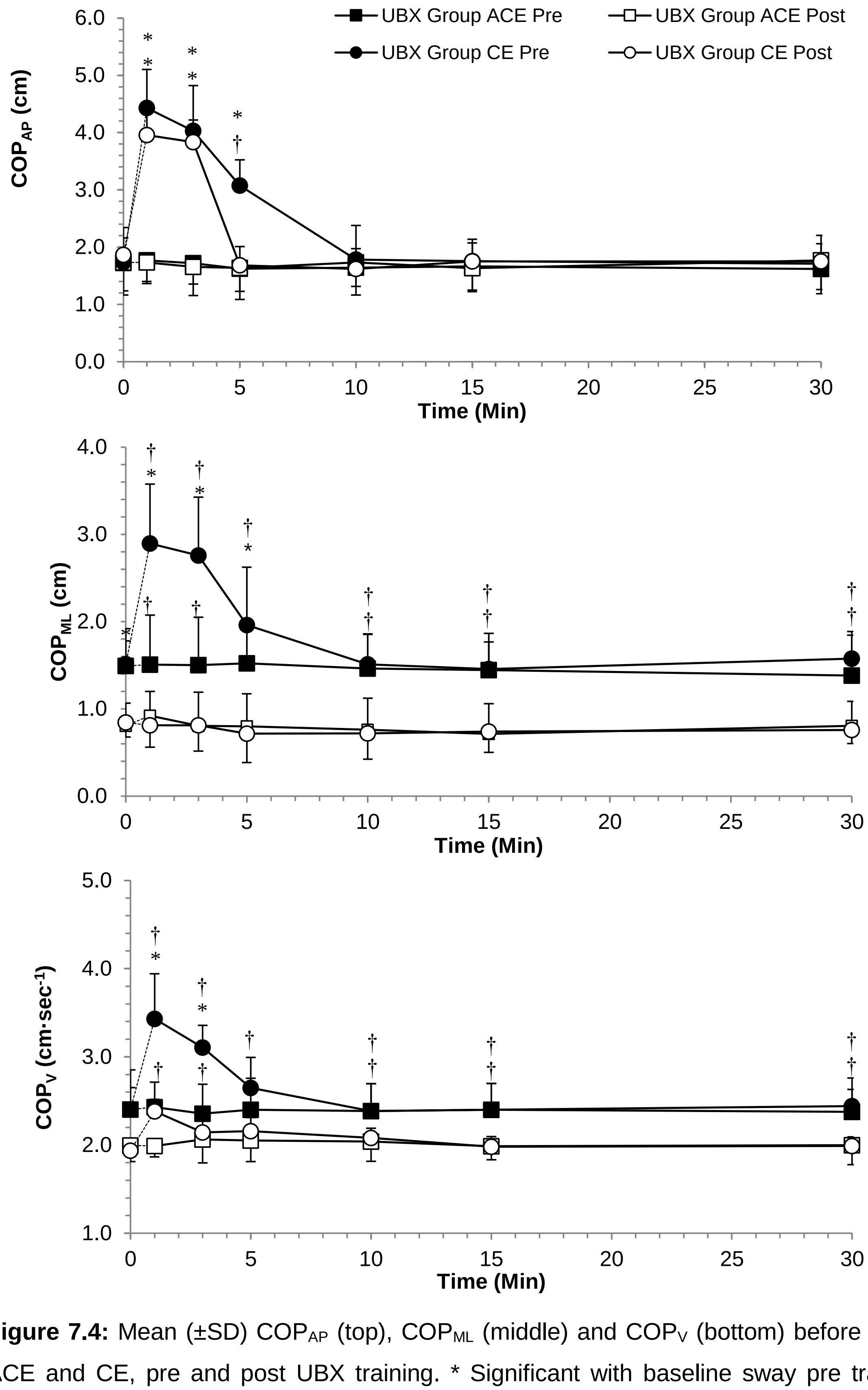


Figure 7.4: Mean (±SD) COP$_{AP}$ (top), COP$_{ML}$ (middle) and COP$_{V}$ (bottom) before and after ACE and CE, pre and post UBX training. * Significant with baseline sway pre training. ** Significant with baseline sway post training. † Significant between pre and post training values for same exercise mode. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

7.3.4 ii Lower body exercise training

Anteroposterior COP displacement

As with the UBX group, a significant interaction of time ^{x} mode observed (P = 0.004). Post hoc analysis showed that pre-training an increase in COP_{AP} was observed after CE, which lasted until 10-min post exercise. There was also an interaction of training status ^{x} mode (P = 0.016). Post hoc tests showed that after LBX training, COP_{AP} was reduced at baseline and following both ACE and CE. Acute ACE had no effects of COP_{AP} ($P \geq 0.05$).

Mediolateral COP displacement

For COP_{ML}, an interaction of time ^{x} mode was found to be significant (P = 0.001). Post hoc analyses revealed that pre and post training CE elicited an increase in COP_{ML}, returning to baseline values within 5-mins of exercise completion. However, a training status ^{x} mode interaction (P = 0.018) was found to be significant for COP_{ML} (Figure 7.5). Post hoc tests showed that LBX training resulted in a significant reduction in baseline and post exercise COP_{ML} for both modes of exercise. Acute ACE had no effects of COP_{ML} ($P \geq 0.05$).

COP mean velocity

Cycling induced a significant increase in COP_{V} (P = 0.001), returning to baseline levels within 10-min of exercise completion. Lower body exercise training removed the effects of CE on the COP_{V} in addition to significant reductions in baseline COP_{V} ($P \leq 0.05$). Acute ACE had no effects of COP_{V} ($P \geq 0.05$).

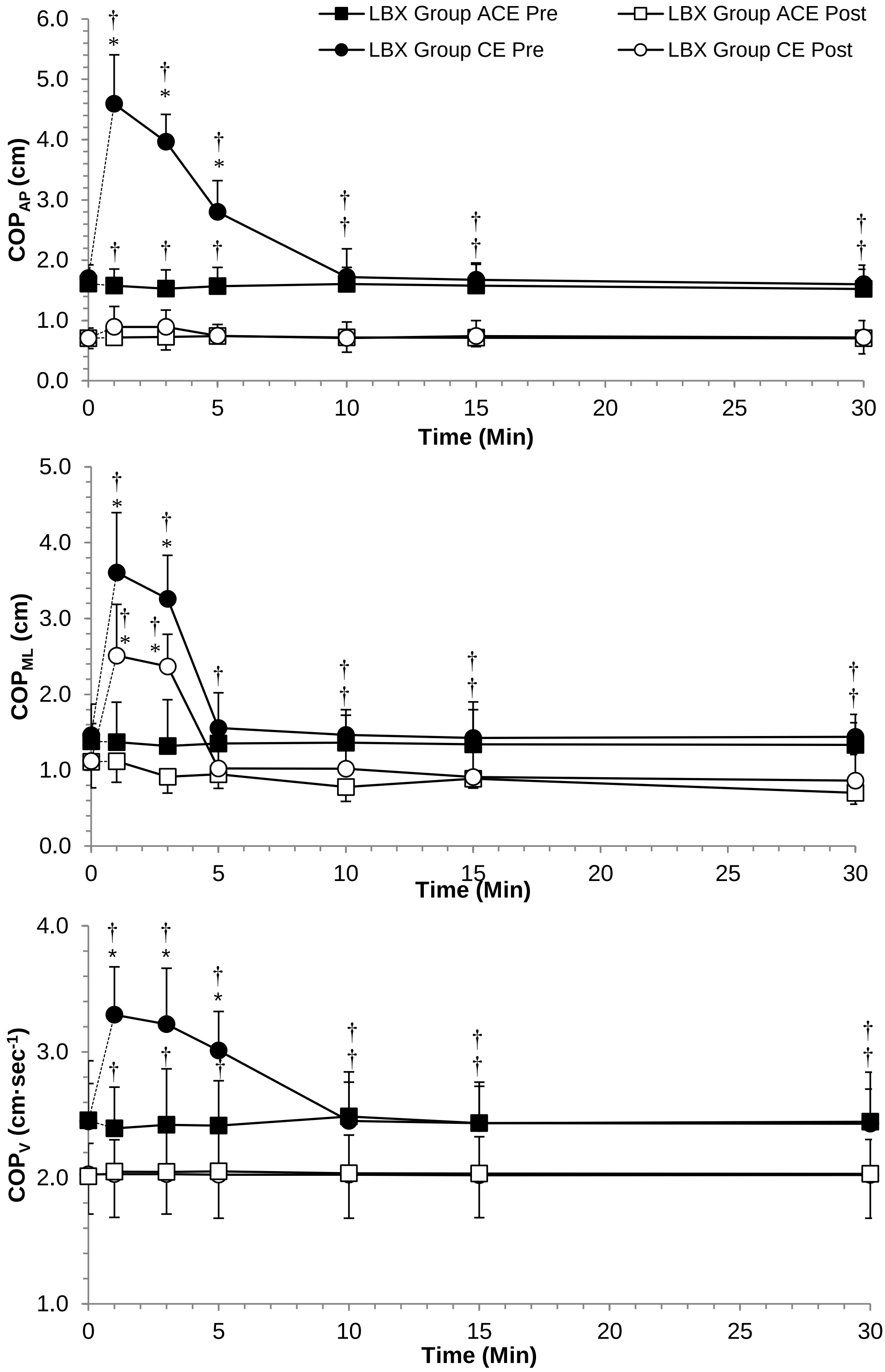


Figure 7.5: Mean (±SD) COP_{AP} (top), COP_{ML} (middle) COP_{V} (bottom) before and after ACE and CE, pre and post LBX training. * Significant with baseline sway pre training. ** Significant with baseline sway post training. † Significant between pre and post training values for same exercise mode. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

7.3.5 EMG responses to UBX and LBX training

7.3.5 i Maximal voluntary isometric contraction

Cycling training resulted in significant improvements for MVIC of the GS (*P* = 0.0002, 64 ± 29 %), TA (*P* = 0.0005, 65 ± 33 %), BF (*P* = 0.001, 44 ± 27 %) and RF (*P* < 0.001) 129 ± 70 %) following LBX training. Following arm training, significant improvements in MVIC were observed for the RA (*P* = 0.0002, 72 ± 34 %) and ES (*P* = 0.006, 33 ± 21 %).

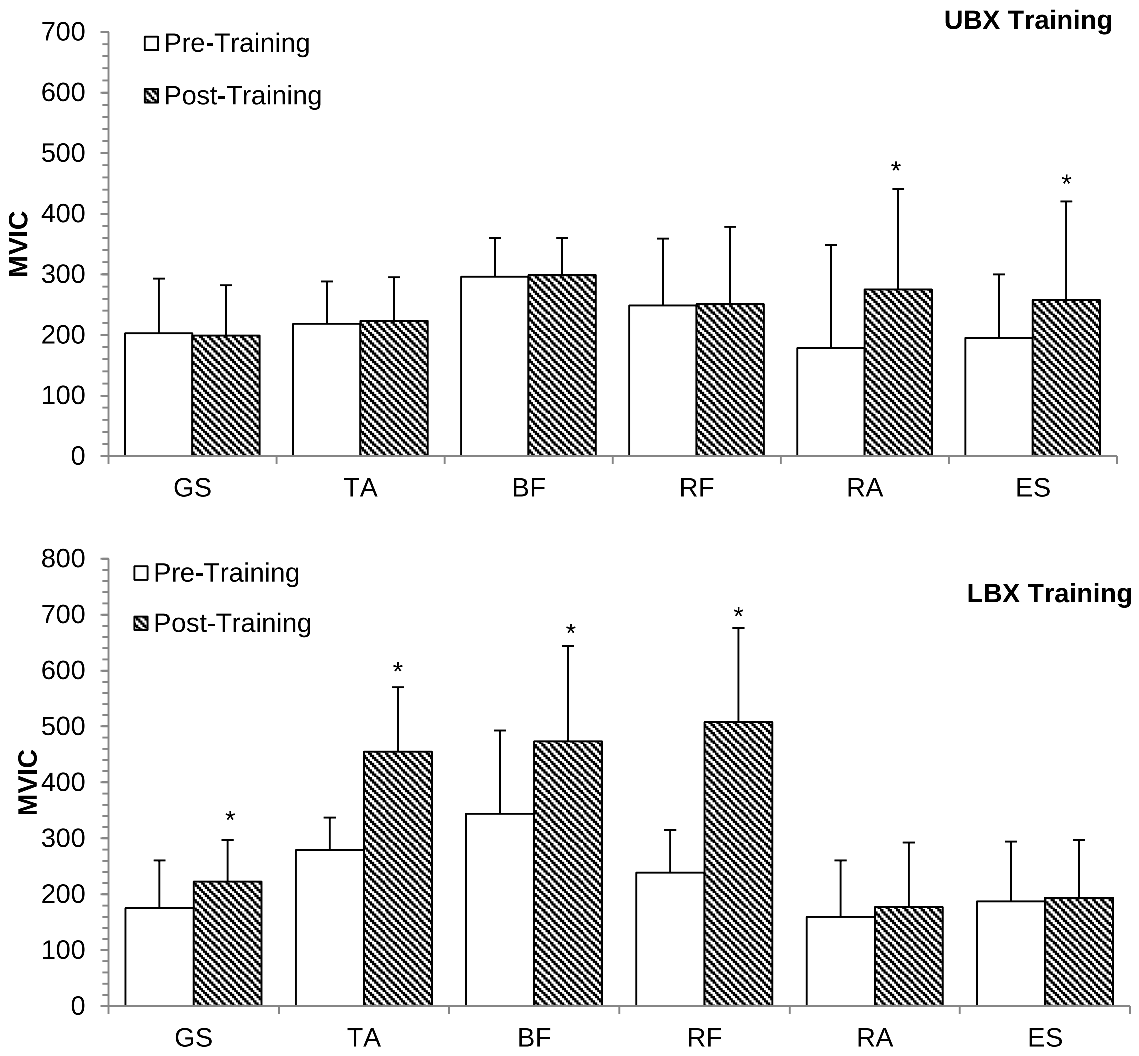


Figure 7.6: Maximal voluntary isometric contraction (MVIC) at pre and post upper (top) and lower body (bottom) exercise training. * Significant increase from pre-training value ($P \leq 0.05$)

7.3.5 ii EMG responses during postural sway

Medial Gastrocnemius

Significant training status × time interactions were observed for EMG amplitude of the GS for the LBX (P = 0.001) and UBX (P = 0.001) training groups. Post hoc analysis revealed a significant increase in amplitude of the EMG_{GS} following both ACE and CE pre-training ($P \leq 0.05$) (Figure 7.7), while no changes were observed in either group post training ($P \geq 0.05$). Post training the amplitude of the EMG_{GS} was significantly lower compared to each time point recorded for both ACE and CE trials ($P \leq 0.05$).

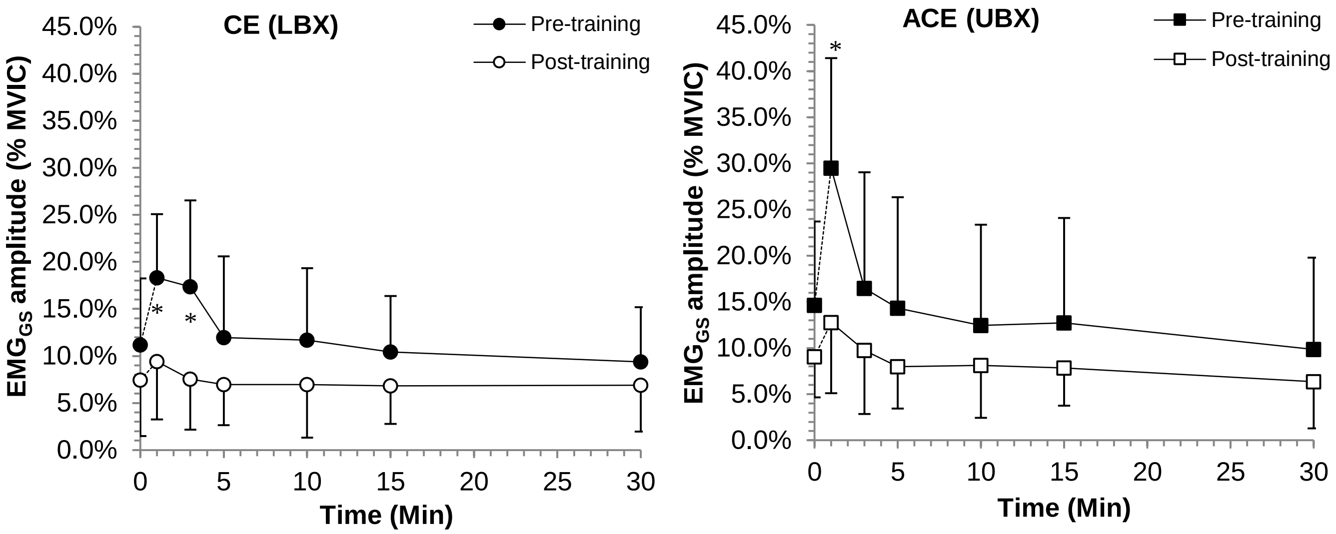


Figure 7.7: Surface EMG of the medial gastrocnemius (GS) recorded before and after CE (left) and ACE (right) during posturographic trials. * Significant with baseline sway pre training in both groups. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

Tibialis anterior

A significant training status × time interaction was observed for the EMG_{TA} in the UBX (P = 0.004) and LBX (P = 0.032) training groups. Post hoc analysis revealed a significant increase in EMG_{TA} amplitude immediately after exercise in both groups during pre-training trials (Figure 7.8). The change in EMG amplitude immedatley after exercise (~ 1 %) was not different between LBX and UBX groups ($P \geq 0.05$). Interestingly, smaller SD's were observed for all post training time points compared to pre training values.

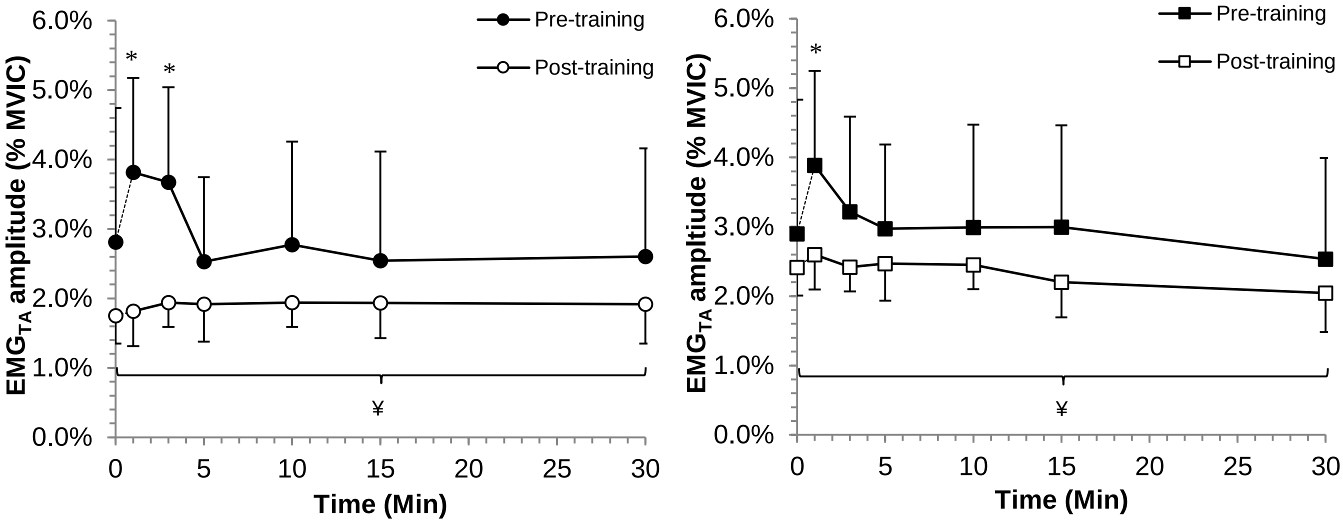


Figure 7.8: Surface EMG of the tibialis anterior (TA) recorded before and after CE (left) and ACE (right) during posturographic trials. * Significant with baseline sway pre training in UBX and LBX training groups. ¥ Significant difference between pre and post trainingn UBX and LBX training groups. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

Rectus femoris

A significant training status × time interaction was observed for the EMG_{RF} in the LBX (P = 0.001) but not the UBX group (P = 0.995) (Figure 7.9). Pre and post training, the EMG_{RF} amplitude decreased significantly returning back to baseline values after 5 min of recovery ($P \leq 0.05$). For the LBX group, baseline amplitude of the EMG_{RF} was not different post training ($P \geq 0.05$). In addition, the first two time points post CE were not different after training ($P \geq 0.05$). In contrast, after 5 min of recovery the EMG_{RF} amplitude was significantly lower post-training compared to pre training trials ($P \leq 0.05$).

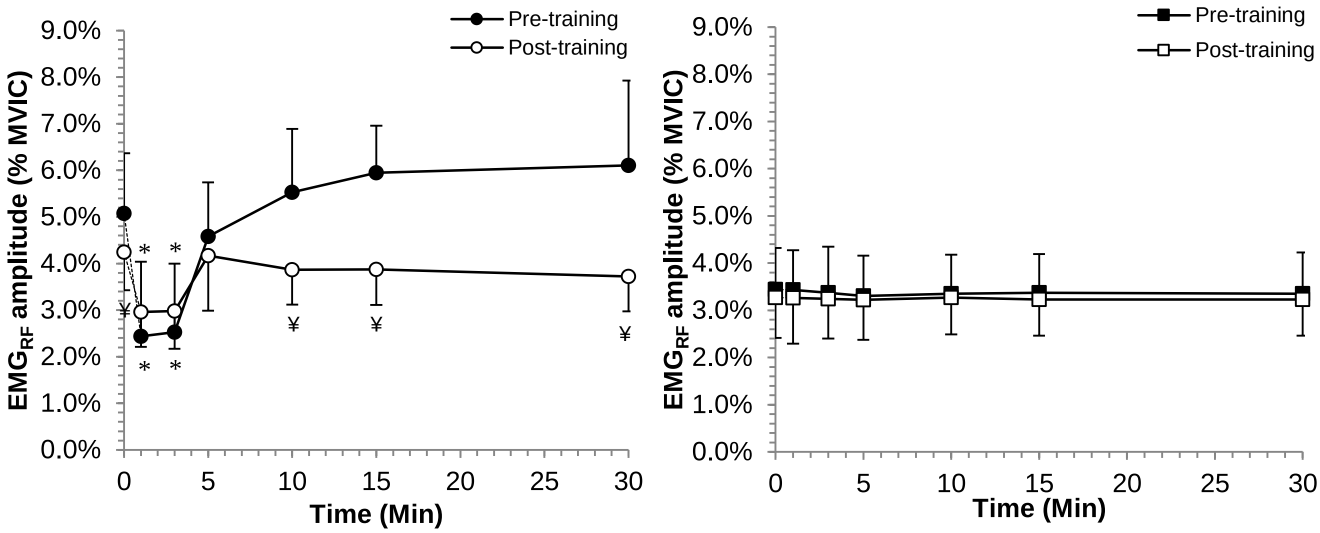


Figure 7.9: Surface EMG of the rectus femoris (RF) recorded before and after CE (LBX) (left) and ACE (UBX) (right) during posturographic trials. * Significant with baseline sway pre training in LBX group. ¥ Significant difference between pre and post training in LBX group. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

Rectus abdominus

A significant training status $\times$ time interaction was observed for the EMG_{RA} in the UBX (P = 0.001) but not the LBX group (P = 0.438). Pre and post UBX training, the EMG_{RA} amplitude decreased significantly returning back to baseline values after 10 min of recovery ($P \leq 0.05$) (Figure 7.10). Post hoc analysis revealed a reduction in baseline EMG_{RA} amplitude for the UBX group ($P \leq 0.05$). The amplitude of the EMG_{RA} after ACE (1 – 5 min) was the same before and after training ($P \geq 0.05$). However, there was a significant reduction in EMG_{RA} amplitude at 10, 15 and 30 minutes compared to pre training trials ($P \leq 0.05$). Following CE training there was no difference in EMG_{RA} amplitude during all time points compared to pre training.

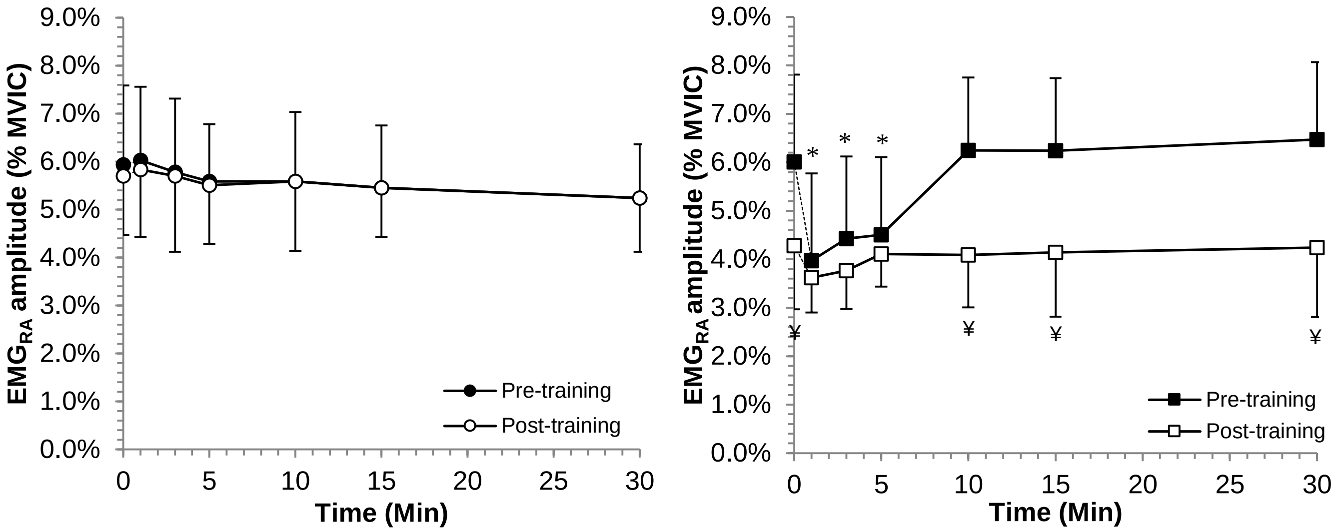


Figure 7.10: Surface EMG of the rectus abdominus (RA) recorded before and after CE (left) and ACE (right) during posturographic trials. * Significant with baseline sway pre training in UBX training group. ¥ Significant difference between pre and post training in UBX training group. Dashed lines represent transition from pre- to post-exercise. Time point 0 represents an average of three pre-exercise trials.

Biceps femoris and Erector spinae

No training status × time interactions were observed for the amplitude of the EMG_{BF} for the LBX (P = 0.534) or UBX (P = 0.389) training groups. Similarly, no training status × time interactions were observed for the EMG_{ES} amplitude for the LBX (P = 0.996) or UBX (P = 0.372) training groups. In addition, there were no main mode or time effects for either muscle in each group.

7.3.6 Postural sway on a compliant surface

When standing on a compliant surface a significant training status × mode interaction was observed for COP_{AP} during EO (P = 0.031) and EC (P = 0.003) conditions (Figure 7.11). Post-hoc analyses revealed a significant reduction in COP_{AP} following LBX only. For COP_{ML} a training status × mode interaction was observed during EO (*P* = 0.008) and EC (*P* = 0.047) conditions, indicating an improvement following UBX training only.

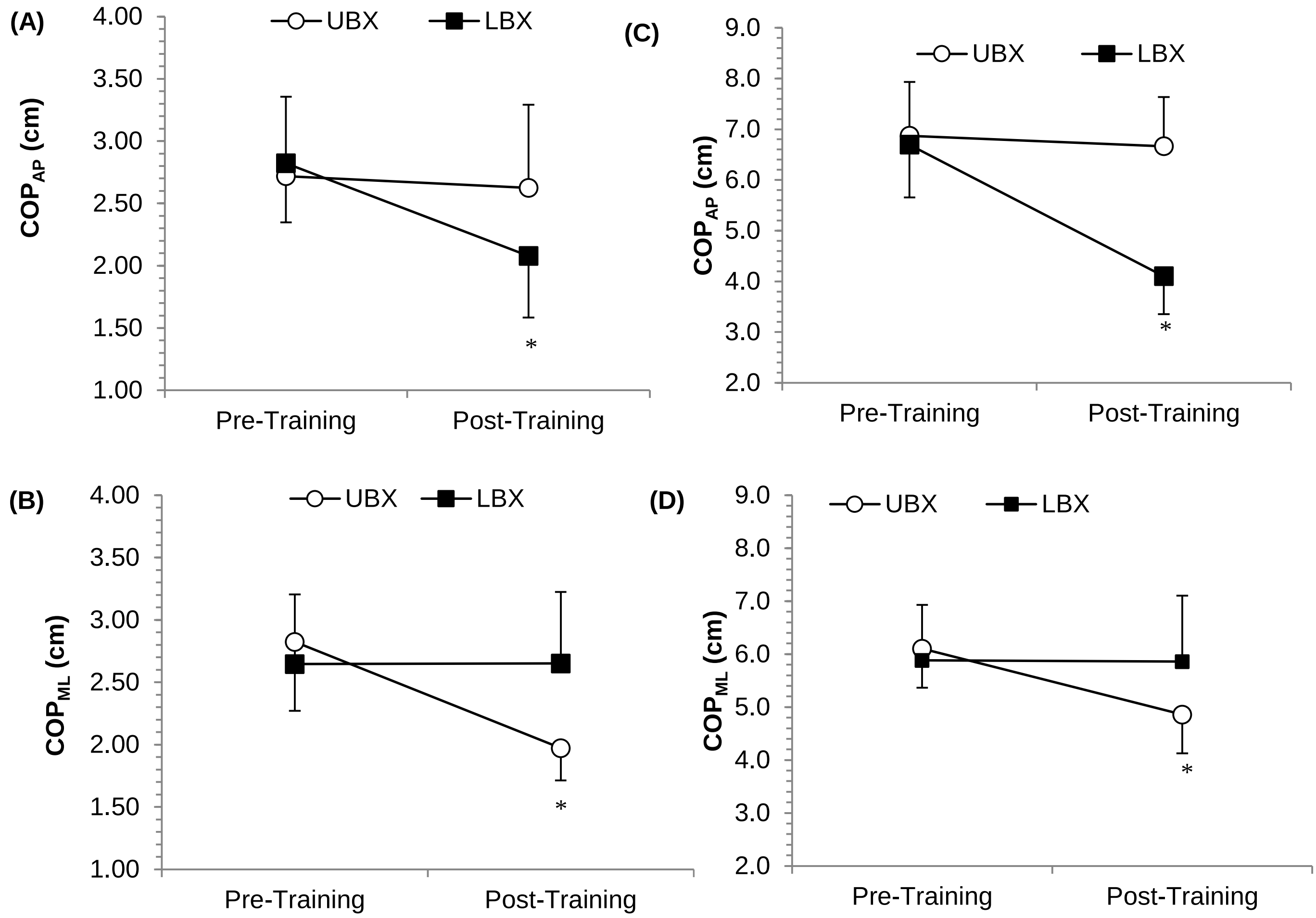
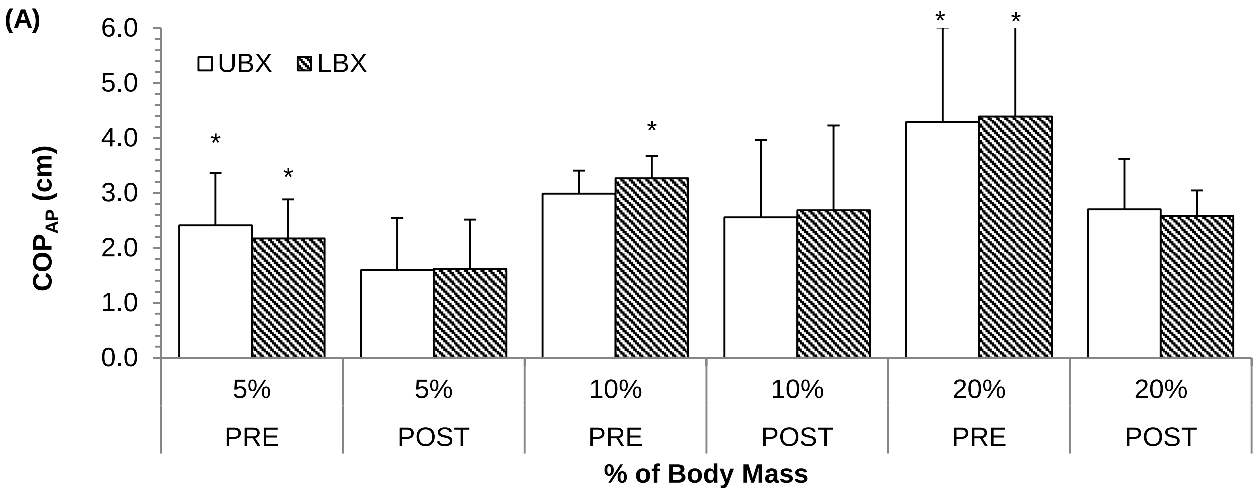


Figure 7.11: COP_{AP} with eyes open (A) and closed (C) and COP_{ML} with eyes open (B) and eyes closed (D) on foam surface before and after upper and lower body exercise training. * *P* $\leq$ 0.05 significant difference with pre training values.

7.3.7 Postural sway while holding a load

For the UBX group COP_{AP} decreased significantly post training when holding a bag at 5% (*P* = 0.024) and 20% (*P* = 0.011) of body mass. The COP_{ML} decreased post UBX when holding a bag at 10% (*P* =0.049) of body mass. Following LBX training, COP_{AP} decreased when holding a bag at 5% (*P* =0.001), 10% (*P* =0.018) and 20% of body mass (*P* =0.048), while COP_{ML} decreased when holding a bag at 5% body mass (*P* = 0.009) (Figure 7.12).



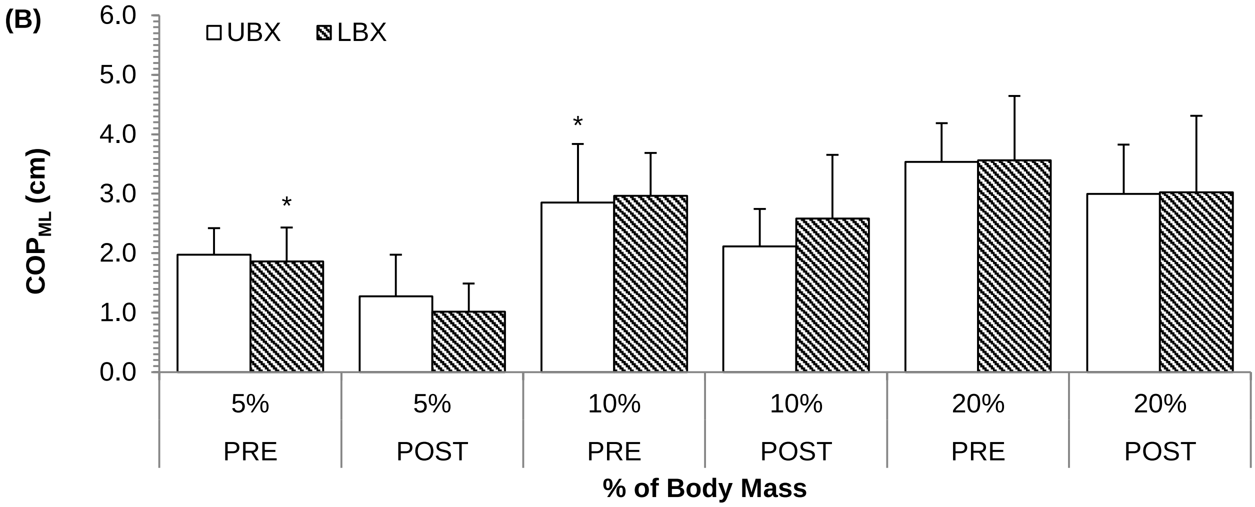


Figure 7.12: Anteroposterior (A) and mediolateral (B) COP displacement while carrying bags at different weights (% of body mass) before and after upper and lower body exercise training

7.3.8 Multi Directional Functional Reach Test

No differences in pre training reach distance were observed between UBX and LBX training groups (*P* $\geq$ 0.05). Functional reach in all directions improved following UBX training (all *P* $\leq$

0.05), but remained unchanged following LBX (all $P \geq 0.05$) (Table 7.6). The greatest improvement in functional reach was observed in the backward direction (Δ 10 ± 5 cm).

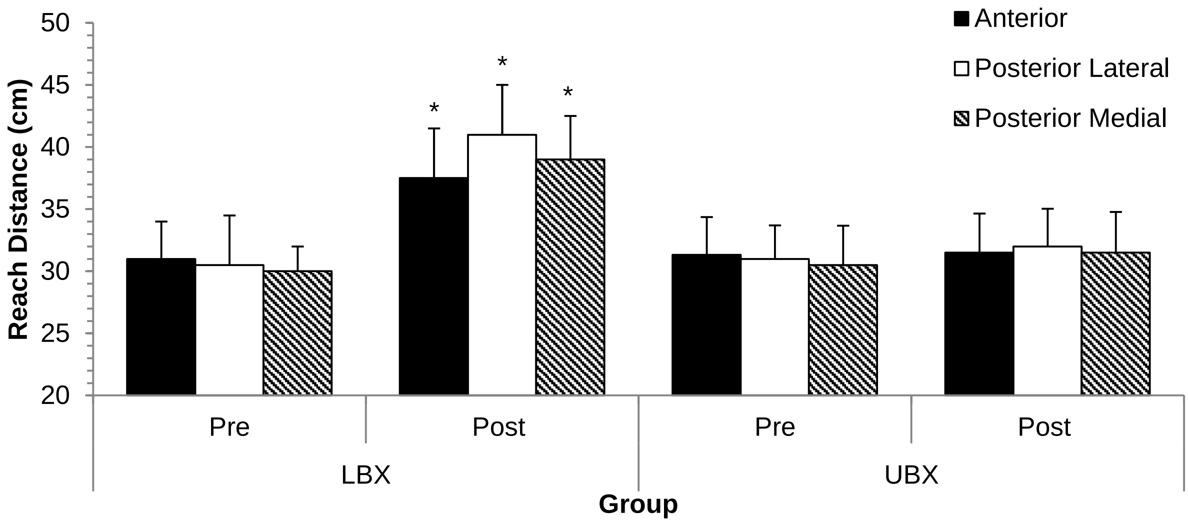
Table 7.6: Functional reach distance before and after UBX and LBX training in four directions

Reach Direction	**UBX**			**LBX**			***P* value (group * training status)**
	Pre	Post	Δ %	Pre	Post	Δ %	
Forward	25 ± 7	33 ± 6*	42 ± 24	26 ± 2	27 ± 3†	4 ± 11	0.017
Backward	17 ± 9	27 ± 9*	60 ± 22	18 ± 4	20 ± 6†	10 ± 13	0.012
Right	22 ± 7	27 ± 6*	27 ± 17	22 ± 4	22 ± 5†	1 ± 7	0.015
Left	22 ± 5	26 ± 5*	18 ± 11	22 ± 5	23 ± 4†	6 ± 9	0.002

* Significant difference between pre and post training; † Significant difference with UBX.

7.3.9 Lower body Dynamic Balance

There was a significant training status * group interaction on anterior (right limb; $P = 0.009$, left limb; $P = 0.001$), posteriormedial (right limb; $P = 0.023$, left limb; $P = 0.033$) and posterolateral (right and left limb; $P = 0.004$) reach directions (Figure 7.13). For all reach directions no significant improvements in the Y balance test were observed after UBX training ($P \geq 0.05$). After 6-weeks of cycling training, anterior reach distance increased by 21 ± 9 and 22 ± 7 % for right and left limb stance, respectively ($P = 0.025$). Greater relative increases in posteriorlateral ($P = 0.045$) (right limb; 36 ± 12 %, left limb; 30 ± 11 %) and posteriormedial ($P = 0.048$) (right limb; 29 ± 11 %, left limb; 28 ± 11 %) were observed compared to the anterior direction.



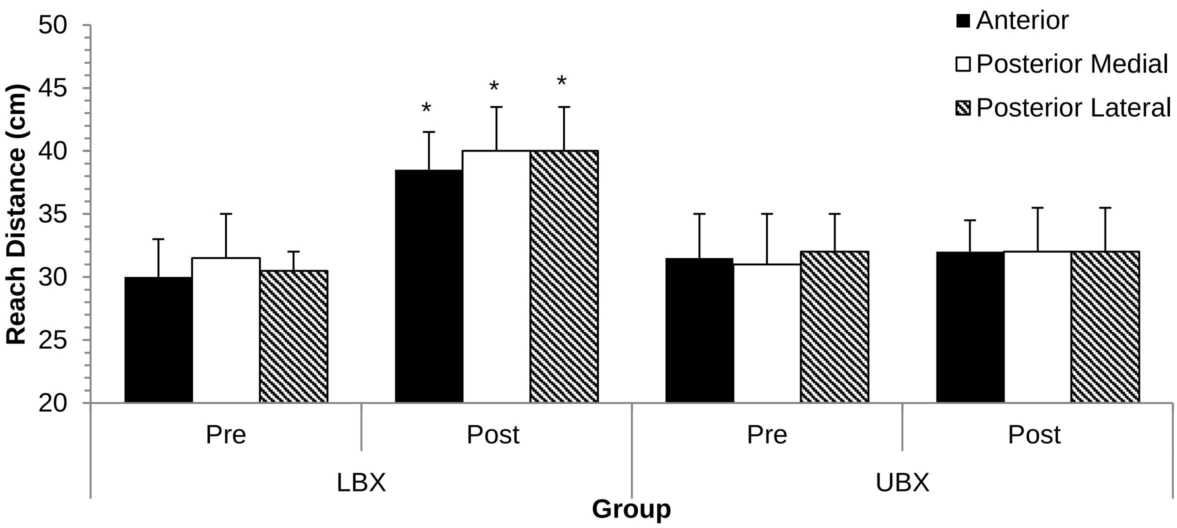
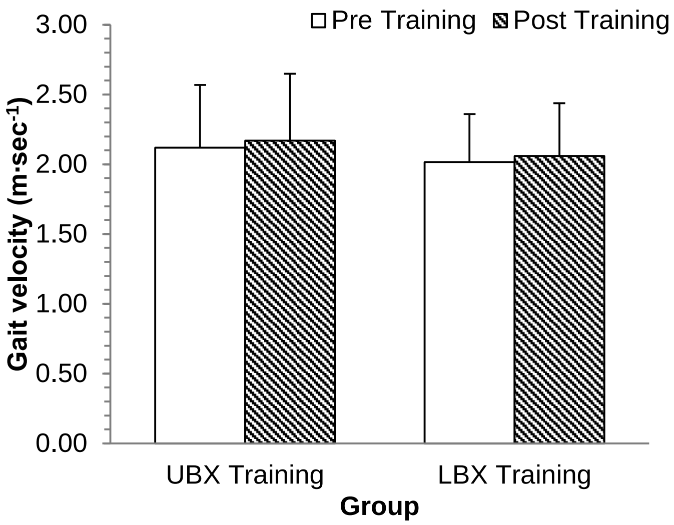


Figure 7.13: Effects of UBX and LBX training on Y balance reach performance for the right (top) and left (bottom) limb stance. * Indicates significant main effect of training ($P \leq 0.05$)

7.3.10 Timed gait speed

A non-significant change in the time taken to walk eight metres was observed for the LBX training group (P = 0.276) and the UBX training group (P = 0.152). The time taken to complete the TUG test significantly decreased in both the LBX (P = 0.001) and the UBX training groups

(P = 0.010). The reduction in TUG speed was not different between LBX (Δ - 9 ± 4 %) and UBX (Δ - 7 ± 6 %) training groups ($P \geq 0.05$) (Figure 7.14).



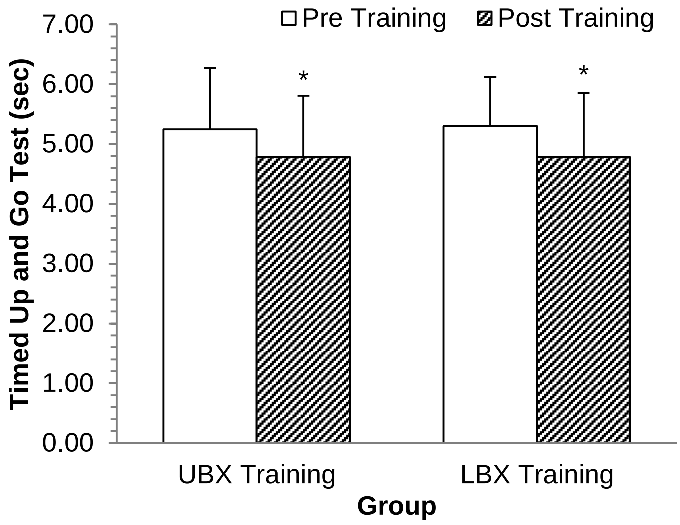


Figure 7.14: Fast walking speed (left) and Timed Up and Go Test (right) before and after 6-weeks of upper and lower body exercise training. * Indicates significant main effect of training ($P \leq 0.05$)

7.3.11 Timed Bridge Test

An improvement in the amount of time spent in the prone bridge position increased significantly from pre to post training for both UBX (P = 0.004) and LBX (P = 0.005) training groups. Post UBX training, the time spent in the prone bridge position increased from 24 ± 12 sec to 49 ± 25 sec (Δ121 ± 72 %). At baseline, prone bridge time was similar in the LBX group (23 ± 18 sec) to the UBX group, however an increase of 9 ± 3 sec post training was not as great (63 ± 43 %), but still significant (P = 0.005).

7.3.12 Grip Strength

There was a significant training status × group interaction for dominant hand grip strength (*P* = 0.001). Following 6-weeks of UBX training, dominant hand grip strength increased by 8 ± 3 kg from 27 ± 8 to 34 ± 9 kg (Δ 31 ± 15 %). In the LBX group, grip strength remained unchanged from pre to post training (28 ± 12 vs. 28 ± 12 kg, respectively). Similarly, a significant group × time interaction was observed for non-dominant hand grip strength (*P* = 0.003). Grip strength increased from pre (27 ± 10 kg) to post-training (33 ± 11 kg) following UBX (Δ 7 ± 4 kg; 26 ± 18 %), but remained the same following CE (28 ± 13 to 29 ± 12, respectively).

7.3.13 Arm Curl Test

A significant group × time interaction was observed for dominant (*P* = 0.001) and non-dominant (*P* = 0.001) arm curls. Following UBX training, the number of arm curls in 30 s increased significantly from pre- to post training for both the right (20 ± 3 to 26 ± 3, respectively) and left (20 ± 2 to 26 ± 3, respectively) arms. Baseline arm curl was not different between UBX and LBX groups (*P* > 0.05). No changes in arm endurance were observed in the LBX group from pre to post training for the dominant (20 ± 4 to 21 ± 5, respectively) and non-dominant arms (19 ± 5 to 21 ± 6, respectively) (*P* > 0.05).

7.3.14 Seated Medicine Ball Throw

A group × time interaction was observed for medicine ball throw distance at both 1.5 kg (*P* = 0.001) and 3.0 kg (*P* = 0.002) weights. Throw distance increased following UBX training, but remained the same following LBX training for both medicine ball weights (Figure 7.15). Interestingly, a greater improvement in throw distance was observed in the heavier ball condition (Δ 32 ± 31 %) compared to the lighter ball condition (Δ 21 ± 15 %) (*P* = 0.041) (Figure 7.12). There was a significant negative correlation between the percentage change in medicine ball throw distance for the 1.5 kg and 3.0 kg weights with the percentage reduction in TUG time in the UBX group (1.5 kg; r = - 0.891; P = 0.005, 3.0 kg; r = - 0.880; P = 0.004). However, no significant correlation was observed between throw distance and fast gait speeds ($P \geq 0.05$).

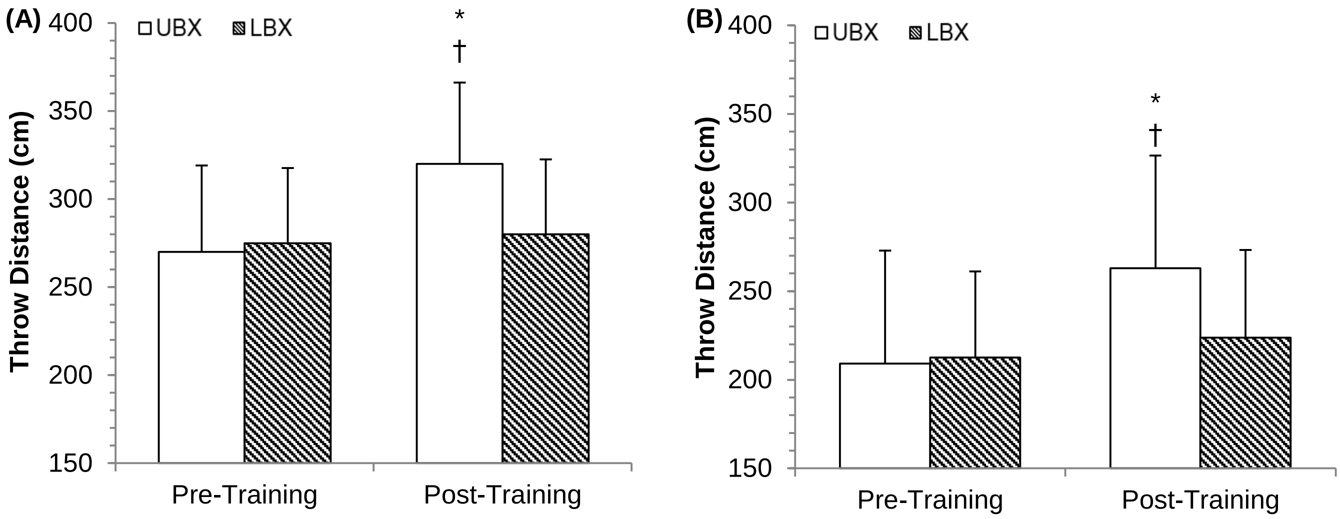


Figure 7.15: Pre and post training values of 1.5 kg throw distance (A) and 3.0 kg throw distance (B) for arm cranking (ACE) and cycle ergometry (CE) groups. Values are mean ± SD; * $P \leq 0.05$ vs. pre training; † $P \leq 0.05$ vs. CE group

There was a significant negative correlation between the percentage increase in EMG_{RA} maximal voluntary contraction with the percentage reduction in COP_{ML} in the UBX group (r = -.864, P = 0.001). In addition, the improvement in the timed prone bridge test was negatively correlated with the reduction in COP_{ML} (r = -.920, P = 0.001) and positively correlated with the improvement in EMG_{RA} (r = .730, P = 0.017) and EMG_{ES} (r = .759, P = 0.011) maximal voluntary isometric contraction.

7.4 Discussion

The purpose of the present study was threefold. Firstly, the specific and cross transfer effects of upper and lower body exercise training on maximal and submaximal exercise capacity in healthy older adults were compared. Secondly, the effects of arm ergometery and cycle ergometery exercise on postural sway were measured before and after upper and lower body endurance training. Thirdly, the efficacy of arm and leg training on a range of balance and functional tasks which would allow real life applications to daily living were determined. The first finding of this study was that 6-weeks of endurance training using either the arms or legs elicited similar improvements in specific (~ 25 %) as well as cross transfer (~ 13 %) $\dot{V}O_{2PEAK}$ during maximal incremental and submaximal exercise tests. These findings indicate that approximately half of the improvements in exercise capacity are central in origin and may be transferable to the untrained muscle mass (e.g., different exercise type). Secondly, this is the first study to report that endurance training mitigates and/or removes the effects of acute exercise on postural sway. Thirdly, during non-fatigued conditions arm training improved upper limb and trunk strength, endurance and stability thus improving functional reach distance. Conversely, leg training increased lower limb strength therefore improving lower body dynamic balance.

7.4.1 Peak physiological responses

When compared to Pogliaghi et al. (2006) lower W_{MAX} and $\dot{V}O_{2PEAK}$ for both ACE and CE modes was observed. The lower peak values reported in the present study likely reflects a lower initial fitness level of the current participants compared to participants by Pogliaghi and colleagues. For example, in the present study lower body $\dot{V}O_{2PEAK}$ ranged between 17 to 22 $ml \cdot min^{-1} \cdot kg^{-1}$, which is considered poor for both females ($\leq$ 25 $ml \cdot min^{-1} \cdot kg^{-1}$) and males ($\leq$ 30 $ml \cdot min^{-1} \cdot kg^{-1}$) over the age of 60 years for lower body exercise (Heyward 2006). However, upon volitional exhaustion of all trials, participants in both UBX and LBX groups attained an

RER greater than 1.15, a HR_{MAX} within 10% of the age predicted maximum and a local/central RPE of at least 18. Therefore, a maximal effort was likely attained. Furthermore, HR_{MAX} reported for ACE and CE in the present study were similar to those as reported by Pogliaghi et al. (2006) using a similar age cohort (67 ± 5 years). The 6-week aerobic exercise training program for both the UBX and LBX groups elicited a significant increase in values of W_{PEAK} and $\dot{V}O_{2PEAK}$ for all incremental tests in the absence of increases in HR_{MAX}. The greater peak values observed post training demonstrates that the training intensity (50 - 70 % W_{PEAK}), frequency (3 days per week) and duration (20 – 35 min) employed in the present study were adequate to elicit an increased exercise tolerance. Specific and cross transfer responses are discussed in the following sections.

7.4.1 i Specific training effects

In the present study 6-weeks of endurance exercise training completed with either the arms or legs elicited similar mode specific exercise improvements of ~ 25 %. These values are in agreement with previous lower body training interventions of a similar (8-weeks; Perini et al. 2002) somewhat longer (12-weeks; Ahmaidi et al. 1998; 28-weeks; Hagberg et al. 1989) and significantly longer durations (52 weeks; Spina et al. 1993) in the elderly. However, this study provides further support of the findings by Pogliaghi et al. (2006) that ACE training elicits a similar increase in $\dot{V}O_{2PEAK}$ to CE training. These improvements in arm ergometry $\dot{V}O_{2PEAK}$ are comparable with those reported in the literature for healthy younger (~ 19 %; Bottoms and Price 2014; ~35%, Lewis et al. 1980; ~16%, Magel et al. 1978; ~19%, Stamford et al. 1978) middle-aged (~16%, Bhambhani, Eriksson and Gomes 1991) and older adults (~22%, Pogliagi et al. 2006). Therefore, the improvements in $\dot{V}O_{2PEAK}$ in the present study are comparable to previous studies for ACE training.

Since muscle mass and potential for maximal power output are different between the arms and legs a different training potential between these modes of exercise might be expected.

When training was performed at the same relative intensity between UBX and LBX groups, there were no differences in the benefits of training on mode specific $\dot{V}O_{2PEAK}$ (~ 26 %). There is conflict in the literature with regards to the specific mechanisms for the improvement in aerobic fitness with limb specific adaptations. Some studies suggest that peripheral changes such as arterial–venous oxygen difference are the predominant cause for mode specific adaptations (Volianitis et al. 2004). Training is limb specific (Loftin et al. 1988; Volianitis et al. 2004) and therefore a significant proportion of the adaptations to endurance training are attributed to peripheral factors to the trained limb alone such as increases in capillarisation (Coggan et al. 1992), conversion of type IIb muscles fibres to type IIa (Coggan et al. 1992), decreases in the activity of some glycolytic enzymes, particularly lactate dehydrogenase (Apple and Roger 1986), increased blood flow (Volianitis et al. 2004) and marked increases in mitochondrial respiratory enzyme levels (Meredith et al. 1989). Recently, Bottoms and Price (2014) reported that peripheral adaptations are evidenced by showing that improvements in W_{MAX} following ACE training were significantly correlated with increased bicep circumference when tensed, therefore hypertrophy was likely to have produced the increased power output. In the present study, an improvement in combined right and left hand grip strength following UBX training was positively correlated with the percentage increase in ACE peak power output. Therefore, the improved maximal exercise capacity for ACE is at least partly attributable to improvements in forearm and hand strength. It remains unknown whether training for a further 6-week period would elicit additional cross transfer benefits or whether improvements would be limited to mode specific adaptations.

7.4.1 II Cross transfer training effects

Previous studies have reported that improvements in exercise performance on a given exercise modality could be transferred to a different exercise mode i.e., a cross transfer effect (Loftin et al. 1988; Pogliaghi et al. 2006; Tordi et al. 2001). The present study reports a significant improvement of LBX training on $\dot{V}O_{2PEAK}$ (~ 12 %) and W_{PEAK} for ACE (+ 11 %).

Furthermore, UBX training elicited an improvement in $\dot{V}O_{2PEAK}$ (~ 12 %) and W_{PEAK} for CE (~ 12 %). The transferability of the training effects has largely been used as evidence of the central nature of the adaptive response (Loftin et al. 1988; Pogliaghi et al. 2006; Tordi et al. 2001). Loftin et al. (1988) previously reported that arm training elicited significant central (stroke volume and cardiac output) and peripheral (a-vO_2 difference) adaptations to support improvements in $\dot{V}O_{2PEAK}$ during both arm and leg work, with the more pronounced effects specific to the upper body. The investigators reported that low initial $\dot{V}O_{2PEAK}$ values during arm and leg exercise may have allowed for potential improvements in both central and peripheral metabolic and circulatory function. In the present study where all of the participants were considered as unfit, both specific and cross transfer benefits were reported for both exercise groups. It is likely that if participants trained for an additional 6-weeks only specific improvements in exercise capacity may be observed due to the improved training status of either the arms or legs.

Some studies have reported no cross transfer effects of ACE training on CE performance in young (Magel et al. 1978; Stamford et al. 1978; Tordi et al., 2001) and middle aged adults (Bhambhani, Eriksson and Gomes 1991). The latter studies suggested that the intensity of arm training (50 – 60 % $\dot{V}O2_{PEAK}$) was not great enough to elicit significant cardiovascular training adaptations in already relatively fit adults (Bhambhani, Eriksson and Gomes 1991; Tordi et al. 2001). In the present study, in agreement with other studies which demonstrated cross transfer effects, the initial fitness level of participants was low (Lewis et al. 1980; Pogliaghi et al. 2006). Therefore, exercise with the arms would likely yield a sufficient cardiovascular challenge to elicit a central adaptation in relatively unfit, but otherwise healthy older adults only. If individuals are not untrained it is likely that they would require a greater exercise intensity. From a practical standpoint, these findings demonstrate that ACE training could provide an effective alternative training stimulus for healthy young and older adults who are initially less fit.

7.4.2 Submaximal physiological responses

The intensity (50 % W_{PEAK}) and duration (20-min) of experimental trials were consistent with exercise intensity prescribed for cardiovascular training in older adults (Nelson et al. 2007), similar to the exertion older adults are exposed to in everyday life (Donath et al. 2013) and comparable with previous studies in this area (Stemplewski et al. 2013). Before and after training $\dot{V}O_2$, $\dot{V}_E$, and HR were lower during ACE compared to CE trials in both groups. This is not surprising since exercise trials were performed at a relative percentage of mode specific W_{PEAK}, which reflects a greater absolute intensity for CE. Pre training, all trials were well matched for RER and RPE_L and RPE_C. As a result, exercise intensity was well matched in terms of the fuel metabolised and perceived exertion during arm and leg exercise.

As expected, 6-weeks of UBX and LBX training elicited significant reductions in $\dot{V}O_2$, $\dot{V}_E$, HR and RER for mode specific exercise at the same absolute intensity which are comparable with the improvements reported by Pogliaghi et al. (2006). These data suggest that specific improvements in exercise capacity are not only observed during maximal, but also submaximal intensities. The present study reports a significant reduction in $\dot{V}O_2$, $\dot{V}_E$, and HR (~ 10 – 13 %) for ACE after LBX training. Similarly, UBX training elicited a reduction – 10 – 12 % reduction in the same variables during CE exercise. As with maximal data, these cross transfer effects were approximately half of the specific exercise effects, as measured by oxygen uptake, providing further evidence that the increase in exercise tolerance is ‘transferable’ to an exercise mode using different muscles.

7.4.3 The effects of ACE and CE on postural sway pre training

Pre training postural sway responses to ACE and CE were similar to the findings reported in previous chapters (Chapter 4 and 5). In both groups, an increase in COP_{AP} and COP_{ML} was observed following CE, suggesting that healthy older adults showed elevated fall risk following exercise engaging the lower body. The destabilising effects of CE on COP measures lasted for 10 and 5-min post exercise for COP_{AP} and COP_{ML} directions, respectively. However, as with our previous findings, ACE did not elicit any post exercise balance impairments. The underlying mechanisms associated with these responses have been described in previous sections and appear to be consistent and reproducible.

7.4.4 The effects of ACE and CE on postural sway following UBX training

7.4.4 I Pre-exercise postural sway

Arm crank ergometry training did not elicit any improvements in COP_{AP} post training (ACE or CE). However, UBX training did elicit a significant reduction in baseline COP_{ML} before ACE (28 ± 20 %) and CE (35 ± 15 %) trials, equating to a reduction of 0.44 ± 0.24 and 0.52 ± 0.27 cm respectively. When compared to norm reference data reported in Chapter 6, the reductions in COP_{ML} represent a reversal of approximately four decades of the age associated decline in COP_{ML} sway (Chapter 6, Figure 6.2). Mediolateral stability is dependent up on movements of the hips and trunk to maintain the centre of mass within the base of support (Winter et al. 1993; 1996). Older adults have poorer trunk repositioning error (difference between target trunk position and intended trunk position from an initial position) compared to healthy young adults (Goldberg, Hernandez and Alexander 2005). Therefore, the ability of the trunk to contribute to the maintenance of upright standing may be comprised among the older population, thus reducing control of COP_{ML} sway. This is particularly important among older adults as they preferentially adopt a hip strategy to maintain balance in the bipedal stance (Woollacott et al. 1986).

There are indications that core strength training can reduce COP_{ML} and mean velocity during quiet bipedal standing (Kaji et al. 2010). It is conceivable that the isometric contraction of the abdominal and spinal musculature during arm cranking (Sawka 1986) improves trunk strength and stability as a result of stabilisation of the upper body during ACE. In the present study, a significant improvement in EMG_{RA} isometric strength (determined by MVIC) was observed post UBX training. A significant negative correlation between the reduction in COP_{ML} during quiet stance and the percentage increase in the MVIC of the EMG_{RA} was also observed. These findings suggest that the abdominal musculature are a major influence in upper body exercise and in the control of COP_{ML} sway. The UBX group also demonstrated a significant improvement in the timed bridge test. Prone bridge exercises have been shown to challenge the quadratus lumborum and abdominal muscles to enhance spine stability (McGill, Childs and Liebenson 1999) suggesting increased lumbar spine stabilisation after UBX training (Schellenberg et al., 2007). Trunk strength and/or stability adaptations are evidenced by the fact that the increase in the time spent in the prone bridge position was also significantly correlated with the reduction in COP_{ML} and also with improvements in EMG_{RA} and EMG_{ES} maximal voluntary contraction.

Collectively, these correlative analyses suggest that UBX training improved trunk musculature strength, stability and endurance which in part is likely to have produced the reduction in COP_{ML} during quiet stance. Such improvements in strength and stabilisation appear to result in smaller COP displacement in directionally sensitive postural muscles which are responsible for mediolateral balance adjustments (Winter et al., 1993, 1996). The improved trunk strength with arm endurance training has important implications for fall risk when considering the association between poor trunk strength and increased incidence of falls among the older population (Granacher et al. 2012). However, LBX training also resulted a decrease in COP_{ML} so therefore not all the improvements were due to the abdominal musculature. This is discussed in more detail in section 7.4.4.

7.4.4 II Post-exercise postural sway

Before 6-weeks of UBX training acute CE elicited an increase in both COP_{AP} and COP_{ML}. Following 6-weeks of UBX training the acute effects of CE on COP_{ML} were completely removed, while acute cycling effects on COP_{AP} were mitigated when post training trials were performed at the same absolute intensity as pre-training trials. It is likely that the effects of CE on postural sway may have still been present to some extent if exercise was performed at the same relative intensity.

Exercise increases energy requirements and therefore augmenting cardiac and respiratory contractions (Bove et al. 2007; Paillard 2012). When respiratory demand remains increased following participation in exercise, COP_{ML} sway is increased as a result of respiration-related changes in trunk muscle activity, due to competition of the respiratory muscles for ventilation and control of sway (Smith et al. 2010). The central origin of training adaptations discussed previously in this chapter may help explain how UBX training was able to remove the acute effects of cycling on COP_{ML}. Specifically, UBX training resulted in cross transfer reductions in $\dot{V}O_2$, $\dot{V}_E$, and HR (~ 10 – 13 %), indicating that cardiac and respiratory contractions were accentuated compared to pre training trials for CE. Therefore, following UBX training CE trials represented a lower relative exercise intensity (50 ± 16 % of pre training $\dot{V}O_{2PEAK}$) compared to pre training trials (61 ± 16 % of pre training $\dot{V}O_{2PEAK}$) thus reducing any competition between respiratory muscles for respiration and balance. Therefore, improvements in trunk strength and stability following UBX training may have reduced the COP_{ML} sway and the central effect of UBX training reduced the respiratory related effects induced by cycling on COP_{ML} as post training trials were performed at a lower percentage of maximal capacity which resulted in an easier intensity.

The central nature of training adaptations may have mitigated the negative effects of lower limb exercise on COP_{AP}. For example, Tew et al. (2009) reported increased tissue O_2

saturation (reflecting the O_2 delivery – utilisation balance) in the calf musculature during submaximal treadmill exercise after ACE training, which may offset peripheral fatigue effects in COP_{AP} sway. Peripheral fatigue of sagittal plane muscles induced by cycling is caused by a decrease in metabolic substrates for muscle contraction (e.g, ATP, PCr and glycogen) as well an increase in metabolites (i.e., lactate and associated hydrogen ions) thus affecting muscle spindle afferents and subsequent disturbance to postural stability predominantly in the COP_{AP} direction (Paillard 2012; Surenkok et al., 2008). Therefore, a better matching between O_2 delivery and O_2 utilisation following arm training likely enhances untrained muscle performance by delaying the accumulation of metabolites that cause muscle fatigue and balance impairment. The cross transfer effects observed in the UBX group on CE may explain the quicker return to baseline sway in the COP_{AP} after arm training (i.e., less stressful intensity).

7.4.5 The effects of ACE and CE on postural sway following LBX training

7.4.5 i Pre-exercise postural sway

After the 6-week intervention, the LBX group showed a marked improvement in COP_{AP} at rest. This finding is in agreement with previous literature demonstrating an improvement in COP_{AP} following cycling training when measured during an un-fatigued resting state (Hassanlouei et al. 2014; Rissel et al., 2013). The observed reduction in COP_{AP} of - 27 ± 10 % (EO) and - 36 ± 17 % (EC) after LBX training represents a reversal of approximately 4 decades of the age associated decline in balance (Study 3, Figure 6.2). Similarly, a reduction in COP_{ML} of - 28 ± 14 % (EO) and -19 ± 10 % (EC) after UBX training also represents a reversal of approximately 4 decades of the age associated decline in balance.

Muscular contributions are an important factor for balance control (Lion et al. 2009). In this regard, muscular strength seems to be related with good postural stability in older adults

(Gauchard et al., 1999). Accordingly, cycling mainly involves lower limb muscle activity in the sagittal plane (e.g., flexors and extensors of the ankles, knees and hips) (Ericson et al., 1985) which are the same muscles responsible for controlling COP_{AP} sway (Winter et al. 1996). The reduced COP_{AP} sway during none-fatigued resting condition after cycling training provide further evidence of the directionally sensitive activity of postural muscles (Vuillerme and Hintzy 2007; Winter et al. 1996). Gauchard et al. (2002) reported that being physically active limits the loss of proprioception, therefore allowing antigravity muscles to detect larger postural sway more quickly and responding with a shorter latency. Furthermore, Perrin et al. (1999) suggested that physically active older adults (running, cycling and swimming) are able to regulate somatosensory inputs more efficiently than sedentary individuals which leads to smaller postural sway values.

The present study reports a significant increase in MVIC of the EMG_{MG}, EMG_{TA}, EMG_{BF} and EMG_{RF} post LBX training. Such findings provide potential evidence that the stimulus induced by cycling training was likely sufficient to elicit neuromuscular changes that are consistent with studies showing increased motor unit recruitment, firing rate and synchronisation following exercise training (Aagaard 2003). Thus, it is likely that improvements in force production of sagittal plane movers reduced COP_{AP} as a function of neural adaptations commonly observed in the first 6 weeks of training (Sale 1992). Bouillon, Sklenka and Ver (2009) reported that improved dynamic postural control after cycling training might be due to adaptations in neural drive, such as increasing fusimotor firing rate, motor-neuron excitability and increased levels of central descending neural pathways and a decrease in neural inhibition. The increase in neural drive during quiet standing may improve the activation of motor neurons of postural muscles and facilitates the integration of afferent information (Nardone et al. 1998) thus translating into better control of postural sway.

7.4.5 ii Post-exercise postural sway

Before 6-weeks of LBX training cycling transiently disturbed COP_{AP} and COP_{ML}. In contrast, LBX training completely removed the effects of CE on COP_{AP} and mitigated the effects on COP_{ML}. Pre and post training CE trials were performed at 62 ± 8 and 44 ± 5 % of pre-training $\dot{V}O_{2PEAK}$ trials. These findings suggest that the post training trial intensity represents an intensity below the threshold where cycling exercise disturbs COP_{AP} in older adults. However, the CE_{ABS} trial in Chapter 4 was performed at 37 ± 6 % of $\dot{V}O_{2PEAK}$ and still elicited an increase in postural sway.

As previously discussed in this thesis acute exercise is associated with impairments in sensory proprioceptive information and/or integration within the CNS and/or a decrease in muscular system efficiency and force production (Derave et al. 2002; Vuillerme and Hintzy 2007). During cycling metabolites released by exercising muscle fibres likely disrupt the control of COP_{AP} (Paillard 2012). At workloads above 65 – 70 % of $\dot{V}O_{2PEAK}$, carbohydrates (primarily muscle glycogen) are the dominant fuel source for exercise (LeBlanc et al. 2004). Indeed, a respiratory exchange ratio (RER) of 0.99 in the pre-training CE trial indicates a significant shift to carbohydrate metabolism and a resultant increased production of carbon dioxide (CO_2). The increase in blood CO_2, resulting in carbonic acid formation and subsequent decreases in blood pH which are also associated with the dissociation of lactic acid into lactate and hydrogen ions (H+) has previous been shown to affect postural control (Surenkok et al. 2008).

Endurance training elicits marked increases in the oxidative capacity of skeletal muscle as a result of an increase in the size and number of mitochondria and an increase in the concentration of enzymes of the Krebs cycle and electron transport chain (Jones and Carter 2000) thus delaying anaerobiosis and subsequent blood lactate concentrations which may increase resistance to fatigue (Gaesser and Poole 1988). A final RER of 0.89 in the post training CE trial suggests a lowered carbohydrate metabolism and thus potentially reducing the adverse effects of metabolic products on postural sway. Therefore, the underlying mechanisms which explain the decreased effects of CE on postural sway post cycling training are likely related to improvements in submaximal exercise tolerance. Post training, values for $\dot{V}O_2$, $\dot{V}_E$, and HR reduced by 15 - 22 %, thus participants were working at a lower relative intensity post training. It is likely that an improvement in muscle endurance due to cycling training may increase the fatigue resistance of the muscle and therefore potentially mitigate or complete remove the destabilising effects of exercise on COP_{AP} sway.

As previously discussed, the increases in COP_{ML} immediately after acute cycling exercise likely reflects respiration-related changes in trunk muscle activity (Smith et al. 2010). Compensatory strategies adopted by the CNS to activate non-fatigued postural muscles following ACE may explain the lack of destabilising effects of upper limb exercise on postural sway. In contrast, following acute CE, fatigued postural muscles may not be able to compensate for the increased respiratory demand thus resulting in increased COP_{ML}. Post LBX training, the destabilising effects of CE on COP_{ML} appear to be mitigated. It is possible that the improvement in submaximal exercise tolerance resulted in less exacerbated respiratory contractions and subsequently a reduced balance impairment. These findings suggest there may be a cross over effect for postural sway in a similar manner for those reported in exercise capacity (Section 7.4.2 and 7.4.3).

7.4.6 Electromyographic responses

7.4.6 i Baseline EMG responses

As a result of the reduced COP_{AP} and COP_{ML} sway after UBX and LBX training, a reduced amplitude of the EMG_{TA} and EMG_{GS} was observed for both groups at baseline. The reduction in EMG amplitude of the calf musculature may suggest that the reduced postural sway post UBX and LBX is clinically beneficial as the body’s centre of gravity was moving over a shorter distance. Therefore, upright balance was more easily maintained, reducing the demands placed on the postural control system and more specifically motor contributions to standing. In addition, following UBX training a reduction in baseline EMG_{RA} was observed, while a reduction in EMG_{RF} was reported following LBX training. These findings suggest that the key muscles for controlling upright stance (GS and TA) were reduced at baseline however, there were differences in the muscles which were altered by UBX and LBX training. The improvement in COP_{ML} sway after UBX training might be explained by a lower amplitude of the EMG_{RA} while reduced COP_{AP} sway after LBX training may be due to the decreased amplitude o the EMG_{RF}.

7.4.6 ii Post exercise EMG responses

Following acute ACE the amplitude of the EMG_{RA} was significantly reduced by ~ 23 % and ~ 8 % during pre and post training trials, respectively. This was accompanied by a significant increase in the amplitude of the EMG_{GS} and EMG_{TA}. Similarly, immediately after CE the amplitude of the EMG_{RF} was reduced by ~ 50 % and ~ 35 % pre and post training, which was also accompanied by an increase in the amplitude of the EMG_{GS} and EMG_{TA}. The reduction in muscle activation of the EMG_{RA} (after ACE) and EMG_{RF} (after CE) returned to baseline relatively quickly ($\leq$ 5 min). An increase in post exercise muscle activity in non-fatigued muscles may partially compensate for the reduction in EMG_{RA} and EMG_{RF} activation in the UBX and LBX groups, respectively (Morris and Allison 2006). Several studies have observed an increased EMG amplitude and earlier onset of activation (anticipatory postural adjustments) for unfatigued muscles, and weaker activation of fatigued muscles (Morris and Allison 2006; Strang and Berg 2007; Strang et al. 2009) following fatigue on non-postural muscles. For example, Kanekar et al., (2008) selectively fatigued the deltoid muscles and reported that increased activation of lower limb muscles (e.g., soleus, gastrocnemius, semitendinosus and biceps femoris) were able to compensate the perturbations to maintain postural stability in bipedal stance. The earlier onset of anticipatory postural activity reported by Kanekar et al. (2008) and the increased amplitude of EMG_{RA} and EMG_{RF} following ACE and CE, respectively, may represent a functional adaptation by the central nervous system to preserve postural stability in the presence of fatigue.

The greater muscle activation of some muscles and weaker activation of others following ACE and CE in the present study may suggest a functional adaptation by the central nervous system to preserve postural stability in the presence of fatigue. Since postural tasks do not maximally activate muscles involved in postural regulation, the CNS can increase the activation of these different muscles to compensate for other fatigued muscles (Paillard 2012; Strang et al. 2009).

A reduced amplitude of the EMG_{RA} and EMG_{RF} was observed post UBX and LBX training, respectively, which might indicate that these muscle became more resistant to fatigue likely due to increased strength and power (Hassanlouei et al. 2014). Endurance training could enhance neural function by reducing response latency and improving the interpretation of sensory information thus improving balance (Orr et al. 2006). Neural adaptations are known to precede morphological adaptations (Moritani and Devries 1979). These neural adaptations include increased neural drive to agonist muscles achieved by increased motor-unit recruitment and increased and earlier firing rate (Enoka 1997) resulting in the ability to exert more force. Therefore, neural adaptations following arm and leg endurance training may result in an improved force control (ability to produce force steadily, thus reducing COP displacement) which might occur due to reduced motor-unit discharge variability (Barry and Carson 2004). Furthermore, the increase in neural drive improves the activation of motor neurons of postural muscles and facilitates the integration of afferent information (Nardone et al. 1998) thus translating into better control of postural sway. An improved control of muscle force may result in a more efficient muscle contraction and therefore a reduction in the EMG amplitude of the muscle during quiet standing.

7.4.7 Postural sway while standing on a complaint surface

When balance was progressively challenged by increasing task difficulty (i.e., standing on a compliant surface with eyes open and closed) COP_{AP} and COP_{ML} increased compared to bipedal standing on a firm surface. It is well known that for such a test age-related differences in postural sway are particularly evident when inputs from two of the three sensory systems are not available (Bisson et al. 2014; Chapter 6). In the present study, UBX training elicited significant reductions in COP_{ML} during EO (- 28 ± 7 %) and EC (- 19 ± 10 %) conditions. In contrast, the LBX group demonstrated reductions of 27 ± 10 % (EO) and 35 ± 16 % (EC) for COP_{AP}. As noted previously, the respective improvements in both groups represented an age related reversal of approximately four decades when considered in relation to data presented in Chapter 6. Therefore, these findings possess significant clinical implications for fall risk in that older adults are better able to compensate for altered sensory information (e.g., closing the eyes and standing on a compliant surface) after an improvement in training status and due to increases in strength. However, neither group were able to improve both COP_{AP} and COP_{ML} postural sway, therefore highlighting the importance of training the arms as well as the legs to elicit a generic overall improvement in balance.

7.4.8 Functional reach

This is the first study to present data pertaining to ACE and functional reach performance. Following ACE training, significant improvements in reach distance were observed in forward (42 %), backwards (60 %), right (27 %) and left (18 %) directions whereas LBX had no beneficial effects. While not directly comparable, previous studies have reported that core strength training results in increased trunk strength and subsequent improvements in reach distance (Granacher et al. 2010). The present study demonstrates improvements in maximal isometric voluntary contraction (MVIC) of the EMG_{RA} (34 %) and EMG_{ES} (33 %) providing some evidence that arm training improved functional reach distance as a result of increased core strength. This is further substantiated by the strong relationship observed between the increase in posterior functional reach and the improvement in the MVIC of the EMG_{RA}. The present study also reports a significant relationship between improvements in prone bridge time and posterior functional reach. An improvement in prone bridge time may suggest increased lumbar spine stabilisation (Schellenberg et al., 2007) and increased endurance capacity of the ES. These correlative analyses indicate that training related improvements in trunk stability and strength could have an effect on postural stability and functional performance following arm training (Granacher et al. 2010).

7.4.9 Dynamic balance

This study found that cycling training increased reach distance expressed as a percentage of leg length in all three directions of the Y balance test, whereas UBX did not. Such improvements in reach distance support similar findings after cycling training in middle aged adults (Bouillon, Sklenka and Ver 2009). These authors speculated that improved knee extensor force production (not directly tested) as a result of cycling training may have accounted for improvements in dynamic balance due to better control of the standing leg. The present study supports these findings by reporting an improvement in the MVIC of the EMG_{RF} and EMG_{BF} muscles following cycling training. Improvements in knee extensor and flexor strength have previously been reported following cycling training in older adults (Macaluso et al. 2003).

Interestingly, Earl and Hertel (2001) found that individuals show greater activity of the BF and medial hamstring during posterior reach directions of the star excursion balance test, whereas the vastus medialis obliquus and vastus lateralis were more active in the anterior reach directions. Therefore, performance of the Y balance results in different lower extremity muscle activation patterns (Earl and Hertel 2001). In the present study, increases in MVIC of the RF were significantly correlated with the increased reach distance in the anterior direction, while increases in MVIC of the BF were significantly correlated with increases in both the posterolateral and posteromedial directions following cycling training. Therefore, the present study provides novel findings that the improvements in anterior and posterior reach direction are independent of improvements in anterior and posterior leg muscles, respectively. Pre-training Y balance performance was the same as similar age groups reported in Chapter 6. However, post LBX training reach distance was ~ 40 cm for all directions which represents an approximate 2 – 3 decade reversal in the age related reductions in dynamic balance performance.

In the present study timed walking tasks were included as measures of mobility performance. This study found that after UBX and LBX training the TUG test was performed faster than pre-training. Interestingly, the improvement in TUG performance was not different between UBX and LBX training groups ($P \geq 0.05$) suggesting that arm and leg training retain a similar potential for improving mobility performance. The improvements in TUG performance after UBX (7 %) and LBX (9 %) training in the present study are consistent with the improvements in TUG performance reported after cycling training in middle-aged adults (Bouillon, Sklenka and Ver 2009). Despite a clear improvement in TUG performance after UBX and LBX training, no improvements in fast gait speed were observed in either group. The TUG test differs from generic walking because it includes rising from a chair, standing, turning and sitting (Viccaro, Perrera and Studenski 2011). While walking requires good dynamic balance ability (Podsiadlo ad Richardson 1991) rising from the chair during the TUG test requires muscle strength and power of the lower limbs and trunk (Viccaro, Perrera and Studenski 2011). These findings suggest that UBX and LBX training may have improved individuals ability to rise quickly from the chair and sit back down without increasing walking speed. Anecdotally, it was noted that participants in both cohorts appeared to reach the initial 3 m marker faster post training, which might support a more rapid rise from the chair and initiation into the gait cycle. It is worthy of mention that the TUG is a timed test and there is a possibility of human error in recording performance time. This is eliminated in the timed gait speed test with the use of photoelectric time gates. Despite the possibility of human error the TUG test was performed using conventional protocols (Podsiadlo and Richardson 1991) and has been replicated by many other studies (Bohannon 2008; Isles et al. 2004). Therefore, the author is confident that the data is reliable and valid.

7.4.10 Upper Body Strength

This study makes an original contribution to ACE research by providing evidence of improved hand grip strength following UBX training. Previous studies have shown that grip strength declines by ~ 20 – 25 % after 60 years of age (Chapter 6; Vianna, Oliveira and Araujo 2007). The results of the present study demonstrate that UBX training can increase grip strength by ~ 25 – 30 % in healthy older adults in this age range. These strength gains therefore represent a reversal of approximately one decade of the age associated decline in hand grip strength (Study 3, Table 6.5). The improvement in hand grip strength following arm training likely reflects the contribution of the hands and forearms when gripping the ergometer handles during the pushing and pulling movements of each duty cycle (Sawka 1986). In the present study ACE training was performed using a neutral grip. Bressel et al. (2001) showed that the neutral grip results in greater activation of the brachioradialis muscle, which are prime muscles for hand grip strength.

While measuring neuromuscular responses during ACE was not within the scope of this study, it was qualitatively noted by several participants that ‘their hands ached quiet significantly’ after each ACE training session in the first three weeks. Interestingly seven of the ten participants stated that they could noticeably feel the training had improved their ability to perform difficult manual duties with the hands. Some of the most notable comments included; opening glass jars and tins, using a trowel in the garden, playing a brass instrument, picking up their grandchild, emptying the washing machine with cold hands, and pouring drinks from the kettle. When asked, the remaining three participants did not feel the training had improved their strength. Interestingly, these three individuals had the greatest hand grip strength prior to training and showed the smallest improvements in strength after training. Therefore, while subjective, the comments were received with some confidence in the context of real life improvements. In addition, upper body power, as measured by the seated medicine ball throw, increased (Harris et al., 2011). According to Harris et al. (2011) the movements during the medicine ball throw and activated musculature are similar to those incorporated in activities of daily living such as rising from a chair, lifting loads and pushing open doors (Harris et al., 2011). This study found a significant negative correlation between the percentage increase in throw distance (combined 1.5 kg and 3.0 kg) with the improvement in TUG test. Therefore, those who had the greatest improvement in upper body power had the greatest reduction in TUG test time, suggesting an important contribution of the trunk and arms during the TUG test. Significant improvement in arm endurance were also observed after UBX training. According to Rikli and Jones (2013) the arm curl test correlates with arm strength and is important for the performance of various daily tasks such as carrying objects, lifting handbags and washing / combing ones hair. Thus the improvement in functional arm endurance as measured by the 30 s arm curl test possesses important implications for older adults when performing everyday tasks.

7.4.11 Summary

In summary, 6-weeks of arm crank ergometry and cycle ergometry training elicited similar improvements in specific and cross transfer exercise tolerance at both maximal and submaximal exercise intensities. Approximately half of the improvements in exercise tolerance were considered as specific to the trained muscle mass suggesting peripheral adaptations to training. The other half of the training adaptation is non-specific since it influences an alternative exercise modality and is probably due to central adaptations. Accordingly, this is the first study to show that endurance training performed with either the arms or legs can mitigate the acute detrimental effects of cycling exercise on postural sway. The data presented in this study is useful for clinicians who may want to incorporate diversity into rehabilitation programs, such as endurance training. Similarly, ACE training offers a novel approach to improving cardiovascular health and aspects of COP_{ML} sway and functional reach performance. In addition, ACE training elicits a range of muscle specific benefits which may be applied to daily life, such as increased hand strength, upper body power, arm endurance and core stability. A combination of upper and lower limb endurance training may be more effective than either form of exercise alone. The favourable adaptations in balance after upper and lower body exercise training may contribute to better aerobic fitness, reduced fall risk and lesser exertion during normal daily activities. In addition, it is likely that exercise training is likely to result in a range of psychological and cognitive benefits which are likely to result in continued physical activity and limiting a loss of independence with advancing age. For some older subgroups with reduced lower limb exercise capacity (e.g., obesity, arthritis, recovering from injury or surgery), ACE training offers a suitable alternative mode of exercise which can contribute to healthier lifestyles and limit the age related physical decline among older adults. This is important because upper body exercise capacity remains well retained in older adults (Aminoff et al. 1997). Therefore, ACE might be a novel way of engaging previously sedentary adults in physical activity.

Chapter 8

General Discussion

The aim of the present thesis was to investigate how acute and chronic upper and lower body exercise affect postural sway in both healthy young and older adults. As outlined in section 1.1, four main objectives were established:

- To examine the effects of upper and lower body exercise effects on postural sway in young healthy adults
- To examine the effects of upper, lower and whole body exercise on postural sway in healthy older adults
- To analyse the differences in postural stability, walking speed and dynamic balance in young, middle aged and elderly adults in order to develop a range of suitable tests which can be used to examine potential improvements in postural stability and functional ability following aerobic training
- To examine the effects of 6-weeks of aerobic exercise training using either the upper or lower body on a range of balance tests as identified in objective 3

This chapter will discuss the main findings of the experimental chapters, addressing each of the aforementioned objectives. Differences in postural sway responses to ACE and CE in healthy young and older adults will be discussed. Consideration will also be given to the adaptations to postural sway and functional abilities following ACE and CE training. Methodological limitations of the four studies will be discussed and subsequent recommendations will be made for future work. Final conclusions will be presented with the practical applications of the present research.

8.1 Main Findings

Study 1 examined the effects of upper and lower body exercise effects on postural sway in healthy young adults. This study involved participants exercising to maximal intensity with subsequent tests involving ACE and CE at the same absolute and relative submaximal exercise intensities. Extensive prior work has documented the adverse effects of both localised fatigue of the lower body (Bizid et al., 2009; Caron 2003, 2004; Corbeil et al. 2003; Harkins et al. 2005; Ledin et al. 2004; Madigan et al. 2006; Paillard et al., 2010; Vuillerme et al. 2002; Yaggie and McGregor, 2002) and lower body endurance exercise (Bove et al. 2007; Burdet and Rougier 2004; Derave et al. 1998; Derave et al. 2002; Gauchard et al. 2002; Mello, Oliveira and Nadal 2010; Nardone et al. 1997, 1998; Thomas, Van Lunen and Morrison 2012; Vuillerme and Hintzy 2007) on postural sway in young adults. In line with previous studies (Gauchard et al. 2002; Mello, Oliveira and Nadal 2010) study 1 reported greater increases in postural sway after maximal compared to submaximal cycling. However, this study was the first to explore the time course effects of cycling on standing balance. It was shown that the adverse effects of maximal and submaximal cycling on postural sway lasted for approximately 15 and 5 min, respectively. It is likely that the destabilising consequences of cycling on postural sway were elicited by a combination of local factors, such as reduced muscle force generating capacity (Enoka and Suart 1992) and proprioceptive alterations (Hiemstra, Lo and Folwer 2001) and central factors, such as increases in respiratory and cardiac contractions (Conforto et al., 2001; Caron et al., 2000).

Study 1 presented further novel findings by addressing an inconsistency in prior evidence by showing that exercise performed with the upper body does not elicit a post exercise balance impairment when performed at the same absolute and relative intensity as cycling. The predominant use of the upper body during ACE (Smith et al. 2008) means that this mode of exercise potentially removes the negative effects of lower limb muscle fatigue on balance. While it is likely that increased ventilation after lower body exercise adversely affects postural sway (Bove et al. 2007), following ACE young healthy adults were able to limit any disturbances to postural stability by potentially re-weighting sensory inputs from visual, vestibular and proprioceptive systems (Paillard 2012; Vuillerme et al. 2006). For example, any deficit of proprioceptive information can be compensated for by increasing the contribution of the visual system. A number of additional compensatory strategies have been identified which reduce the likelihood of a fall following physical activity. For example, earlier anticipatory postural adjustments and increased activation of postural muscles have been reported after upper (Kanekar et al. 2008) and lower (Strang et al. 2008) body fatigue which may allow muscles more time to achieve the required force to maintain stability. Exploratory EMG data recorded from a single participant in study 1 revealed that there was an increase in muscle activation of the gastrocnemius medialis and tibialis anterior following ACE, which potentially reflects a compensatory increase in muscle activation to overcome the negative effects of respiratory demand following ACE. Several studies have provided evidence that upper limb fatigue is compensated by lower body postural muscles (Kanekar et al. 2008; Morris and Allison 2006). This novel finding possesses important new implications as it indicates that young healthy adults may be able to adopt postural strategies to counteract or limit the potential disturbance of postural control caused by upper body exercise. However, this may not be the case in older adults because proprioceptive and neuromuscular systems are less efficient in older compared to younger adults (Bisson et al. 2011) and therefore the negative consequences of exercise might be more exacerbated in older adults.

Despite an increase in fall risk with advancing age, the effects of cycling (Stemlewski et al. 2012; 2013), treadmill walking (Donath et al. 2013) and arm cranking (Smith et al. 2010) on postural sway have been poorly explored. Study 2 sought to expand on the findings in study 1 by further investigating the effects of ACE and CE in elderly adults. The addition of treadmill exercise (TM) was included because some studies had suggested this mode of exercise impaired balance to a greater extent than CE (Lepers et al. 1997; Nardone et al. 1997) and therefore allowed a more comprehensive understanding of the effects of exercise mode on postural sway. While it is known what aspects of postural control are affected by exercise in elderly men (Stemlewski et al. 2012; 2013) no studies to date have reported changes in postural sway following exercise in older females. This is important because the life expectancy of females is greater than males in the UK (Office for National Statistics 2014), therefore there will likely be an increasing number of older female adults who live sedentary lifestyles resulting in increased chronic disease and reduced mobility. Furthermore, approximately 70 % of adults treated for falls in accidents and emergency departments are female (Stevens and Sogolow 2005). As such, females are the most prevalent population undergoing exercise rehabilitation for fall related injuries representing a key target for examining exercise effects of postural sway.

The findings of study 2 showed that ACE does not elicit a post exercise balance impairment in older females, agreeing with previous findings in healthy young males (Hill et al. 2014; Chapter 4). The non-significant change in postural sway after ACE suggests that older adults were able to limit any potential postural sway alterations equally as well as young adults (study 1), possibly by using other available proprioceptive inputs, which agrees with recent findings by Bission et al. (2014). When exercise was performed at the same relative intensity (% W_{MAX}), both between and within subjects, postural sway increased after CE and TM. Interestingly, an increase in COP_{ML} sway was observed following submaximal trials, which is consistent with findings observed in older males after cycling (Stemplewski et al. 2012; 2013). An increase in COP_{ML} sway after exercise has previously been interpreted as a temporary increase in the risk of falling (Egerton et al. 2009). Study 2 also reported that breathing frequency and tidal volume at the end of submaximal trials was correlated with the increase in both COP_{ML} and COP_{AP} sway following CE and TM. These findings suggest that increased breath frequency and depth are likely to have elicited some of the disturbance to postural sway after CE and TM however, the lack of change in sway after ACE suggests that other factors may have contributed to the increase in sway after lower body exercise (i.e., postural muscle fatigue).

To date, no studies have examined the time course effects of such exercises on postural sway in older adults. Addressing the time course effects of exercise on postural sway provided data allowing us to estimate the window of an exercise induced increased risk of falling. Following CE and TM, postural sway measures remained significantly increased up until 10 min post exercise. These findings provide important new implications in the context that the first 10 min following exercise suggest a potential open window of exercise induced elevated risk of falling.

Despite growing interest in the effects of exercise on postural sway in both young and older adults, few studies have directly compared the effects of local muscle fatigue on postural sway in young and older adults (Bission et al. 2014; Davidson et al. 2009; Lin et al. 2009; Parreira et al. 2013). Typically, under quiet standing conditions, localised muscle fatigued elicits a similar absolute increase in postural sway in both young and older adults (Bission et al. 2014). Conversely, the postural sway responses to exercise in study 1 and study 2 do not support the findings in localised fatigue protocols. For example, following 30 min cycling at 50 % W_{MAX} in study 1, small increases in COP_{ML} sway of ~ 20 % (~ 0.2 cm) were reported. In contrast, following 20 min cycling at 50 % HR_E is study 2, COP_{ML} increased by ~ 80 – 120 % (3 – 4 cm). Therefore, both absolute and relative changes in COP_{ML} following CE were greater in older adults. In contrast, following 30 min cycling in study 1, increases in COP_{AP} sway of ~ 70 – 100 % were observed, compared to increases of ~ 50 – 80 % in older adults (Figure 8.1). Despite the smaller relative percentage increase in COP_{AP} sway, the absolute increase in sway was greater in older (~ 5 cm) compared to younger (~ 1 cm) adults. Therefore, it appears that lower body exercise elicits adverse effects mainly on COP_{AP} sway in young adults, while eliciting a more overall adverse effect in the COP_{AP} and COP_{ML} in older adults. The increase in COP_{ML} sway may be practically significant to fall risk as mediolateral aspects of balance are related to fall risk among older adults (Maki, Holliday and Troppe 1994; Piirtola and Era 2006).

The underlying mechanisms explaining the difficulty in maintaining COP_{ML} sway in older adults, especially after exercise, remain unclear. Older adults tend to adopt a hip strategy to maintain balance (Woollacott and Shumway Cook 1986). Specifically, mediolateral balance is primarily dependent upon movement of the hip and trunk to maintain the centre of mass within the base of support (Winter et al. 1996). Because respiratory demand is increased after exercise the contribution of the trunk musculature to control balance may be impaired in older adults.

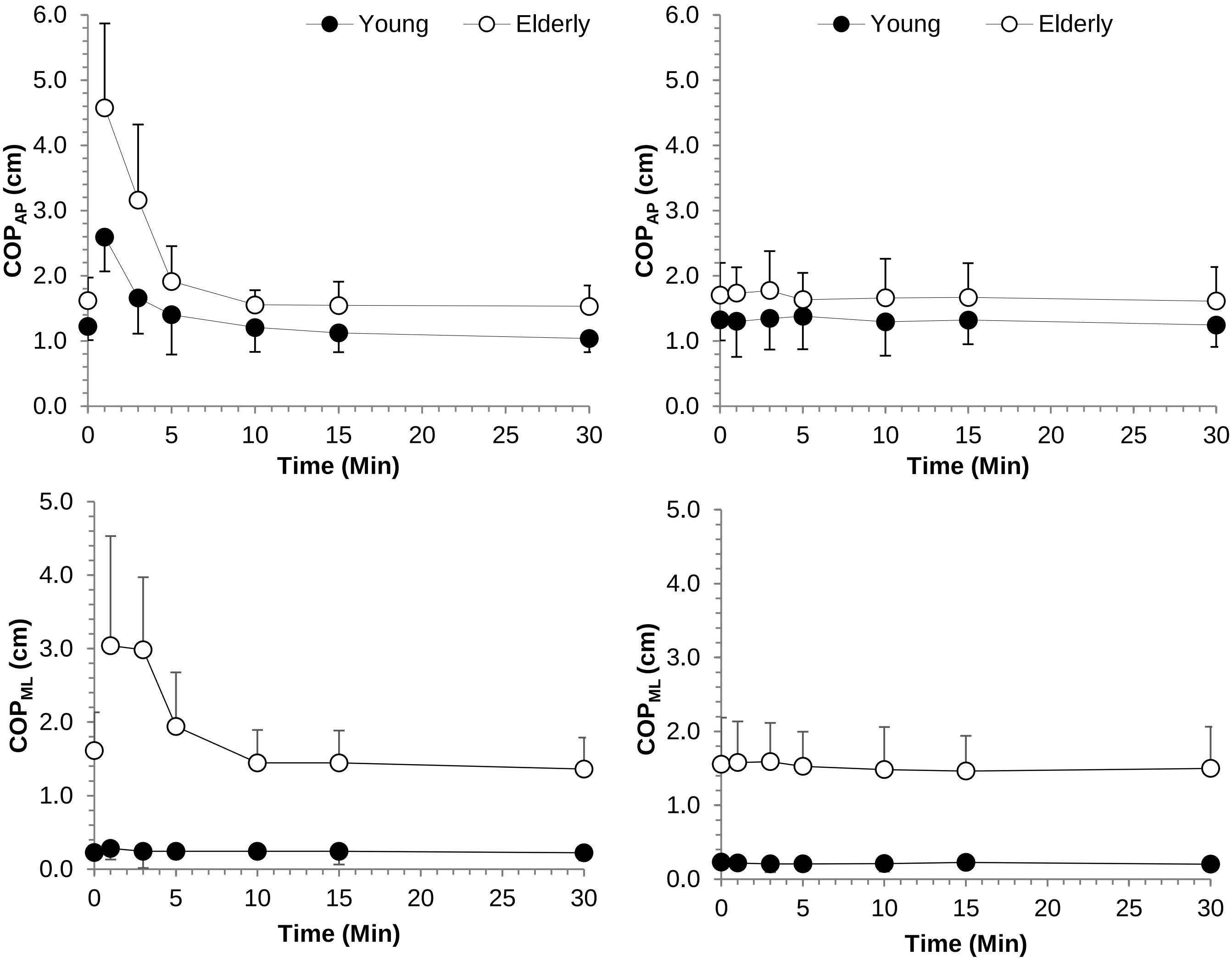


Figure 8.1: Mean ± SD COP_{AP} and COP_{ML} pre and post upper (right) and lower (left) body exercise recorded in study 1 (30-min, 50 % W_{MAX}) and 2 (20-min, 50 % HR_E)

Several potential mechanisms may explain the greater disturbance to postural sway following exercise at the same relative intensity in older compared to younger adults. Ageing results in neural and musculoskeletal changes that lead to muscle weakness, poor muscular endurance, sensorimotor deficits and subsequently postural control impairment (Orr 2010). Factors regulating standing balance are multifaceted (Parreira et al. 2013) and therefore exercise which impacts upon any of the aforementioned postural control systems may have more adverse effects on older compared to younger adults. Extensive work suggests that older adults are more fatigue resistant than younger adults (Hakinen 1995; Kent-Braun et al. 2002; Rubinstein and Kamen 2005) due to an increase in type I fibres and decrease in type II fibres with age (Nikolic et al. 2010), a shorter fascicle length and lower ratio of active muscle volume and force production (which might be an energetic advantage) (Mademli and Arampatzis 2008) and improved metabolic economy as evidenced by less metabolite accumulation (Lanza, Larsen and Kent-Braun 2007). Therefore, based on these findings one might expect the effects of exercise on postural sway to be less severe in older compared to younger adults. It is likely that the greater adverse effects on postural sway in healthy older compared to younger adults observed in study 1 and study 2 reflects habituation to exercise. Specifically, none of the participants in study 2 had walked on a treadmill or exercised on a stationary cycle ergometer or arm crank ergometer. Despite being familiarised to the exercise apparatus all exercise modes were novel to the older adults.

While the effects of cycling on postural sway were generally greater in older compared to younger adults the recovery times were similar in both cohorts. To the author's knowledge, only one study investigated postural sway recovery from fatigue in older compared to younger adults (Lin et al. 2009). It was reported that COP_V recovered after 11 and 2 minutes for young and older adults, respectively. Lin et al. (2009) cited increased fatigue resistance as the predominant explanation for quicker recovery times following fatigue in older adults. The author is not aware of any studies which have investigated the recovery time following endurance exercise in young and older adults. Bove et al. (2007) reported a linear relationship between the recovery in COP_L and oxygen consumption post exercise. These findings suggest that the magnitude and recovery after exercise could be related to the rapid recovery of oxygen uptake post exercise. Indeed, chapter 5 showed that breathing frequency and tidal volume recorded in the final 30 s of exercise was significantly correlated to the percentage increase in COP_{ML} and COP_{AP}, but only after CE and TM trials. The non-significant correlation between respiratory variables at the end of ACE and changes in postural sway provide strong evidence that increased ventilatory responses elicit adverse effects on balance after lower body exercise, but other factors must be contributing to the disturbance.

When all CE trials were considered a significant correlation was observed between the percentage change in COP_{AP} sway from baseline to immediately post exercise with $\dot{V}O_2$ expressed as a percentage of $\dot{V}O_{2PEAK}$ (r = 0.620, P = 0.0001) (Figure 8.2). Therefore, the disturbance to postural sway post exercise is at least partly explained by the intensity of exercise. Distinct thresholds were not evident, which likely reflects the multifactorial nature of postural stability. These findings suggest that working at more severe intensities relative to individual's maximum is likely to result in greater balance disturbances and thus increased likelihood of falling.

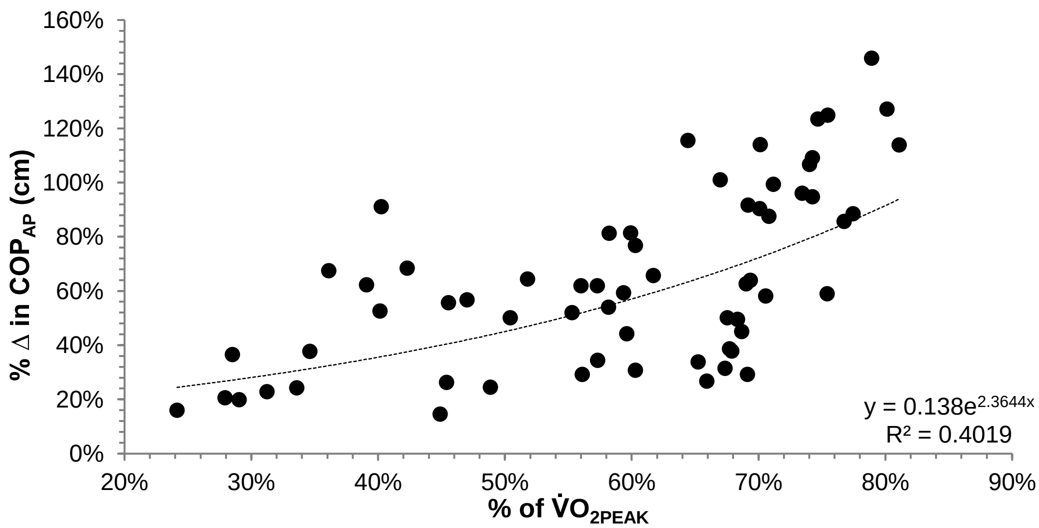


Figure 8.2: Exercise intensity relative to individuals maximum and percentage change in COP_{AP} sway following cycling exercise

Study 3 aimed to determine whether postural sway could predict functional performance with the view to increasing the clinical relevance of COP measures with regards to balance impairment. This study aimed to determine the posturographic and functional balance measures that would yield the best discrimination between age groups. This study makes an original contribution to balance research by addressing an important gap in the published literature with regards to effects of aging on balance control. Firstly, study 3 provided data to overcome equivocal findings in the literature (Nolan et al. 2010; Iling et al. 2010) by showing that performance of functional balance tests, such as timed walking speed, functional reach and dynamic stability begin to decline at 50 years of age, which was 10 years earlier than the significant increases in postural sway were evident ($\geq$ 60 years). The reported changes in functional balance occurred at an age younger than usually focused on for the testing of balance and introduction of pre-emptive exercise interventions to improve balance (Isles et al. 2004). This is important because functional tests might provide an earlier warning sign identifying individuals who present an increase in fall risk. This study was also the first to report data for the SEBT among older adults. The finding that SEBT scores decreased with age may suggest that this test discriminates age related changes in balance equally as well as other functional (i.e., MDFR, and gait speed) and quantitative (i.e., posturographic) assessments.

As expected, postural sway showed weak to moderate correlations with functional outcome measures, suggesting that postural sway is not a useful tool to predict success in functional performance tests. Therefore, it was concluded that functional tests do not necessarily furnish the same information regarding balance mechanisms as centre of pressure data derived from a force platform. As a result, it was acknowledged that future interventions should consider a more comprehensive range of balance tasks which more thoroughly investigate changes in balance. This study also showed that age was a significant factor for all outcome measures with poorer performance being associated with older age. The results suggested that the measures considered were sensitive enough to observe age related changes and, as a result, might be modifiable by training.

The purpose of study 4 was threefold. Firstly, to determine the benefits of arm and leg exercise training on specific and cross transfer exercise tolerance in older adults. Secondly, to examine the effects of upper or lower body exercise training on the experimental procedures used in study 1 & 2. Thirdly, to determine the effects of upper and lower body exercise training on a range of functional outcome measures which were identified from study 3.

Growing evidence suggests that endurance training interventions can enhance postural control under static and dynamic conditions (Buchner et al. 1997; Bouillon, Sklenka and Ver 2009; Donath et al. 2014; Due Jakobsen et al. 2011; Rissel et al. 2013; Vilarinho et al. 2009). However, these studies are limited as postural stability was measured in non-fatigued resting conditions. Study 4 provides new insights into the effects of exercise training on balance by measuring postural sway before and after acute exercise trials, pre and post training. The results of this study showed that both LBX and UBX training reduced and/or completely removed the negative effects of acute cycling on postural sway when exercise was performed at the same absolute intensity pre and post training. The impact of this work is important as it indicates that while acute lower body exercise elicits adverse effects on standing balance, the deleterious consequences of cycling to balance control can be removed with an improvement in training status. Importantly, UBX and LBX training improved different aspects of balance. For example, UBX training improves balance tasks where the trunk musculature contributes significantly to balance adjustments (i.e., standing on a compliant surface) or tasks (i.e., functional reach distance). Conversely, LBX training improved balance tasks which are dependent upon the contribution of lower limb muscles, such as quiet standing on a fixed surface and dynamic lower body reach distance.

Study 4 also presents novel data in terms of compensatory postural strategies as measured by EMG. Numerous studies have reported that following local muscle fatigue, non-fatigued postural muscles are activated earlier (Strang and Berg 2007; Strang et al. 2009), at a greater amplitude (Morris and Allison, 2006) and for longer (Strang et al. 2007), while fatigued muscles show a weaker activation (Morris and Allison, 2006). This is the first study to show similar responses following endurance exercise. Following acute ACE, a smaller amplitude was observed for rectus abdominus with a greater activation of the gastrocnemius medialis and tibialis anterior. This might suggest that the central nervous system adopts a local postural compensatory strategy by increasing the activation of postural muscles which were not fatigued. Indeed, this is further substantiated by findings that the amplitude of the rectus abdominus decreased significant following acute CE, which was accompanied by an increase in the amplitude of the EMG_{GS} and EMG_{TA}, in a similar manner to the responses following ACE. This is an important finding which suggests that older adults are able to effectively use different compensatory postural strategies to counteract or limit the disturbance of postural control elicited by endurance exercise (Paillard 2012). These findings contribute to a better understanding of the feed forward mechanisms of postural control with important implications for elderly individuals. Determining the impact of exercise on protective compensatory strategies may provide a new possibility for the development of rehabilitation strategies that can result in adaptations at the peripheral muscle level while also modifying central protective strategies.

In young adults it is still debated whether training with a specific muscle group can also enhance exercise performed with a different muscle group (cross transfer effect) (Bhambhani et al. 1991; Lewis et al. 1980; Loftin et al. 1988; Tordi et al. 2001). Available data in this area is scarce in healthy older adults. Study 4 provided supporting evidence to earlier findings (Pogliaghi et al. 2006) that training modalities bought about by the arms and legs elicited similar improvements in specific (24 - 26 %) and cross transfer (10 - 12 %) effects in older adults. Therefore, for older adults who are relatively unfit, or those with lower limb disability, ACE is an alternative mode of exercise which can elicit significant health and fitness benefits. These findings supplement the novel data pertaining to the directionally sensitive changes in postural muscles and support the notion that, for healthy older adults, training the arms as well as the legs will provide functional benefit with the likely attenuation of postural sway in addition to improving general fitness and real life exercise capacity.

8.2 Limitations and Future Research

Despite all methods being thoroughly planned according to limitations noted within the literature, a number of methodological limitations are acknowledged and future research should consider these observations.

It is acknowledged that the generalisability of the results in the current thesis to an at risk population are limited due to the healthy cohorts recruited. Although adults over the age of 60 are at increased risk of falling, none in the current thesis were balance impaired, as validated by the Berg Balance Scale (BBS). Whether the present effects would be observed in individuals with impaired postural control in individuals with typically lower fitness levels (i.e., Parkinsons's disease, stroke or peripheral arterial disease) where the negative consequences of exercise may be more exacerbated remains to be investigated. This thesis aimed to determine the effects of UBX and postural sway effects in healthy cohorts, with the effects in clinical patients aimed at future work. Further cross-sectional and prospective studies are required in those prone to falls and clinical populations as it seems likely that older adults with comorbidities may experience a greater alteration of postural sway following acute submaximal exercise.

While postural sway during quiet bipedal stance has been shown to be robust and sensitive enough to detect the effects of lower limb exercise in both young (Study 1) (Gouchard et al., 2002; Mello, de Oliveira, & Nadal, 2010) and older adults (Study 2 & 4; Stemplewski et al. 2012; 2013; Egerton, Brauer & Cresswell 2009) this task does not represent all aspects of balance and fall risk factors. The experimental design used a simple task (bipedal stance) which participants could easily and safely perform. The effects of ACE and CE on other more challenging tasks (i.e., standing on foam) might elicit different findings. However, the adoption of this stance was necessary as this task was that examined in many previous studies. Therefore, the present results are comparable to those of previously published results.

It might be questioned whether the increases in postural sway observed post exercise were clinically significant to fall risk (Derave et al. 1998). The increases in postural sway observed in the present thesis and in previous literature suggest that the COP was still well within the support surface of the feet therefore the increase in sway was too little to elicit a fall (Derave et al., 2002). Therefore, the practical applications of the present findings are based on the interpretation that an increase in postural sway post exercise indicates an increased risk of falling. Nevertheless, postural sway during quiet bipedal standing can be improved with training.

Several authors have recently cited that an increase in fall risk post exercise might be evident through other measures. For example, Egerton, Brauer & Cresswell (2009) and Mello, de Oliviera & Nadal (2010) both suggest that prolonged feelings of tiredness might affect attention, thus increasing fall risk. It is well known that postural control requires attentional resources (Brauer, Woollacott & Shumway-Cook 2001), and these resources are reduced when fatigued (Vuillerme, Forestier & Nougier 2002). Therefore, the possible interaction between fatigue-induced reductions in attentional resources and postural sway warrant further investigation. Currently, there appears to be a shift in research focus to studies which are investigating the effects of fatigue on postural sway while completing a dual task. At present, only localised muscle fatigue protocols have been investigated (Bisson et al. 2011).

Post exercise postural sway assessment was carried out rapidly after exercise to ensure any changes following activity were captured. This meant that we could not examine other tasks such as functional reach distance, or gait velocity which might have improved the clinical relevance of the first two studies. However, by adopting a pre and post exercise comparison, this task still allowed us to investigate changes in postural sway, which was the primary aim of the first two studies.

As already discussed acute ACE did not elicit a post exercise increase in postural sway, which might be linked to the relationship between respiratory activity and the standing task. For example, ACE disturbs postural sway in patients with chronic obstructive pulmonary disease, but not in healthy older controls (Smith et al., 2010). Increased activity of the trunk musculature and diaphragm enhances trunk stiffness, thus reducing the contribution of the trunk muscle movements for balance control. While previous research has reported a strong relationship between recovery of post exercise oxygen consumption and the recovery of the COP path length following treadmill running (Bove et al., 2007) the present thesis did not measure any post exercise gas variables, namely $\dot{V}O_2$ and $\dot{V}_E$. The recovery process from fatigue is often considered to be the biggest limitation for all fatigue experiments, therefore we felt this was not a practical approach for ensuring that the time between the end of exercise and the first sway trial was kept within a maximum of 30 s.

Similar to previous studies, this thesis documented the effects of exercise on conventional COP sway measures, without reporting other complementary measures such as joint stiffness and proprioception. Interpretation of findings may have benefited from these measures allowing for a more complete conclusion of the findings. However, Study 1 and Study 4 did investigate muscle activation before and after exercise thus allowing for a better understanding of the compensatory strategies elicited in responses to acute exercise.

Due to dropouts randomisation was not perfect with regards to gender distribution, body mass and baseline physical activity in Study 4. As a result, there were relatively more males in the UBX training group (n = 4) compared to the LBX training group (n = 2). However, when males and females were compared, there were no differences in initial aerobic fitness or the improvements in postural sway and exercise tolerance.

While the data from study 4 provided strong evidence of a positive adaptation in balance and other functional abilities, the present findings do not allow conclusions to be made with regards to absolute fall risk. Using retrospective and prospective experimental designs, some studies have shown that aerobic endurance training improves falls incidence, but not indices of balance (i.e., postural sway, timed gait, functional reach) (Buchner et al., 1997). Therefore, it is likely that exercise training can reduce fall risk by mechanisms other than affecting gait and balance, such as behavioural (i.e., walking on ice), or psychological changes (i.e., fear of falling). Future training studies should consider a prospective research design to determine future fall rates, in addition to behaviour and psychological questionnaires to more fully determine the effects of exercise training and associated benefits on fall risk. However, what is clear is that UBX training improves more components likely to contribute to balance in untrained individuals than LBX training.

In study 4, pre and post training experimental trials were performed at the same absolute intensity in order to determine the absolute changes in physiological and postural sway responses to exercise. It could be argued that significant improvements would intuitively be observed due to post exercise trials representing a lower relative percentage of post training peak power output. Therefore, it would have been advantageous to also perform experimental trials at the same relative intensity as post training peak power output.

The training intervention in this thesis elucidates to the potential importance of the abdominal musculature for ACE. It would be of interest to acutely fatigue the torso musculature and determine its effects on subsequent ACE performance and post exercise balance impairment. Furthermore, it is of keen interest to investigate the effects of abdominal training on ACE performance and generic balance adaptations.

The extensive literature documenting the effects of exercise on postural sway advocate that individuals should exercise caution following lower body exercise. The time has come to broaden our understanding to implement preventative strategies to remove the adverse effects of exercise on postural sway. Specifically, future work should focus upon the intensity and duration of a warm down required to remove the effects of lower body exercise on balance impairment. Given the novel findings that ACE does not disrupt sensorimotor control during quiet standing, it seems plausible to explore the effects of multidimensional exercise protocols involving cycling and arm cranking on postural sway.

The cross transfer training adaptations following ACE training remains an area of inconsistency within the literature. The current understanding is that cross transfer benefits are limited to individuals who are inherently unfit before training. In order to address this concern more closely, it is recommended that older adults engage in 12-weeks of ACE training, where fitness would be assessed at baseline, 6-weeks and 12-weeks. It remains to be determined whether further cross transfer adaptations are observed after the initial 6-weeks when the upper body muscle mass is trained.

8.3 Practical applications

This thesis provides an original contribution to the literature through the finding that the adverse effects of lower body exercise on postural sway are not present following upper body exercise in healthy young or older adults. These findings imply that upper limb exercise does not result in major alterations in sensorimotor control mechanisms responsible for standing balance in healthy adults. This original contribution has several practical applications which are of use in a range of settings including exercise scientists, community healthcare providers, physiotherapists and those working in rehabilitation settings. The novel finding that ACE has no impact upon quiet bipedal standing challenges current precepts for the relationship between exercise and postural sway, which will enable the potential risks of injury and falls during exercise training to managed and anticipated during exercise rehabilitation. Upper and lower body endurance training reduces and/or completely removes the deleterious effects of lower body exercise on postural sway in older adults. A combination of upper and lower body exercise is more effective in older adults than either mode of exercise alone and should therefore be utilised by practitioners prescribing exercise in older adults. This novel information will be of interest for clinicians who may want to incorporate variety into rehabilitation training, such as the use of aerobic exercise. A specific focus within rehabilitation after fall related injuries should be directed towards endurance training with the arms, as this mode of exercise is safe and effective and can easily be modified and implemented by clinicians and nurses for fall risk populations. As upper body exercise training possesses a range of functional and health benefits, this mode of exercise might be ideally suited to a number of clinical populations who have reduced lower body exercise capacity to allow these individuals to get back on their feet and engage in complementary lower body exercise.

Age-related musculoskeletal functional decline includes sarcopenia and reduced balance, limiting the capacity to perform daily tasks and subsequently increasing fall risk and accelerating the progression of disability and a loss of independence. While therapeutic activities attempt to mitigate this decline, current practices typically have poor adherence, potentially due to difficulty in engaging in conventional exercises. For example, some subgroups with lower limb diseases (e.g., peripheral arterial disease, diabetes induced foot ulcerations, arthritis etc.), obese individuals, those recovering from lower limb surgery and/or injury and elderly who are very sedentary typically cannot tolerate lower limb exercise resulting in poor attrition and limited benefit. However, chair based ACE training may be more suitable for older individuals as fall risk is reduced and seems to be well tolerated by a range of clinical groups. The current research project(s) has generated new information and has confirmed the efficacy and feasibility of upper limb training to improve fitness and functional outcomes among older populations.

As life expectancy continues to rise in the UK, there will likely be a corresponding increase in the number of people falling. Therefore, the need to develop more effective and efficient therapies, which target a range of impairments in a single session (e.g., fitness, strength, balance, functional ability etc.) is essential, particularly as poor adherence with current therapeutic practices results in largely ineffective outcomes in older populations. The primary finding that seated arm training results in improved balance outcomes in addition to the known benefits on reversing losses in muscles mass and fitness, might be influential in providing guidelines for planners and policy makers as to the most effective therapies to target fall risk in older adults. In the long term these guidelines may indirectly enhance quality of life and prolong independent living in older adults while simultaneously reducing time demands placed on clinicians and patients and subsequently reducing the financial burden on the state.

References

Aagaard, P. (2003). Training-induced changes in neural function. *Exercise and Sport Sciences Reviews*, 31 (2), 61-67.

Ahmaidi, S., Masse-Biron, J., Adam, B., Choquet, D., Freville, M., Libert, J. P., & Prefaut, C. (1998). Effects of interval training at the ventilatory threshold on clinical and cardiorespiratory responses in elderly humans. *European Journal of Applied Physiology and Occupational Physiology*, 78 (2), 170-176.

Ahlborg, G. & Jensen-Urstad, M. (1991). Metabolism in exercising arm vs. leg muscle. *Clinical Physiology*, 11, 459-468.

Adlerton, A.K., Moritz, U., & Moe-Nilssen, R. (2003). Forceplate and accelerometer measures for evaluating the effect of muscle fatigue on postural control during one-legged stance. *Physiotherapy Research International*, 8 (4), 187-199.

Allen, D. G., Lamb, G. D., & Westerblad, H. (2008). Skeletal muscle fatigue: cellular mechanisms. *Physiological reviews*, 88 (1), 287-332.

Allen, T. J., Leung, M., & Proske, U. (2010). The effect of fatigue from exercise on human limb position sense. *The Journal of Physiology*, *588* (8), 1369-1377

Allum, J.H.J., Carpenter, M.G., Honegger, F., Adkin, A.L., & Bloem, B. R. (2002). Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man. *The Journal of Physiology*, 542 (2), 643-663.

Aminoff, T., Smolander, J., Korhonen, O., & Louhevaara, V. (1996). Physical work capacity in dynamic exercise with differing muscle masses in healthy young and older men. *European Journal of Applied Physiology and Occupational Physiology*, 73 (1-2), 180-185.

Aminoff, T., Smolander, J., Korhonen, O., & Louhevaara, V. (1997). Cardiorespiratory and subjective responses to prolonged arm and leg exercise in healthy young and older men. *Journal of Applied Physiology and Occupational Physiology*, 75 (4), 363-368.

Andersson, B., Jonsson, B., and Ortengren, R. (1974). Myoelectric activity in individual lumbar erector spinae muscles in sitting: A study with surface and wire electrodes. *Scandinavian Journal of Rehabilitation Medicine*, 3, 91–108.

Apple, F.S., & Rogers, M.A. (1986). Skeletal muscle lactate dehydrogenase isozyme alterations in men and women marathon runners. *Women*, 14 (8.7), 9-9.

Astrand, P.O. & Rodahl, K. (1986). *Textbook of Work Physiology*. 3rd. New York, McGraw Hill International Editions.

Astrand, P.O., Ekblom, B., Messin, R., Saltin, B. & Stenberg, J. (1965). Intra-arterial blood pressure during exercise with different muscle groups. *Journal of Applied Physiology*, 20, 253-256.

Bar-or, O. & Zwiren, L.D. (1975). Maximal oxygen consumption test during arm exercise reliability and validity. *Journal of Applied Physiology*, 38, 424-426.

Barnett, A., Smith, B., Lord, S.R., Williams, M., & Baumand, A. (2003). Community-based group exercise improves balance and reduces falls in at-risk older people: a randomised controlled trial. *Age and Ageing*, 32 (4), 407-414.

Barry, B. K., & Carson, R. G. (2004). The consequences of resistance training for movement control in older adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 59 (7), M730-M754.

Basmajian, J. & DeLuca, C. (1985) *Muscles Alive. Their Functions Revealed by Electromyography*. (5th ed.) Baltimore: Williams and Wilkins.

Bauer, C., Gröger, I., Rupprecht, R., & Gaßmann, K.G. (2008). Intrasession reliability of force platform parameters in community-dwelling older adults. *Archives of Physical Medicine and Rehabilitation*, 89 (10), 1977-1982.

Bellew, J.W., Panwitz, B.L., Peterson, L., Brock, M.C., Olson, K. E., & Staples, W. H. (2009). Effect of acute fatigue of the hip abductors on control of balance in young and older women. *Archives of Physical Medicine and Rehabilitation*, 90 (7), 1170-1175.

Bellew, J.W., Yates, J. W., & Gater, D.R. (2003). The initial effects of low-volume strength training on balance in untrained older men and women. *The Journal of Strength & Conditioning Research*, 17 (1), 121-128.

Benvenuti, F., Mecacci, R., Gineprari, I., Bandinelli, S., Benvenuti, E., Ferrucci, L., & Stanhope, S.J. (1999). Kinematic characteristics of standing disequilibrium: reliability and validity of a posturographic protocol. *Archives of Physical Medicine and Rehabilitation*, 80 (3), 278-287

Berg, K.O., Wood-Dauphinee, S.L., Williams, J.I., & Maki, B. (1992). Measuring balance in the elderly: Validation of an instrument. *Canadian Journal of Public Health,* 83, S7-11.

Berger, L.L., Regueme, S.C., & Forestier, N. (2010). Unilateral lower limb muscle fatigue induces bilateral effects on undisturbed stance and muscle EMG activities. *Journal of Electromyography and Kinesiology,* 20, 947–952.

Bernasconi, S., Tordi, N., Perrey, S., Parratte, B., & Monnier, G. (2006). Is the VO_2 slow component in heavy arm-cranking exercise associated with recruitment of type II muscle fibers as assessed by an increase in surface EMG? *Applied Physiology, Nutrition, and Metabolism*, 31 (4), 414-422.

Besson, H., Ekelund, U., Brage, S. Ø. R. E. N., Luben, R., Bingham, S., Khaw, K. T., & Wareham, N. J. (2008). Relationship between subdomains of total physical activity and mortality. *Medicine and Science in Sports and Exercise*, 40 (11), 1909-1915.

Betik, A.C., & Hepple, R.T. (2008). Determinants of V O2 max decline with aging: an integrated perspective. *Applied Physiology, Nutrition, and Metabolism*, 33 (1), 130-140.

Bevegard, S., Freyschuss, U., & Strandell T. (1966). Circulatory adaptation to arm and leg exercise in supine sitting position. *Journal of Applied Physiology*, 12, 37–46.

Bhambhani, Y.N., Eriksson, P., & Gomes, P.S. (1991). Transfer effects of endurance training with the arms and legs. *Medicine and Science in Sports and Exercise,* 23, 1035–1041.

Bhattacharya, A., Morgan, R., Shukla, R., Ramakrishan, H.K., Wang, L. (1987). Non-invasive estimation of afferent inputs for postural stability under low levels of alcohol. *Annals of Biomechanical Engineering,* 15, 533-50.

Bird, M.L., Hill, K., Ball, M.J., & Williams, A.D. (2009). Effects of resistance and flexibility exercise interventions on balance and related measures in older adults. *Journal of Aging and Physical Activity*, 17(4), 444-454

Bisson, E.J., McEwen, D., Lajoie, Y., & Bilodeau, M. (2011). Effects of ankle and hip muscle fatigue on postural sway and attentional demands during unipedal stance. *Gait and Posture*, 33, 83-87.

Bisson, E.J., Chopra, S., Azzi, E., Morgan, A., & Bilodeau, M. (2010). Acute effects of fatigue of the plantarflexor muscles on different postural tasks. *Gait and Posture*, 32 (4), 482-486.

Bisson, E.J., McEwen, D., Lajoie, Y., & Bilodeau, M. (2011). Effects of ankle and hip muscle fatigue on postural sway and attentional demands during unipedal stance. G*ait & posture*, 33(1), 83-87.

Bizid, R., Margnes, E., Franc¸ ois, Y., Gonzalez, G., Jully, J.L., Dupui, P., Paillard, T., (2009). Effects of knee and ankle muscle fatigue on postural control in the unipedal stance. *European Journal of Applied Physiology*, 106, 375–380.

Bjerkefors, A., & Thorstensson, A. (2006). Effects of kayak ergometer training on motor performance in paraplegics. *International Journal of Sports Medicine*, 27(10), 824-829.

Bjerkefors, A., Carpenter, M. G., & Thorstensson, A. (2007). Dynamic trunk stability is improved in paraplegics following kayak ergometer training. *Scandinavian Journal of Medicine & Science in Sports*, 17 (6), 672-679.

Björklund, M., Crenshaw, A. G., Djupsjöbacka, M., & Johansson, H. (2000). Position sense acuity is diminished following repetitive low-intensity work to fatigue in a simulated occupational setting. *European Journal of Applied Physiology*, 81 (5), 361-367.

Black, F. O., Wall III, C., Rockette Jr, H.E., & Kitch, R. (1982). Normal subject postural sway during the Romberg test. *American Journal of Otolaryngology*, 3 (5), 309-318.

Blaszczyk, J.W., Lowe, D.L., & Hansen, P.D. (1994). Ranges of postural stability and their changes in the elderly. *Gait & Posture*, 2 (1), 11-17.

Bloem, B., Allum, J.H.J., Carpenter, M., Verschuuren, J.J.G.M., & Honegger, F. (2002). Triggering of balance corrections and compensatory strategies in a patient with total leg proprioceptive loss. *Experimental Brain Research*, 142 (1), 91-107.

Blum, L., & Korner-Bitensky, N. (2008). Usefulness of the Berg Balance Scale in stroke rehabilitation: a systematic review. *Physical Therapy*, 88 (5), 559-566.

Bohannon, R. W. (2008). Population representative gait speed and its determinants. *Journal of Geriatric Physical Therapy*, 31 (2), 49-52.

Bohannon, R.W. (1997). Comfortable and maximum walking speed of adults aged 20—79 years: reference values and determinants. *Age and Ageing*, 26 (1), 15-19.

Borg, G. (1982). Psychophysical bases of perceived exertion. *Medicine and Science in Sports and Exercise*, 14, 377–381.

Borg, G. (1998). *Borg's perceived exertion and pain scales*. Human kinetics

Borg, G., Hassmén, P., & Lagerström, M. (1987). Perceived exertion related to heart rate and blood lactate during arm and leg exercise. *European Journal of Applied Physiology and Occupational Physiology*, 56 (6), 679-685.

Bottoms, L., & Price, M. (2014). The effect of arm training on thermoregulatory responses and calf volume during upper body exercise. *European Journal of Applied Physiology*, 114 (6), 1113-1122.

Boullion, L.E., Sklenka, D.K., & Driver, A.C. (2009). Comparison of training between 2 cycle ergometers on dynamic balance for middle-aged women. *Journal of Sport Rehabilitation*, 18 (2), 316.

Bouillon, L. E., & Baker, J. L. (2011). Dynamic balance differences as measured by the star excursion balance test between adult-aged and middle-aged women. *Sports Health: A Multidisciplinary Approach*, 3 (5), 466-469.

Bouisset, S., Duchêne, J.L., (1994). Is body balance more perturbed by respiration in seating than in standing posture? *Neuro Report,* 5, 957–960.

Bove, M., Faelli, E., Tacchino, A., Lofarno, F., Cogo, C.E., & Ruggeri, P. (2007) Postural control after a strenuous treadmill exercise. *Neuroscience Letters*, 18, 279–281

Brauer, S.G., Woollacott, M., & Shumway-Cook, A. (2001). The interacting effects of cognitive demand and recovery of postural stability in balance-impaired elderly persons. *Journals of Gerontology. Series A, Biology Sciences and Medical Sciences*, 56, M489–M496.

Bressel, E. & Heise, G. D. (2004). Effect of Arm Cranking Direction on EMG, Kinematic, and Oxygen Consumption Responses. *Journal of Applied Biomechanics*, 20, 129-143

Bressel, E., Bressel, M., Marquez, M. & Heise, G.D. (2001). The effect of handgrip position on upper extremity neuromusuclar responses to arm cranking exercise. *Journal of Electromyography & Kinesiology*, 11, 291-298.

Bressel, E., Yonker, J. C., Kras, J., & Heath, E. M. (2007). Comparison of static and dynamic balance in female collegiate soccer, basketball, and gymnastics athletes. *Journal of Athletic Training*, 42 (1), 42.

Bruniera, C. A., Rogerio, F. R., & Rodacki, A. L. (2013). Stabilometric response during single-leg stance after lower limb muscle fatigue. *Brazilian Journal of Physical Therapy*, 17 (5), 464-469.

Bryant, E.C., Trew, M.E., Bruce, A.M., Kuisma, R.M.E., & Smith, A.W. (2005). Gender differences in sway performance at the time of retirement. *Clinical Biomechanics*, 20, 330-335.

Buchner, D.M., Cress, M.E., de Lateur, B.J., Esselman, P.C., Margherita, A.J., Price, R., & Wagner, E.H. (1997a). The effect of strength and endurance training on gait, balance, fall risk, and health services use in community-living older adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 52 (4):M218-24.

Buchner, D.M., Cress, M.E., De Lateur, B.J., Esselman, P.C., Margherita, A.J., Price, R., & Wagner, E.H. (1997b). A comparison of the effects of three types of endurance training on balance and other fall risk factors in older adults. *Aging Clinical and Experimental Research*, 9 (1-2), 112-119.

Burden, A., & Bartlett, R. (1999). Normalisation of EMG amplitude: an evaluation and comparison of old and new methods. *Medical Engineering & Physics*, 21 (4), 247-257.

Burdet, C., & Rougier, P. (2004). Effects of utmost fatigue on undisturbed upright stance control. *Science and Sports*, 19, 308-316.

Buskirk, E.R., & Hodgson, J.L. (1987). Aging and aerobic power: the rate of change in men and women. *Federation Proceedings*, 46, 1824–1829.

Cappozzo, A., Catani, F., Della Croce, U., & Leardini, A. (1995). Position and orientation in space of bones during movement: anatomical frame definition and determination. *Clinical Biomechanics*, 10 (4), 171-178.

Carpenter, M.G., Frank, J.S., Winter, D.A., & Peysar, G.W. (2001). Sampling duration effects on centre of pressure summary measures. *Gait and Posture,* 13, 35–40.

Caron, O. (2004). Is there interaction between vision and local fatigue of the lower limbs on postural control and postural stability in human posture? *Neuroscience Letters*, 363, 18–21.

Celli, B., Criner, G., & Rassulto, J. (1985). Ventilatory muscle recruitment during unsupported arm exercise in normal subjects. *Journal of Applied Physiology*, 64, 1936-41.

Chiari, L., Rocchi, L., & Cappello, A. (2002). Stabilometric parameters are affected by anthropometry and foot placement. *Clinical Biomechanics*, *17*(9), 666-677.

Clausen, J.P., Trap-Jensen, J., & Lassen, N.A. (1970). The effects of training on the heart rate during arm and leg exercise. *Scandinavian Journal of Clinical & Laboratory Investigation*, 26 (3), 295-301.

Clifford, A.M., & Holder-Powell, H. (2010). Postural control in healthy adults. *Clinical Biomechanics*, 25, 546-551.

Coggan, A.R., Spina, R.J., King, D.S., Rogers, M.A., Brown, M., Nemeth, P.M., & Holloszy, J.O. (1992). Histochemical and enzymatic comparison of the gastrocnemius muscle of young and elderly men and women. *Journal of gerontology*, *47* (3), B71-B76.

Collett, M.E., & Liljestrand, G. (1924) The minute volume of the heart in man during some different types of exercise. *Skandinavisches Archiv Für Physiologie,* 45, 29–42.

Collins, J.J., & De Luca, C.J. (1994). Random walking during quiet standing. *Physical Review Letters*, 73 (5), 764.

Conforto, S., Schmid, M., Camomilla, V., D'Alessio, T., Cappozzo, A. (2001). Hemodynamics as a possible internal mechanical disturbance to balance. *Gait & Posture*, 14, 28–35.

Corbeil, P., Blouin, J. S., Bégin, F., Nougier, V., & Teasdale, N. (2003). Perturbation of the postural control system induced by muscular fatigue. *Gait & Posture*, *18* (2), 92-100.

Couser, J.I., Martinez, F.J., & Celli, B.R. (1992). Respiratory response and ventilatory muscle recruitment during arm elevation in normal subjects. *CHEST Journal*, 101 (2), 336-340.

Cress, M.E., Buchner, D.M., Questad, K.A., Esselman, P.C., DeLateur, B.J., & Schwartz, R.S. (1996). Continuous-scale physical functional performance in healthy older adults: a validation study. *Archives of Physical Medicine and Rehabilitation*, 77, 1243–1250.

Creath, R., Kiemel, T., Horak, F., Peterka, R., & Jeka, J. (2005). A unified view of quiet and perturbed stance: simultaneous co-existing excitable modes. *Neuroscience Letters*, 377 (2), 75-80.

Cummings, S.R., Nevitt, M.C., Browner, W.S., Stone, K., Fox, K.M., Ensrud, K. E. & Vogt, T. M. (1995). Risk factors for hip fracture in white women. *New England Journal of Medicine*, 332 (12), 767-774.

Davidson, B.S., Madigan, M.L., & Nussbaum, M.A. (2004). Effects of lumbar extensor fatigue and fatigue rate on postural sway. *European Journal of Applied Physiology*, 91, 183-189.

Davies, C.J. & Sargeant, A.J. (1975). Effects of training on the physiological response to one- and two-leg work. *Journal of Applied Physiology*, 38, 377-381.

Davis, J.A., Vodak, P., Wilmore, J.H., Vodak, J., & Kurtz, P. (1976). Anaerobic threshold and maximal aerobic power for three modes of exercise. *Journal of Applied Physiology*, 41 (4), 544-550.

Davy, K. P., Johnson, D. G., & Seals, D. R. (1995). Cardiovascular, plasma norepinephrine, and thermal adjustments to prolonged exercise in young and older healthy humans. *Clinical Physiology*, 15 (2), 169-181.

DeJong, A.T., Bonzheim, K., Franklin, B.A., & Saltarelli, W. (2009). Cardiorespiratory responses to maximal arm and leg exercise in national-class marathon runners. *Physician and Sports Medicine*, 37, 120–126.

Deliagina, T.G., Orlovsky, G.N., Zelenin, P.V., & Beloozerova, I.N. (2006). Neural bases of postural control. *Physiology*, 21 (3), 216-225

Deliagina, T.G., Zelenin, P.V., Beloozerova, I.N., & Orlovsky, G.N. (2007). Nervous mechanisms controlling body posture. *Physiology & Behaviour*, 92(1), 148-154.

Demura, S., & Uchiyama, M. (2009). Influence of anaerobic and aerobic exercises on the centre of pressure during an upright posture. *Journal of Exercise Science & Fitness*, 7, 39-47.

Derave, W., DeClercq, D., Bouckaert. J., & Pannier, J.L. (1998). The influence of exercise and dehydration on postural stability. *Ergonomics*, 41, 782-789.

Derave, W., Tombeux, N., Cottyn, J., Pannier, J.L., & De Clercq, D. (2002). Treadmill exercise negatively affects visual contribution to static postural stability. *International Journal of Sports Medicine*, 23, 44–49.

Di Blasio, A., Sablone, A., Civino, P., D'Angelo, E., Gallina, S., & Ripari, P. (2009). Arm vs. combined leg and arm exercise: Blood pressure responses and ratings of perceived exertion at the same indirectly determined heart rate. *Journal of Sports Science & Medicine*, 8 (3), 401.

Diener, H.C., Dichgans, J., Guschlbauer, B., & Mau, H. (1984). The significance of proprioception on postural stabilization as assessed by ischemia. *Brain Research*, 296 (1), 103-109.

Dijkers, M.P., Kropp, G.C., Esper, R.M., Yavuzer, G., Cullen, N., & Bakdalieh, Y. (2002). Reporting on reliability and validity of outcome measures in medical rehabilitation research. *Disability and Rehabilitation*, 24, 819–27.

Donath, L., Zahner, L., Roth, R., Fricker, L., Cordes, M., Hanssen, H., Schmidt-Truckass, A., & Faude, O. (2013). Balance and gait performance after maximal and submaximal endurance exercise in seniors: is there a higher fall-risk? *European Journal of Applied Physiology*. 113: 611-619.

Donath, L., Faude, O., Roth, R., & Zahner, L. (2014). Effects of stair-climbing on balance, gait, strength, resting heart rate, and submaximal endurance in healthy seniors. *Scandinavian Journal of Medicine & Science in Sports*, 24 (2), e93-e101.

Douris, P.C., Handrakis, J.P., Gendy, J., Salama, M., Kwon, D., Brooks, R., Salama, N., & Southard, V. (2011). Fatiguing upper body aerobic exercise impairs balance. *Journal of Strength and Conditioning Research*, 25, 3299-3305.

Doyle, R.J., Hsiao-Wecksler, E.T., Ragan, B.G., & Rosengren, K.S. (2007). Generalizability of center of pressure measures of quiet standing. *Gait & Posture*, 25 (2), 166-171.

Duarte, M., & Freitas, S. M. (2010). Revision of posturography based on force plate for balance evaluation. *Brazilian Journal of Physical Therapy*, 14 (3), 183-192.

Duarte, M., & Sternad, D. (2008). Complexity of human postural control in young and older adults during prolonged standing. *Experimental Brain Research*, 191 (3), 265-276.

Duncan, P. W., Weiner, D. K., Chandler, J., & Studenski, S. (1990). Functional reach: a new clinical measure of balance. *Journal of Gerontology*, 45 (6), M192-M197.

Duncan, G. E., Howley, E. T., & Johnson, B. N. (1997). Applicability of VO2max criteria: discontinuous versus continuous protocols. *Medicine and Science in Sports and Exercise*, 29 (2), 273-278.

Du Pasquier, R.A., Blanc, Y., Sinnreich, M., Landis, T., Burkhard, P., & Vingerhoets, F.J.G. (2003). The effect of aging on postural stability: a cross sectional and longitudinal study. *Clinical Neurophysiology*, 33 (5), 213-218.

Earl, J. E., & Hertel, J. (2001). Lower-extremity muscle activation during the Star Excursion Balance Tests. *Journal of Sport Rehabilitation*, 10 (2), 93-104.

Egerton, T., Brauer, S.G., Cresswell, A.G. (2009). The immediate effect of physical activity of standing sway in healthy and sway- impaired older people. *Australasian Journal of Ageing* 28, 93-96.

Egerton, T., Brauer, S. G., & Cresswell, A. G. (2010). Dynamic postural stability is not impaired by moderate-intensity physical activity in healthy or balance-impaired older people. *Human Movement Science*, 29 (6), 1011-1022.

Ehsani, A.A., Ogawa, T., Miller, T.R., Spina, R.J., & Jilka, S.M. (1991). Exercise training improves left ventricular systolic function in older men. *Circulation*, 83 (1), 96-103.

Eklund, G. (1972). General features of vibration-induced effects on balance. *Journal of Medical Sciences*, 77 (2), 112-124.

Enoka, R.M. (2008). *Neuromechanics of Human Movement* (4th ed.). Champaign: Human Kinetics.

Enoka, R.M., & Stuart, D.G. (1992). Neurobiology of muscle fatigue. *Journal of Applied Physiology*, 72 (5), 1631-1648.

Enoka, R.M. (1997). Neural adaptations with chronic physical activity. *Journal of Biomechanics*, 30 (5), 447-455.

Era, P., Konttinen, N., Mehto, P., Saarela, P., & Lyytinen, H. (1996). Postural stability and skilled performance—a study on top-level and naive rifle shooters. *Journal of Biomechanics*, 29 (3), 301-306.

Era, P., Sainio, P., Koskinen, S., Haavisto, P., Vaara, M., & Aromaa, A. (2006). Postural balance in a random sample of 7,979 subjects aged 30 years and over. *Gerontology*, 52 (4), 204-213.

Era, P., & Heikkinen, E. (1985). Postural sway during standing and unexpected disturbance of balance in random samples of men of different ages. *Journal of Gerontology*, 40 (3), 287-295.

Ericson, M.O., Nisell, R., Arborelius, U.P., Ekholm, J. (1985). Muscular activity during ergometer cycling. *Scandinavian Journal of Rehabilitation Medicine*, 17, 53–61.

Eston, R.G., & Brodie, D.A. (1986). Responses to arm and leg ergometry. *British Journal of Sports Medicine*, 20 (1), 4-6.

Fairbarn, M. S., Blackie, S. P., McElvaney, N. G., Wiggs, B. R., & Pardy, R. L. (1994). Prediction of heart rate and oxygen uptake during incremental and maximal exercise in healthy adults. *CHEST Journal*, 105 (5), 1365-1369.

Fisher, N.M., Pendergast, D.R., & Calkins, E. (1991). Muscle rehabilitation in impaired elderly nursing home residents. *Archives of Physical Medicine and Rehabilitation*, 72 (3), 181-185

Fitzpatrick, R., & McCloskey, D.I., (1994). Proprioceptive, visual and vestibular thresholds for the perception of sway during standing in humans. *Journal of Physiology*, 478, 173–186.

Fitzpatrick, R.C., Gorman, R.B., Burke, D., & Gandevia, S.C., (1992). Postural proprioceptive reflexes in standing human subjects: bandwidth of response and transmission characteristics. *Journal of Physiology*, 458, 69–83.

Forbes, S. C., & Chilibeck, P. D. (2007). Comparison of a kayaking ergometer protocol with an arm crank protocol for evaluating peak oxygen consumption. *Journal of Strength and Conditioning Research*, 21, 1282–1285.

Forestier, N., Teasdale, N., & Nougier, V. (2002). Alteration of the position sense at the ankle induced by muscular fatigue in humans. *Medicine and Science in Sports and Exercise*, 34 (1), 117-122.

Fox, Z.G., Mihalik, J.P., Blackburn, J.T., Battaglini, C.L., & Guskiewicz, K.M. (2008). Return of postural control to baseline after anaerobic and aerobic exercise protocols. *Journal of Athletic Training*, 43, 456–463.

Fox, S.M., Naughton, J.P., & Haskell, W.L. (1971) Physical activity and the prevention of coronary heart disease. *Annals of Clinical Research,* 3, 404 –432.

Franklin, B. A. (1989). Aerobic exercise training programs for the upper body. *Medicine & Science in Sports & Exercise,* 21, S141-s148.

Franklin, B. A., Whaley, M. H,, Howley, E. T. (2000). *American college of sports medicine guidelines for exercise testing and prescription*. Baltimore: Williams and Wilkins.

Frontera, W. R., Hughes, V. A., Fielding, R. A., Fiatarone, M. A., Evans, W. J., & Roubenoff, R. (2000). Aging of skeletal muscle: a 12-yr longitudinal study. *Journal of Applied Physiology*, 88 (4), 1321-1326

Fujisawa, N., Masuda, T., Inaoka, H., Fukuoka, Y., Ishida, A., & Minamitani, H. (2005). Human standing posture control system depending on adopted strategies*. Medical and Biological Engineering and Computing*, 43 (1), 107-114.

Gaesser, G.A., & Poole, D.C. (1988). Blood lactate during exercise: time course of training adaptation in humans. *International Journal of Sports Medicine*, 9, 284-8.

Gardner, M.M., Robertson, M.C., & Campbell, A.J. (2000). Exercise in preventing falls and fall related injuries in older people: a review of randomised controlled trials. *British Journal of Sports Medicine*, 34, 7–17.

Garland, S.J., & Kaufman, M.P., (1995). *Role of muscle afferents in the inhibition of motoneurons during fatigue*. In: Gandevia, S.C., Enoka, R.M., McComas, A.J., Stuart, D.G., Thomas, C.K. (Eds.), *Fatigue: Neural and Muscular Mechanisms*. Plenum, New York, pp. 271–278.

Gauchard, G.C., Gangloff, P., Vouriot, A., Mallie, J. P., & Perrin, P.P. (2002). Effects of exercise-induced fatigue with and without hydration on static postural control in adult human participants. *International Journal of Neuroscience*, 112, 1191-1206.

Gauchard, G.C., Jeandel, C., Tessier, A., & Perrin, P.P. (1999). Beneficial effect of proprioceptive physical activities on balance control in elderly human subjects. Neuroscience Letters, 273, 81–4.

Gil, A.W., Oliveira, M.R., Coelho, V.A., Carvalho, C.E., Teixeira, D.C., & Silva Jr, R.A.D. (2011). Relationship between force platform and two functional tests for measuring balance in the elderly. *Brazilian Journal of Physical Therapy*, 15 (6), 429-435.

Gledhill, N., Cox, D. & Jamnick, R. (1994). Endurance athletes' stroke Volume does not plateau: major advantage is diastolic function. *Medicine and Science in Sports Exercise*, 26, 1116–1121.

Godfrey, B.S. & Lawrence, P. (2005). *Crime and Justice, 1750–1950*. Portland, OR: Willan Publishing.

Goldie, P. A., Bach, T. M., & Evans, O. M. (1989). Force platform measures for evaluating postural control: reliability and validity. *Archives of Physical Medicine and Rehabilitation*, 70 (7), 510-517.

Goodwin, V.A., Abbott, R.A., Whear, R., Bethel, A., Ukoumunne, O.C., Thompson-Coon, J., & Stein, K. (2014). Multiple component interventions for preventing falls and fall-related injuries among older people: systematic review and meta-analysis. *BMC geriatrics*, 14 (1), 15.

Goosey-Tolfrey, V.L., & Sindall, P. (2007). The effects of arm crank strategy on physiological responses and mechanical efficiency during submaximal exercise. *Journal of Sports Sciences*, 25, 453–460.

Goosey-Tolfrey, V. L., Alfano, H., & Fowler, N. (2008). The influence of crank length and cadence on mechanical efficiency in hand cycling. *European Journal of Applied Physiology*, 102, 189–194.

Granacher, U., Gruber, M., Förderer, D., Strass, D., & Gollhofer, A. (2010). Effects of ankle fatigue on functional reflex activity during gait perturbations in young and elderly men. *Gait & Posture*, 32(1), 107-112.

Granacher, U., Lacroix, A., Muehlbauer, T., Roettger, K., & Gollhofer, A. (2012). Effects of core instability strength training on trunk muscle strength, spinal mobility, dynamic balance and functional mobility in older adults. *Gerontology*, 59 (2), 105-113.

Grange, C.C., Maire, J., Groslambert, A., Tordi, N., Dugue, B., Pernin, J.N., Rouillon, J.D. (2004). Perceived exertion and rehabilitation with arm crank in elderly patients after total hip arthroplasty: a preliminary study. *Journal of Rehabilitation and Research Development*, 41 (4), 611-20.

Granata, K. P., & Lockhart, T. E. (2008). Dynamic stability differences in fall-prone and healthy adults. *Journal of Electromyography and Kinesiology*, 18 (2), 172-178.

Gribble, P. A., & Hertel, J. (2004). Effect of lower-extremity muscle fatigue on postural control. *Archives of Physical Medicine and Rehabilitation*, 85 (4), 589-592.

Gribble, P.A., Hertel, J., 2004a. Effect of hip and ankle muscle fatigue on unipedal postural control. *Journal of Electromyography and Kinesiology*, 14, 641–646.

Gribble, P. A., Hertel, J., & Plisky, P. (2012). Using the Star Excursion Balance Test to assess dynamic postural-control deficits and outcomes in lower extremity injury: a literature and systematic review. *Journal of Athletic Training*, 47 (3), 339-357.

Gribble, P. A., & Hertel, J. (2003). Considerations for normalizing measures of the Star Excursion Balance Test. *Measurement in Physical Education and Exercise Science*, 7 (2), 89-100.

Gribble, P. A., Tucker, W. S., & White, P. A. (2007). Time-of-day influences on static and dynamic postural control. *Journal of Athletic Training*, 42 (1), 35-41.

Grigorenko, A., Bjerkefors, A., Rosdahl, H., Hultling, C., Alm, M., & Thorstensson, A. (2004). Sitting balance and effects of kayak training in paraplegics. *Journal of Rehabilitation Medicine*, 36 (3), 110-116.

Groslambert, A., Grange, C.C., Perrey, S., Maire, J., Tordi, N., Rouillon, J.D. (2006). Effects of Aging on Perceived Exertion and Pain During Arm Cranking in Women 70 to 80 Years Old. *Journal of Sports Science and Medicine*, 5 (2), 208-14.

Gschwind, Y.J., Kressig, R.W., Lacroix, A., Muehlbauer, T., Pfenninger, B., & Granacher, U. (2013). A best practice fall prevention exercise program to improve balance, strength/power, and psychosocial health in older adults: study protocol for a randomized controlled trial. *BMC geriatrics*, 13 (1), 105

Hadders-Algra, M., & Carlberg, E. B. (2008). *Postural control: A key issue in developmental disorders*. London: Mac Keith

Hagberg, J.M., Allen, W.K., Seals, D.R., Hurley, B.F., Ehsani, A.A. & Holloszy, J.O. (1985). A hemodynamic comparison of young and older endurance athletes during exercise. *Journal of Applied Physiology*, 58, 2041–2046.

Hagberg, J.M., Graves, J.E., Limacher, M., Woods, D.R., Leggett, S.H., Cononie, C., Gruber, J.J., Pollock, M.L. (1989). Cardiovascular responses of 70–90 yr-old men and women to exercise training. *Journal of Applied Physiology*, 66, 2589–2594.

Hamilton, N., & Luttgens, K. (2002). *Kinesiology, Scientific Basis of Human Motion*, 10thed. Boston: McGraw-Hill.

Harris, C., Wattles, A.P., DeBeliso, M., Sevene-Adams, P.G., Berning, J. M., & Adams, K. J. (2011). The seated medicine ball throw as a test of upper body power in older adults. *The Journal of Strength & Conditioning Research*, 25 (8), 2344-2348.

Harkins, K.M., Mattacola, C.G., Uhl, T.L., Malone, T.R., & McCrory, J.L. (2005). Effects of 2 ankle fatigue models on the duration of postural stability dysfunction. *Journal of Athletic Training*, 40, 191–196.

Hassanlouei, H., Falla, D., Arendt-Nielsen, L., & Kersting, U. G. (2014). The Effect of Six Weeks Endurance Training on Dynamic Muscular Control of the Knee Following Fatiguing Exercise. *Journal of Electromyography and Kinesiology*. In Press.

Hawkins, S.A., & Wiswell, R.A. (2003). Rate and mechanism of maximal oxygen consumption decline with aging. *Sports Medicine*, 33 (12), 877-888.

Hawkins, S.A., Marcell, T.J., Victoria, J.S., & Wiswell, R.A. (2001). A longitudinal assessment of change in VO2max and maximal heart rate in master athletes. *Medicine and Science in Sports and Exercise*, 33 (10), 1744-1750.

Helge, J. W. (2010). Arm and leg substrate utilization and muscle adaptation after prolonged low-intensity training. *Acta Physiologica*, 199, 519-528.

Hellerstein, H.K., & Franklin, B A. (1984). Evaluating the cardiac patient for exercise therapy. Role of exercise testing. *Clinics in Sports Medicine*, 3 (2), 371-393.

Hellerstein, H.K. (1977). Prescription of vocational and leisure activities. Practical aspects. *Advances in Cardiology*, (24), 105-115.

Hermens, H.J., Freriks, B., Disselhorst-Klug, C. & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology,* 10, 361-374.

Hermansen, L., & Andersen, K. L. (1965). Aerobic work capacity in young Norwegian men and women. *Journal of Applied Physiology*, 20 (3), 425-431.

Hiatt, W.R., Hirsch, A.T., Regensteiner, J.G., & Brass, E.P. (1995). Clinical trials for claudication assessment of exercise performance, functional status, and clinical end points. *Circulation*, 92 (3), 614-621.

Hiemstra, L. A., Lo, I. K., & Fowler, P. J. (2001). Effect of fatigue on knee proprioception: implications for dynamic stabilization. *Journal of Orthopaedic & Sports Physical Therapy*, 31 (10), 598-605.

Hill, M.W., Goss-Sampson, M., Duncan, M., Price, M.J. (2014). The effects of maximal and submaximal arm crank ergometry and cycle ergometry on postural sway. *European Journal of Sport Science*, Ahead of print, 1-9.

Hjeltnes, N. (1977). Oxygen uptake and cardiac output in graded arm exercise in paraplegics with low level spinal lesions. *Scandinavian Journal of Rehabilitation Medicine*, 9 (3), 107-13.

Ho, C. W., Beard, J. L., Farrell, P. A., Minson, C. T., & Kenney, W. L. (1997). Age, fitness, and regional blood flow during exercise in the heat. *Journal of Applied Physiology*, 82 (4), 1126-1135.

Hodges, P.W., Gurfinkel, V.S., Brumagne, S., Cordo, P.C., (2002). Coexistence of stability and mobility in postural control: evidence from postural compensation for respiration. *Experimental Brain Research*, 144, 293–302

Hooker, S.P. & Wells, C.L. (1991). Physiologic responses to arm crank exercise with and without hand grasping. *Clinical Kinesiology* (Online Edition), 45, 3-8.

Hopkins, W., Marshall, S., Batterham, A. & Hanin, J. (2009). Progressive statistics for studies in sports medicine and exercise science. *Medicine & Science in Sports & Exercise* 41, 3-12.

Hopman, M.T.E., Van Teeffelen, W.M., Brouwer, J., Houtman, S. & Binkhorst, R.A. (1995). Physiological responses to asynchronous and synchronous arm-cranking exercise. *European Journal of Applied Physiology & Occupational Physiology*, 72, 111-114.

Horak, F. (1997). Clinical assessment of balance disorders. *Gait & Posture,* 6, 76-84.

Horak, F.B. & Nashner, L.M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. *Journal of Neurophysiology*, 55, 1369-1381.

Horak, F. B. (2006). Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? *Age and Ageing*, 35, ii7-ii11.

Howe, T. E., Rochester, L., Jackson, A., Banks, P. M. H., & Blair, V. A. (2008). Exercise for improving balance in older people. *Cochrane Database of Systematic Reviews*, 4, CD004963

Hufschmidt, A., Dichgans, J., Mauritz, K.H., Hufschmidt, M. (1980). Some methods and parameters of body sway quantification and their neurological application. *Archiv fur Psychiatrie und Nervenkrankheiten,* 228, 135-50.

Hughes, M. A., Duncan, P. W., Rose, D. K., Chandler, J. M., & Studenski, S. A. (1996). The relationship of postural sway to sensorimotor function, functional performance, and disability in the elderly. *Archives of Physical Medicine and Rehabilitation*, *77* (6), 567-572.

Hunter, S.K., Duchateau, J., Enoka, R.M., (2004). Muscle fatigue and the mechanisms of task failure. *Exercise and Sport Sciences Reviews*, 32, 44–49.

Hutchinson, P. L., Cureton, K. J., Outz, H., & Wilson, G. (1991). Relationship of cardiac size to maximal oxygen uptake and body size in men and women. *International Journal of Sports Medicine*, 369, 373.

Ilias, N.A., Xian, H., Inman, C., and Martin, W.H. (2009). Arm exercise testing predicts clinical outcome. *American Heart Journal*,157 (1), 69-76.

Illing, S., Choy, N. L., Nitz, J., & Nolan, M. (2010). Sensory system function and postural stability in men aged 30-80 years. *The Aging Male*, 13 (3), 202-210.

Ishida, A., & Imai, S. (1980). Responses of the posture-control system to pseudorandom acceleration disturbances. *Medical and Biological Engineering and Computing*, 18 (4), 433-438.

Isles, R. C., Choy, N. L., Steer, M., & Nitz, J. C. (2004). Normal values of balance tests in women aged 20–80. *Journal of the American Geriatrics Society*, 52 (8), 1367-1372.

Issekutz, B., Birkhead, N. C., & Rodahl, K. (1962). Use of respiratory quotients in assessment of aerobic work capacity. *Journal of Applied Physiology*, 17 (1), 47-50.

Jackson, A. S., Beard, E. F., Wier, L. T., Ross, R. M., Stuteville, J. E., & Blair, S. N. (1995). Changes in aerobic power of men, ages 25-70 yr. *Medicine and Science in Sports and Exercise*, 27(1), 113-120.

Jackson, A. S., Wier, L. T., Ayers, G. W., Beard, E. F., Stuteville, J. E., & Blair, S. N. (1996). Changes in aerobic power of women, ages 20-64 yr. *Medicine and Science in Sports and Exercise*, 28 (7), 884-891.

Jakobsen, M. D., Sundstrup, E., Krustrup, P., & Aagaard, P. (2011). The effect of recreational soccer training and running on postural balance in untrained men. *European Journal of Applied Physiology*, 111 (3), 521-530

Johnson, M., Polgar, J., Weightman, D., & Appleton, D. (1973). Data on the distribution of fibre types in thirty-six human muscles: an autopsy study. *Journal of the Neurological Sciences*, 18 (1), 111-129.

Joseph, J., & Nightingale, A. (1954). Electromyography of muscles of posture: thigh muscles in males. *The Journal of Physiology*, 126 (1), 81-85.

Joseph, J., & Nightingale, A. (1956). Electromyography of muscles of posture: leg muscles in males. *The Journal of Physiology*, 117, 484-491.

Joseph, J., & Nightingale, A. (1956). Electromyography of muscles of posture: leg and thigh muscles in women, including the effects of high heels. *The Journal of Physiology*, 132 (3), 465-468.

Judge, J. O., Kenny, A. M., & Kraemer, W. J. (2003). Exercise in older adults. *Connecticut Medicine*, 67 (8), 461-464.

Kaji, A., Sasagawa, S., Kubo, T., & Kanehisa, H. (2010). Transient effect of core stability exercises on postural sway during quiet standing. *The Journal of Strength & Conditioning Research*, 24 (2), 382-388.

Kanekar, N., Santos, M.J., & Aruin, A. (2008). Anticipatory postural control following fatigue of postural and focal muscles. *Clinical Neurophysiology*, 11, 2304–2313.

Kang, J., Chaloupka, E. C., Mastrangelo, M. A. & Angelucci, J. (1999) Physiological responses to upper body exercise on an arm and a modified leg ergometer. *Medicine & Science in Sports & Exercise*, 31, 1453-1459.

Kang, J., Chaloupka, E.C., Mastrangelo, M.A., Donnelly, M.S., Martz, W.P. & Robertson, R.J. (1998). Regulating exercise intensity using ratings of perceived exertion during arm and leg ergometry. *European Journal of Applied Physiology & Occupational Physiology*, 78, 241-246.

Kang, J., Robertson, R. J., Goss, F. L., Dasilva, S. G., Suminski, R. R., Utter, A. C., Zoeller, R. F. & Metz, K. F. (1997). Metabolic efficiency during arm and leg exercise at the same relative intensities. *Medicine & Science in Sports & Exercise*, 29, 377-382.

Kapteyn, T.S., Bles, W., Njiokiktjien, C.J., Kodde, L., Massen, C.H., & Mol, J.M.F. (1983) Standardization in platform stabilometry being a part of posturography. *Agressologie,* 24, 321–6

Karvonen, J., & Vurimaa T. (1988). Heart rate and exercise intensity during sports activities. Practical application. *Sports Medicine*, 5, 303-311.

Katzel, L.I., Sorkin, J.D., & Fleg, J.L. (2001). A comparison of longitudinal changes in aerobic fitness in older endurance athletes and sedentary men. *Journal of American Geriatric Society*, 49, 1657–1664.

Kavounoudias, A., Roll, R., & Roll, J. P. (1998). The plantar sole is a ‘dynamometric map’ for human balance control. *Neuroreport*, 9 (14), 3247-3252.

Kent-Braun, J. A. (1999). Central and peripheral contributions to muscle fatigue in humans during sustained maximal effort. *European Journal of Applied Physiology and Occupational Physiology*, 80 (1), 57-63.

Kiemel, T., Oie, K. S., & Jeka, J. J. (2002). Multisensory fusion and the stochastic structure of postural sway. *Biological Cybernetics*, 87 (4), 262-277.

Kiers, H., van Dieën, J., Dekkers, H., Wittink, H., & Vanhees, L. (2013). A systematic review of the relationship between physical activities in sports or daily life and postural sway in upright stance. *Sports Medicine*, 43 (11), 1171-1189.

Kirchner, M., Schubert, P., Schmidtbleicher, D., & Haas, C. T. (2012). Evaluation of the temporal structure of postural sway fluctuations based on a comprehensive set of analysis tools. *Physica A: Statistical Mechanics and its Applications*, 391 (20), 4692-4703.

Kramer, H., Lun, A., Mucke, R., Kuchler, G., (1979). Changes in mechanical and bioelectrical muscular activity and in heart rate due to sustained voluntary isometric contractions and time required for recovery. *Electromyography and Clinical Neurophysiology*, 19, 381–386.

Koppo, K., Bouckaert, J. & Jones, A.M. (2002). Oxygen uptake kinetics during high-intensity arm and leg exercise. *Respiratory Physiology & Neurobiology*, 133, 241-250.

Kouzaki, M., & Masani, K. (2012). Postural sway during quiet standing is related to physiological tremor and muscle volume in young and elderly adults. *Gait & Posture*, 35 (1), 11-17.

Kouzaki, M., & Shinohara, M. (2010). Steadiness in plantar flexor muscles and its relation to postural sway in young and elderly adults. *Muscle & Nerve*, 42 (1), 78-87.

Krishnamoorthy, V., Slijper, H., & Latash, M.L. (2002). Effects of different types of light touch on postural sway. *Experimental Brain Research*, 147 (1), 71-79.

Lafond, D., Corriveau, H., Hébert, R., & Prince, F. (2004). Intrasession reliability of center of pressure measures of postural steadiness in healthy elderly people. *Archives of Physical Medicine and Rehabilitation*, 85 (6), 896-901.

Lanza, I. R., Larsen, R. G., & Kent-Braun, J. A. (2007). Effects of old age on human skeletal muscle energetics during fatiguing contractions with and without blood flow. *The Journal of Physiology*, 583 (3), 1093-1105.

Latash M.L. (2008). *Neurophysiological basis of movement*. Champaign: Human Kinetics

Laughton, C. A., Slavin, M., Katdare, K., Nolan, L., Bean, J. F., Kerrigan, D. C. & Collins, J. J. (2003). Aging, muscle activity, and balance control: physiologic changes associated with balance impairment. *Gait & Posture*, 18 (2), 101-108.

Lazarus, B., Cullinane, E., & Thompson, P. (1981) Comparison of the results and reproducibility of arm and leg exercise tests in men with angina pectoris. American Journal of Cardiology, 47, 1075-9.

Le Clair K, Riach C. (1996).Postural stability measures: what to measure and for how long. Clinical Biomechanics, 11, 176-178.

Lee, D. N., & Lishman, J. R. (1975). Visual proprioceptive control of stance. Journal of *Human Movement Studies,* 1, 87-95.

Lepers, R., Bigard, A. X., Diard, J. P., Gouteyron, J. F., Guezennec, C. Y. (1997). Posture control after prolonged exercise. *European Journal or Applied Physiology*, 76, 55-61.

Lepers, R., Maffiuletti, N. A., Rochette, L., Brugniaux, J., & Millet, G. Y. (2002). Neuromuscular fatigue during a long-duration cycling exercise. *Journal of Applied Physiology*, 92 (4), 1487-1493.

Levy, W. C., Cerqueira, M. D., Abrass, I. B., Schwartz, R. S., & Stratton, J. R. (1993). Endurance exercise training augments diastolic filling at rest and during exercise in healthy young and older men. *Circulation*, 88 (1), 116-126.

Lewis, S., Thompson, P., Areskog, N.H., Vodak, P., Marconyak, M., De- Busk, R., Mellen, S., & Haskell, W. (1980). Transfer effects of endurance training to exercise with untrained limbs. *European Journal of Applied Physiology*, 44, 25–34.

Liaw, M. Y., Chen, C. L., Pei, Y. C., Leong, C. P., & Lau, Y. C. (2009). Comparison of the static and dynamic balance performance in young, middle-aged, and elderly healthy people. *Chang Gung Medical Journal*, *32* (3), 297-304.

Lichtenstein, M.J., Shields, S.L., Shiavi, R.G., & Burger, M.C. (1988). Clinical determinants of biomechanics platform measures of balance in aged women. *Journal of American Geriatric Society*, 36, 996–1002.

Lin, D., Nussbaum, M.A., Seol, H., Singh, N.B., Madigan, M.L., & Wojcik, L.A. (2009). Acute effects of localized muscle fatigue on postural control and patterns of recovery during upright stance: influence of fatigue location and age. *European Journal Applied Physiology*, 106, 425-434.

Lin, D., Seol, H., Nussbaum, M. A., & Madigan, M. L. (2008). Reliability of COP-based postural sway measures and age-related differences. *Gait & Posture*, 28 (2), 337-342.

Lion, A., Bosser, G., Gauchard, G. C., Djaballah, K., Mallié, J. P., & Perrin, P. P. (2010). Exercise and dehydration: A possible role of inner ear in balance control disorder. *Journal of Electromyography and Kinesiology*, 20 (6), 1196-1202.

Loftin, M., Boileau, R. A., Massey, B. H., & Lohman, T. G. (1988). Effect of arm training on central and peripheral circulatory function. *Medicine and Science in Sports and Exercise*, 20, 136–141.

Loram, I. D., & Lakie, M. (2002). Human balancing of an inverted pendulum: position control by small, ballistic-like, throw and catch movements. *The Journal of Physiology*, 540 (3), 1111-1124

Loram, I. D., Kelly, S. M., & Lakie, M. (2001). Human balancing of an inverted pendulum: is sway size controlled by ankle impedance? *The Journal of Physiology*, 532 (3), 879-891.

Loram, I. D., Lakie, M., Di Giulio, I., & Maganaris, C. N. (2009). The consequences of short-range stiffness and fluctuating muscle activity for proprioception of postural joint rotations: the relevance to human standing. *Journal of Neurophysiology*, 102 (1), 460-474.

Loram, I. D., Maganaris, C. N., & Lakie, M. (2005). Human postural sway results from frequent, ballistic bias impulses by soleus and gastrocnemius. *The Journal of Physiology*, 564(1), 295-311.

Lord, S.R., Castell, S., Corcoran, J., Dayhew, J., Matters, B., Shan, A., et al. (2003). The effect of group exercise on physical functioning and falls in frail older people living in retirement villages: a randomized, controlled trial. *Journal of the American Geriatrics Society*, 51 (12),1685–92.

Lord, S.R., Ward, J.A., Williams, P., & Strudwick, M. (1995). The effect of a 12- month exercise trial on balance, strength, and falls in older women: a randomized controlled trial. *Journal of the American Geriatrics Society,* 43 (11), 1198–206.

Lord, S. R. (2006). Visual risk factors for falls in older people. *Age and Ageing*, 35, ii42-ii45.

Lord, S. R., & Menz, H. B. (2000). Visual contributions to postural stability in older adults. *Gerontology*, 46 (6), 306-310.

Lord, S. R., Clark, R. D., & Webster, I. W. (1991). Physiological factors associated with falls in an elderly population. *Journal of the American Geriatrics Society*, 43, 1198-1206.

Lord, S. R., Sherrington, C., Menz, H. B., & Close, J. C. (2007). *Falls in older people: risk factors and strategies for prevention*. Cambridge University Press.

Lord, S.R., Rogers, M.W., Howland, A., & Fitzpatrick, R. (1999). Lateral stability, sensorimotor function and falls in older people. *Journal of the American Geriatrics Society,* 47, 1077–1081

Lord, S.R., Menz, H.B., & Tiedemann, A. (2003). A physiological profile approach to falls risk assessment and prevention. *Physical Therapy*, 83 (3), 237-252.

Lord, S.R., & Ward, J.A. (1994). Age-associated differences in sensori-motor function and balance in community dwelling women. *Age and Ageing*, 23 (6), 452-460.

Loughney, L., West, M., Pintus, S., Lythgoe, D., Clark, E., Jack, S., & Torella, F. (2013). Comparison of oxygen uptake during arm or leg cardiopulmonary exercise testing in vascular surgery patients and control subjects. *British Journal of Anaesthesia*, 370

Low Choy, N. L., Brauer, S. G., & Nitz, J. C. (2007). Age-Related Changes in Strength and Somatosensation during Midlife. *Annals of the New York Academy of Sciences*, 1114 (1), 180-193.

Lyons, S., Richardson, M., Bishop, P., Smith, J., Heath, H. & Giesen, J. (2007). Excess post-exercise oxygen consumption in untrained men following exercise of equal energy expenditure: comparisons of upper and lower body exercise. *Diabetes, Obesity and Metabolism*, 9, 889-894.

Maciaszek, J., Stemplewski, R., & Osiński, W. (2011). Effects of systolic blood pressure (SBP) on a postural stability in elderly men subjected to moderate exercise. *Archives of Gerontology and Geriatrics*, 53 (1), e37-e40.

Mademli, L., Arampatzis, A., & Karamanidis, K. (2008). Dynamic stability control in forward falls: postural corrections after muscle fatigue in young and older adults. *European Journal of Applied Physiology*, 103 (3), 295-306.

Madigan, M. L., Davidson, B. S., & Nussbaum, M. A. (2006). Postural sway and joint kinematics during quiet standing are affected by lumbar extensor fatigue. *Human Movement Science*, 25 (6), 788-799.

Magel, J.R., Foglia, G.F., Mcardle, W. D., Gutin, B. & Pechar, G.S. (1975). Specificity of swim training on maximum oxygen uptake. *Journal of Applied Physiology*, 38, 151-155.

Magnusson, M., Enbom, H., Johansson, R. & Pyykko, I. (1990). Significance of pressor input from the human feet in anterior-posterior postural control. *Acta Otolaryngol (Stockh),* 110, 182-8.

Maire, J., Dugué, B., Faillenet-Maire, A.F., Smolander, J., Tordi, N., Parratte, B., Grange, C., and Rouillon, J.D. (2006). Influence of a 6-week arm exercise program on walking ability and health status after hip arthroplasty: a 1-year follow-up pilot study. Journal of Rehabilitation and Research Development, 43 (4), 445-50.

Maki, B. E., Holliday, P. J., & Topper, A. K. (1994). A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. *Journal of Gerontology*, 49 (2), M72-M84.

Maki, B.E., Holliday, P.J., Fernie, G.R., 1990. Aging and postural control. A comparison of spontaneous- and induced-sway balance tests. *Journal of American Geriatrics Society*, 38, 1–9.

Mallau, S., Vaugoyeau, M., & Assaiante, C. (2010). Postural strategies and sensory integration: no turning point between childhood and adolescence. *PloS one*, 5 (9), e13078.

Mancini, M., & Horak, F. B. (2010). The relevance of clinical balance assessment tools to differentiate balance deficits. *European Journal of Physical and Rehabilitation Medicine*, 46 (2), 239 – 248.

Marais, G., Dupont, L., Vanvelcenaher, J., Clarys, J. P. & Pelayo, P. (2004). Effects of spontaneously chosen crank rate variations on electromyographic responses in sub-maximal arm exercise in inexperienced subjects. *European Journal of Applied Physiology*, 92, 598-601.

Massion, J. (1994). Postural control system. *Current Opinion in Neurobiology*, 4 (6), 877-887.

Massion, J. (1992). Movement, posture and equilibrium: interaction and coordination. *Progress in Neurobiology*, 38: 35–56.

Massion, J., Popov, K., Fabre, J.C., Rage, P. and Gurfinkel, V. (1997). Is the erect posture in microgravity based on the control of trunk orientation or center of mass position? *Experimental Brain Research*, 114, 384–389.

Mathiowetz, V., Rennells, C., & Donahoe, L. (1985). Effect of elbow position on grip and key pinch strength. *The Journal of Hand Surgery*, 10 (5), 694-697.

Mathias, S., Nayak, U. S., & Isaacs, B. (1986). Balance in elderly patients: the" get-up and go" test. *Archives of Physical Medicine and Rehabilitation*, 67 (6), 387-389.

Maurer, C., & Peterka, R. J. (2005). A new interpretation of spontaneous sway measures based on a simple model of human postural control. *Journal of Neurophysiology*, 93 (1), 189-200.

McCombe Waller, S., and Prettyman, M.G. (2012) .Arm training in standing also improves postural control in participants with chronic stroke. *Gait & Posture,* 36 (3), 419-24.

McGill, S. M., Childs, A., & Liebenson, C. (1999). Endurance times for low back stabilization exercises: clinical targets for testing and training from a normal database. *Archives of Physical Medicine and Rehabilitation*, 80 (8), 941-944.

McIlroy, W. E., & Maki, B. E. (1997). Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. *Clinical Biomechanics*, 12 (1), 66-70.

Mello, R.G.T., de Oliveria, L.F., & Nadal, J. (2010). Effects of maximal oxygen uptake test and prolonged cycle ergometer exercise on the quiet standing control. *Gait and Posture*, 32, 220-225.

Meredith, C.N., Frontera, W.R., Fisher, E.C., Hughes, V.A., Herland, J.C., Edwards, J., & Evans, W.J. (1989). Peripheral effects of endurance training in young and old subjects. *Journal of Applied Physiology*, 66 (6), 2844-9

Mergner, T., Maurer, C., & Peterka, R. J. (2003). A multisensory posture control model of human upright stance. *Progress in Brain Research*, 142, 189-201

Metter, E. J., Talbot, L. A., Schrager, M. & Conwit, R. A. (2004). Arm-cranking muscle power and arm isometric muscle strength are independent predictors of all-cause mortality in men. *Scandinavian Journal of Medicine and Science in Sports*, 14, 203-203.

Meyer, P., Oddsson, L,. & De Luca, C. (2004). The role of plantar cutaneous sensation in unperturbed standing. *Experimental Brain Research,* 156, 505-12.

Moore, S. P., Rushmer, D. S., Windus, S. L., & Nashner, L. M. (1988). Human automatic postural responses: responses to horizontal perturbations of stance in multiple directions. *Experimental Brain Research*, 73 (3), 648-658.

Morasso, P.G., & Schieppati, M. (1999). Can muscle stiffness alone stabilize upright standing? *Journal of Neurophysiology*, 82 (3), 1622-1626.

Moreland, J. D., Richardson, J. A., Goldsmith, C. H., & Clase, C. M. (2004). Muscle weakness and falls in older adults: a systematic review and meta-analysis. *Journal of the American Geriatrics Society*, 52 (7), 1121-1129.

Moritani, T., & Devries, H. A. (1980). Potential for gross muscle hypertrophy in older men. *Journal of Gerontology*, 35 (5), 672-682.

Morris, S.L., & Allison, G.T., (2006). Effects of abdominal muscle fatigue on anticipatory postural adjustments associated with arm raising. *Gait & Posture,* 24, 342–348.

Muir, S.W., Gopaul, K., & Odasso, M.M.M. (2012). The role of cognitive impairment in fall risk among older adults: a systematic review and meta-analysis. *Age and Ageing*, 41 (3), 299-308.

Myers, J., Arena, R., Franklin, B., Pina, I., Kraus, W. E., McInnis, K., & Balady, G. J. (2009). Recommendations for Clinical Exercise Laboratories A Scientific Statement From the American Heart Association. *Circulation*, 119 (24), 3144-3161.

Nagai, K., Yamada, M., Mori, S., Tanaka, B., Uemura, K., Aoyama, T., & Tsuboyama, T. (2013). Effect of the muscle coactivation during quiet standing on dynamic postural control in older adults. *Archives of Gerontology and Geriatrics*, 56 (1), 129-133.

Nagai, K., Yamada, M., Uemura, K., Yamada, Y., Ichihashi, N., & Tsuboyama, T. (2011). Differences in muscle coactivation during postural control between healthy older and young adults. *Archives of Gerontology and Geriatrics*, 53 (3), 338-343.

Nagy, E., Toth, K., Janositz, G., Kovacs, G., Feher-Kiss, A., Angyan, L., Horvath, G., (2004). Postural control in athletes participating in an ironman triathlon. *European Journal of Applied Physiology*, 92, 407–413.

Nardone, A., Tarantola, J., Galante, M., Schieppati, M., 1998. Time course of stabilometric changes after a strenuous treadmill exercise. *Archives of Physical Medicine and Rehabilitation*, 79, 920–924.

Nardone, A., Taratola, J., Giordano, A., & Schieppati, M. (1997). Fatigue effects on body sway. *Electroencephalography and Clinical Neurophysiology*, 105, 309-320.

Nashner LM: *Analysis of movement control in man using the movable platform*. In Desmedt JE (ed): Motor Control Mechanisms in Health and Disease: Advances in Neurology. New York, NY, Raven Press, 1983, vol 39, pp 607-619

Nashner, L. M., & McCollum, G. (1985). The organization of human postural movements: a formal basis and experimental synthesis. *Behavioral and Brain Sciences*, 8 (1), 135-150.

Nashner, L. M., Black, F.O., & Wall, C. I.I.I. (1982). Adaptation to altered support and visual conditions during stance: patients with vestibular deficits. *The Journal of Neuroscience*, 2 (5), 536-544.

Nelson, M.E., Rejeski, W.J., Blair, S.N., et al. (2007). Physical activity and public health in older adults: recommendation from the American College of Sports Medicine and the American Heart Association. *Circulation,* 116: 1094–105.

Nelson, M.E., Layne, J.E., Bernstein, M.J., Nuernberger, A., Castaneda, C., Kaliton, D., et al. (2004). The effects of multidimensional home-based exercise on functional performance in elderly people. *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, 59 (2), 154–60.

Nes, B. M., Janszky, I., Wisløff, U., Støylen, A., & Karlsen, T. (2013). Age-predicted maximal heart rate in healthy subjects: The HUNT Fitness Study. *Scandinavian Journal of Medicine & Science in Sports*, *23* (6), 697-704.

Neville, V., Pain, M. T., Kantor, J., & Folland, J. P. (2010). Influence of crank length and crank-axle height on standing arm-crank (grinding) power. *Medicine and Science in Sports and Exercise*, 42, 381–387.

Neville, V., Zaher, N., Pain, M. T., & Folland, J. P. (2009). Lower limb influence on standing arm-cranking (‘‘grinding’’). *International Journal of Sports Medicine*, 30, 713–718.

Newell, K. M., van Emmerik, R. E., Lee, D., & Sprague, R. L. (1993). On postural stability and variability. *Gait & Posture*, 1(4), 225-230.

Newton, R. A. (2001). Validity of the Multi-Directional Reach Test A Practical Measure for Limits of Stability in Older Adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *56* (4), M248-M252.

Nikolić, M., Šoić Vranić, T., Arbanas, J., Cvijanović, O., & Bajek, S. (2010). Muscle loss in elderly. *Collegium antropologicum*, *34* (2), 105-108.

Nitz, J. C., & Choy, N. L. (2004). The efficacy of a specific balance-strategy training programme for preventing falls among older people: a pilot randomised controlled trial. *Age and Ageing*, 33 (1), 52-58.

Nolan, M., Nitz, J., Choy, N. L., & Illing, S. (2010). Age-related changes in musculoskeletal function, balance and mobility measures in men aged 30-80 years. *The Aging Male*, 13(3), 194-201.

Nordahl SH, Aasen T, Dyrkorn BM, Eidsvik S, Molvaer OI (2000). Static stabilometry and repeated testing in a normal population. *Aviation and Space Environmental Medicine*, 71, 889–93.

Ntousis, T., Mandalidis, D., Chronopoulos, E., & Athanasopoulos, S. (2013). EMG activation of trunk and upper limb muscles following experimentally-induced overpronation and oversupination of the feet in quiet standing. *Gait & Posture*, 37 (2), 190-194.

Nussbaum, M.A. (2003). Postural stability is compromised by fatiguing overhead work. *A Journal for the Science of Occupational and Environmental Health and Safety*, 64, 56-61.

Olmsted, L. C., Carcia, C. R., Hertel, J., & Shultz, S. J. (2002). Efficacy of the Star Excursion Balance Tests in detecting reach deficits in subjects with chronic ankle instability. *Journal of Athletic Training*, 37 (4), 501.

O'Sullivan, P. B., Grahamslaw, K. M., Kendell, M., Lapenskie, S. C., Möller, N. E., & Richards, K. V. (2002). The effect of different standing and sitting postures on trunk muscle activity in a pain-free population. *Spine*, 27, (11), 1238-1244.

Ogawa, T., Spina, R. J., Martin, W. H., Kohrt, W. M., Schechtman, K. B., Holloszy, J. O., & Ehsani, A. A. (1992). Effects of aging, sex, and physical training on cardiovascular responses to exercise. *Circulation*, 86 (2), 494-503.

Olivier, N., Legrand, R., Rogez, J., Berthoin, S., Prieur, F., & Weissland, T. (2008). One-leg cycling versus arm cranking: Which is most appropriate for physical conditioning after knee surgery? *Archives of Physical Medicine and Rehabilitation*, 89 (3), 508-512.

Orr, J. L., Williamson, P., Anderson, W., Ross, R., McCafferty, S., & Fettes, P. (2013). Cardiopulmonary exercise testing: arm crank vs cycle ergometry. *Anaesthesia*, 68 (5), 497-501.

Orr, R., De Vos, N. J., Singh, N. A., Ross, D. A., Stavrinos, T. M., & Fiatarone-Singh, M. A. (2006). Power training improves balance in healthy older adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 61 (1), 78-85

Paillard, T. (2012). Effects of general and local fatigue on postural control: a review. *Neuroscience & Biobehaviour Reviews*, 36, 162-176.

Palmer, G. S., Borghouts, L. B., Noakes, T. D., & Hawley, J. A. (1999). Metabolic and performance responses to constant-load vs. variable-intensity exercise in trained cyclists. *Journal of Applied Physiology*, 87 (3), 1186-1196.

Panzer, V. P., Bandinelli, S., & Hallett, M. (1995). Biomechanical assessment of quiet standing and changes associated with aging. *Archives of Physical Medicine and Rehabilitation*, 76 (2), 151-157

Pardasaney, P. K., Latham, N. K., Jette, A. M., Wagenaar, R. C., Ni, P., Slavin, M. D., & Bean, J. F. (2012). Sensitivity to change and responsiveness of four balance measures for community-dwelling older adults. *Physical Therapy*, 92 (3), 388-397.

Parreira, R. B., Amorim, C. F., Gil, A. W., Teixeira, D. C., Bilodeau, M., & da Silva, R. A. (2013). Effect of trunk extensor fatigue on the postural balance of elderly and young adults during unipodal task. *European Journal of Applied Physiology*, 113 (8), 1989-1996.

Park, S., Horak, F. B., & Kuo, A. D. (2004). Postural feedback responses scale with biomechanical constraints in human standing. *Experimental Brain Research*, 154 (4), 417-427.

Park, H., Kim, K. J., Komatsu, T., Park, S. K., & Mutoh, Y. (2008). Effect of combined exercise training on bone, body balance, and gait ability: a randomized controlled study in community-dwelling elderly women. *Journal of Bone and Mineral Metabolism*, 26 (3), 254-259.

Paschalis, V., Nikolaidis, M. G., Giakas, G., Jamurtas, A. Z., Owolabi, E. O., & Koutedakis, Y. (2008). Position sense and reaction angle after eccentric exercise: the repeated bout effect. *European Journal of Applied Physiology*, 103(1), 9-18.

Paterson, D.H., Cunningham, D.A., Koval, J.J., & St. Croix, C.M. (1998) Aerobic fitness in a population of independently living men and women aged 55–86 years. *Medicine and Science in Sports and Exercise*, 31, 1813–1820.

Pavol, M. J. (2005). Detecting and understanding differences in postural sway. Focus on “A new interpretation of spontaneous sway measures based on a simple model of human postural control”. *Journal of Neurophysiology*, 93 (1), 20-21.

Pendergast, D. R. (1989). Cardiovascular, respiratory, and metabolic responses to upper body exercise. *Medicine & Science in Sports & Exercise*, 21, S121-S125.

Pendergast, D., Cerretelli, P. & Rennie, D. W. (1979). Aerobic and glycolytic metabolism in arm exercise. *Journal of Applied Physiology: Respiratory, Environmental And Exercise Physiology*, 47, 754-760.

Perini, R., Fisher, N., Veicsteinas, A., & Pendergast, D. R. (2002). Aerobic training and cardiovascular responses at rest and during exercise in older men and women. *Medicine and Science in Sports and Exercise*, 34 (4), 700-708.

Perrin, P. P., Gauchard, G. C., Perrot, C., & Jeandel, C. (1999). Effects of physical and sporting activities on balance control in elderly people. *British Journal of Sports Medicine*, 33 (2), 121-126.

Perry, S.D. (2006). Evaluation of age-related plantar-surface insensitivity and onset age of advanced insensitivity in older adults using vibratory and touch sensation tests. *Neuroscience Letters*. 392, 62–67

Peterka, R. J. (2002). Sensorimotor integration in human postural control. *Journal of Neurophysiology*, 88 (3), 1097-1118.

Peterka, R. J., & Loughlin, P. J. (2004). Dynamic regulation of sensorimotor integration in human postural control. *Journal of Neurophysiology*, 91 (1), 410-423.

Piirtola, M., & Era, P. (2006). Force platform measurements as predictors of falls among older people–a review. *Gerontology*, 52 (1), 1-16.

Pimental, N. A., Sawka, M. N., Billings, D. S. & Trad, L. A. (1984). Physiological responses to prolonged upper body exercise. *Medicine & Science in Sports & Exercise*, 16, 360-365.

Pincivero, D. M., Gandaio, C. B., & Ito, Y. (2003). Gender-specific knee extensor torque, flexor torque, and muscle fatigue responses during maximal effort contractions. *European Journal of Applied Physiology*, 89 (2), 134-141.

Pinsault, N., & Vuillerme, N. (2009). Test–retest reliability of centre of foot pressure measures to assess postural control during unperturbed stance. *Medical Engineering and Physics,* 31, 276–286.

Podsiadlo, D., & Richardson, S. (1991). The timed" Up & Go": a test of basic functional mobility for frail elderly persons. *Journal of the American geriatrics Society*, 39 (2), 142-148.

Pogliaghi, S., Terziotti, P., Cevese, A., Balestreri, F., & Schena, F. (2006). Adaptions to endurance training in the healthy elderly: arm cranking versus leg cycling. *European Journal of Applied Physiology*, 97, 723-731.

Powell, L. E., & Myers, A. M. (1995). The activities-specific balance confidence (ABC) scale. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 50 (1), M28-M34.

Powers, S.K., Beadle, R.E., & Mangum, M. (1984). Exercise efficiency during arm ergometry: effects of speed and work rate. *Work*, 10, 100.

Price, M. & Campbell, I.G. (1999). Thermoregulatory responses of spinal cord injured and able-bodied athletes to prolonged upper body exercise and recovery. *Spinal Cord,* 37 (11), 772-9.

Price, M. J., Bottoms, L., Smith, P. M., & Nicholettos, A. (2011). The effects of an increasing versus constant crank rate on peak physiological responses during incremental arm crank ergometry. *Journal of Sports Sciences*, 29(3), 263-269.

Price, M. J., Collins, L., Smith, P. M. & Goss-Sampson, M. (2007). The effects of cadence and power output upon physiological and biomechanical responses to incremental arm-crank ergometry. *Applied Physiology, Nutrition & Metabolism*, 32, 686-692.

Price, M.J. (2006). Thermoregulation during exercise in individuals with Spinal Cord injuries. *International Journal of Sports Medicine*, 36 (10), 863-879.

Price, M.J., and Campbell, I.G. (2003) 'Effects of Spinal Cord lesion level upon thermoregulation during exercise in the heat'. *Medicine and Science in Sports and Exercise,* 35 (7), 1100-1107.

Rainoldi, A., Melchiorri, G., & Caruso, I. (2004). A method for positioning electrodes during surface EMG recordings in lower limb muscles. *Journal of Neuroscience Methods*, 134 (1), 37-43.

Ramsbottom, R., Ambler, A., Potter, J., Jordan, B., Nevill, A., Williams, C. (2004). The effect of 6 months training on leg power, balance, and functional mobility of independently living adults over 70 years old. *Journal of Aging & Physical Activity*, 12 (4), 497–510.

Raymakers, J. A., Samson, M. M., & Verhaar, H. J. J. (2005). The assessment of body sway and the choice of the stability parameter (s). *Gait & Posture*, 21(1), 48-58.

Reybrouck, T., Heigenhauser, G. F. & Faulkner, J. A. (1975). Limitations to maximum oxygen uptake in arm and leg, and combined arm-leg ergometry. *Journal of Applied Physiology*, 38, 774-779.

Ribeiro, F., & Oliveira, J. (2007). Aging effects on joint proprioception: the role of physical activity in proprioception preservation. *European Review of Aging and Physical Activity*, 4 (2), 71-76.

Richardson, J. K., Ashton-Miller, J. A., Lee, S. G., & Jacobs, K. (1996). Moderate peripheral neuropathy impairs weight transfer and unipedal balance in the elderly. *Archives of Physical Medicine and Rehabilitation*, 77 (11), 1152-1156.

Rikli, R.E., & Jones, C.J. (2013). *Senior fitness test manual*. Champaign: Human Kinetics.

Ringsberg, K., Gerdhem, P., Johansson, J., & Obrant, K. J. (1999). Is there a relationship between balance, gait performance and muscular strength in 75-year-old women? *Age and Ageing*, 28 (3), 289-293.

Rissel, C., Passmore, E., Mason, C., and Merom, D. (2013). Two pilot studies of the effect of bicycling on balance and leg strength among older adults. Journal of Environmental Public Health (ePub)

Rivera, A.M., Pels, A.E., Sady, S.P., Sady, M.A., Cullinane, E.M., & Thompson, P.D. (1989). Physiological factors associated with the lower maximal oxygen consumption of master runners. *Journal of Applied Physiology*, 66, 949–954

Roberts, H. C., Denison, H. J., Martin, H. J., Patel, H. P., Syddall, H., Cooper, C., & Sayer, A. A. (2011). A review of the measurement of grip strength in clinical and epidemiological studies: towards a standardised approach. *Age and Ageing*, 51

Rocchi, M. B. L., Sisti, D., Ditroilo, M., Calavalle, A., & Panebianco, R. (2005). The misuse of the confidence ellipse in evaluating statokinesigram. *Italian Journal of Sport Science*, 12 (2), 169-172.

Rogers, M. W., & Mille, M. L. (2003). Lateral stability and falls in older people. *Exercise and Sport Sciences Reviews*, 31 (4), 182-187.

Roll, J.P., Roll, R. (1988). From eye to foot. A proprioceptive chain involved in postural control, in: B. Amblard, A. Berthoz, F. Clarac, F. (Eds.), Posture and Gait. Elsevier, Amsterdam, pp. 155-164.

Rubenstein, L. Z., & Josephson, K. R. (2006). Falls and their prevention in elderly people: what does the evidence show?. *Medical Clinics of North America*, 90 (5), 807-824.

Rugelj, D., & Sevšek, F. (2011). The effect of load mass and its placement on postural sway. *Applied ergonomics*, 42 (6), 860-866.

Runge, C. F., Shupert, C. L., Horak, F. B., & Zajac, F. E. (1999). Ankle and hip postural strategies defined by joint torques. *Gait & Posture*, 10(2), 161-170.

Russ, D. W., Lanza, I. R., Rothman, D., & Kent-Braun, J. A. (2005). Sex differences in glycolysis during brief, intense isometric contractions. *Muscle & Nerve*, 32 (5), 647-655.

Ruzic, L., Prpic, T., Madarevic, T., Radman, I., Tudor, A., Rakovac, I., & Sestan, B. (2014). Physiological load and posture control thresholds. *Gait & Posture*, 39 (1), 415-419.

Santos, B. R., Delisle, A., Larivière, C., Plamondon, A., & Imbeau, D. (2008). Reliability of centre of pressure summary measures of postural steadiness in healthy young adults. *Gait & Posture*, 27(3), 408-415.

Santos, M. J., Kanekar, N., & Aruin, A. S. (2010). The role of anticipatory postural adjustments in compensatory control of posture: 1. Electromyographic analysis. *Journal of Electromyography and Kinesiology*, 20 (3), 388-397.

Santos, M. J., Kanekar, N., & Aruin, A. S. (2010). The role of anticipatory postural adjustments in compensatory control of posture: 2. *Biomechanical analysis. Journal of Electromyography and Kinesiology*, 20 (3), 398-405.

Sawka, M. N. (1986). Physiology of upper body exercise. *Exercise and Sport Sciences Reviews*, 14, 175-211.

Sawka, M. N., Foley, M. E., Pimental, N. A., Toner, M. M. & Pandolf, K. B. (1983). Determination of maximal aerobic power during upper-body exercise. *Journal of Applied Physiology: Respiratory, Environmental and Exercise Physiology*, 54, 113-117.

Sawka, M.N., Gonzalez, R.R., Drolet, L.L., & Pandolf, K.B. (1984). Heat exchange during upper- and lower-body exercise. *Journal of Applied Physiology: Respiratory, Environmental and Exercise Physiology*, 57 (4), 1050-4.

Sawka, M.N., Miles, D.S., Petrofsky, J.S., Wilde, S.W., & Glaser, R.M. (1982). Ventilation and acid-base equilibrium for upper body and lower body exercise. *Aviation and Space Environmental Medicine*, 53 (4), 354-9.

Saxton, J.M., Zwierska, I., Blagojevic, M., Choksy, S.A., Nawaz, S., and Pockley, A.G. (2011). Upper- versus lower-limb aerobic exercise training on health-related quality of life in patients with symptomatic peripheral arterial disease. *Journal of Vascular Surgery*, 53 (5), 1265-73.

Schellenberg, K. L., Lang, J. M., Chan, K. M., & Burnham, R. S. (2007). A clinical tool for office assessment of lumbar spine stabilization endurance: prone and supine bridge maneuvers. *American Journal of Physical Medicine & Rehabilitation*, 86 (5), 380-386.

Schepens, B., & Drew, T. (2004). Independent and convergent signals from the pontomedullary reticular formation contribute to the control of posture and movement during reaching in the cat. *Journal of Neurophysiology*, 92 (4), 2217-2238.

Schieppati, M., Nardone, A,. & Schmid, M. (2003). Neck muscle fatigue affects postural control in man. Neuroscience, 121, 277-285

Schmidt, R. A., & Lee, T. (1988). *Motor Control and Learning*, (5th Ed.). Champaign: Human Kinetics

Schneider, D. A., Wing, A. N. & Morris, N. R. (2002). Oxygen uptake and heart rate kinetics during heavy exercise: a comparison between arm cranking and leg cycling. *European Journal of Applied Physiology*, 88, 100-106.

Seals, D. R., & Mullin, J. P. (1982). VO_2 Max in Variable Type Exercise Among Well-Trained Upper Body Athletes. *Research Quarterly for Exercise and Sport*, 53 (1), 58-63.

Seals, D. R., Hagberg, J. M., Spina, R. J., Rogers, M. A., Schechtman, K. B., & Ehsani, A. A. (1994). Enhanced left ventricular performance in endurance trained older men. *Circulation*, 89 (1), 198-205.

Secher, N. H., Ruberg-Larsen, N., Binkhorst, R. A. & Bonde-Petersen, F. (1974). Maximal oxygen uptake during arm cranking and combined arm plus leg exercise. *Journal of Applied Physiology*, 36, 515-518.

Sell, T. C. (2012). An examination, correlation, and comparison of static and dynamic measures of postural stability in healthy, physically active adults. *Physical Therapy in Sport*, *13* (2), 80-86.

Serrador, J. M., Lipsitz, L. A., Gopalakrishnan, G. S., Black, F. O., & Wood, S. J. (2009). Loss of otolith function with age is associated with increased postural sway measures. *Neuroscience Letters*, 465 (1), 10-15.

Shaffer S & Harrison A (2007) Aging of the somatosensory system: A translational perspective. *Physical Therapy* **87** 193-207.

Sharpe, M.H., & Miles, T.S. (1993) Position sense at the elbow after fatiguing contractions. Experimental Brain Research, 94 (1), 179–182

Sheldon J.H. (1963). The effect of age on the control of sway. *Clinical Gerontologist*, 5,129-38.

Sherrington, C.S. (1906). *The integrative action of the nervous system*. New Haven, CT: Yale University Press.

Shumway-Cook, A., & Woollacott, M. H. (2011). *Motor control: translating research into clinical practice.* Lippincott Williams & Wilkins

Shumway-Cook, A., Brauer, S., & Woollacott, M. (2000). Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. *Physical Therapy*, 80 (9), 896-903.

Skinner, H. B., Wyatt, M. P., Hodgdon, J. A., Conard, D. W., & Barrack, R. L. (1986). Effect of fatigue on joint position sense of the knee. *Journal of Orthopaedic Research*, 4 (1), 112-118.

Smith, M.D., Chang, A.T., Seale, H.E., Walsh, J.R, & Hodge, P.W. (2010) Balance is impaired in people with chronic obstructive pulmonary disease. *Gait & Posture* 31, 456-460.

Smith, P. M. & Price, M. J. (2007) Upper-body exercise. In: E. M. Winter., A. M. Jones., R. C. R. Davison., P. D. Bromley. & T. H. Mercer. (eds). BASES Sport and Exercise Physiology Testing Guidelines. pp. 138 - 144.

Smith, P. M., Chapman, M. L., Hazlehurst, K. E. & Goss-Sampson, M. A. (2008) The influence of crank configuration on muscle activity and torque production during arm crank ergometry. *Journal of Electromyography & Kinesiology*, 18, 598-605.

Smith, P. M., Doherty, M. & Price, M. J. (2007a) The effect of crank rate strategy on peak aerobic power and peak physiological responses during arm crank ergometry. *Journal of Sports Sciences*, 25, 711-718.

Smith, P. M., Doherty, M., Drake, D., & Price, M. J. (2004). The influence of step and ramp type protocols on the attainment of peak physiological responses during arm crank ergometry. *International Journal of Sports Medicine*, 25 (8), 616-621

Smith, P. M., Price, M. J., & Doherty, M. (2001). The influence of crank rate on peak oxygen consumption during arm crank ergometry. *Journal of Sports Sciences*, *19*(12), 955-960.

So, R.C., Ng, J.K.F., & Ng, G.Y. (2005). Muscle recruitment pattern in cycling: a review. *Physical Therapy in Sport*, 6 (2), 89-96

Soderberg, G. L., & Knutson, L. M. (2000). A guide for use and interpretation of kinesiologic electromyographic data. *Physical Therapy*, 80 (5), 485-498.

Speers, R. A., Kuo, A. D., & Horak, F. B. (2002). Contributions of altered sensation and feedback responses to changes in coordination of postural control due to aging. *Gait & Posture*, 16 (1), 20-30.

Spina, R. J., Ogawa, T., Kohrt, W. M., Martin, W., Holloszy, J. O., & Ehsani, A. A. (1993). Differences in cardiovascular adaptations to endurance exercise training between older men and women. *Journal of Applied Physiology*, 75, 849-849

Springer, B.K., & Pincivero, D.M. (2009). The effects of localized muscle and whole-body fatigue on single-leg balance between healthy men and women. *Gait & Posture*, 30, 50-54.

Stambolieva, K., Diafas, V., Bachev, V., Christova, L., & Gatev, P. (2012). Postural stability of canoeing and kayaking young male athletes during quiet stance. *European Journal of Applied Physiology*, 112 (5), 1807-1815

Stamford, B. A., Cuddihee, R. W., Moffatt, R. J., & Rowland, R. (1978). Task specific changes in maximal oxygen uptake resulting from arm versus leg training. *Ergonomics*, 21 (1), 1-9

Steffen, T. M., Hacker, T. A., & Mollinger, L. (2002). Age-and gender-related test performance in community-dwelling elderly people: Six-Minute Walk Test, Berg Balance Scale, Timed Up & Go Test, and gait speeds. *Physical Therapy*, *82* (2), 128-137

Stel, V. S., Smit, J. H., Pluijm, S. M., & Lips, P. (2003). Balance and mobility performance as treatable risk factors for recurrent falling in older persons. *Journal of Clinical Epidemiology*, 56(7), 659-668.

Stemplewski R, Maciaszek J, Salamon A, Tomczak M, Osinski W (2012). Effect of moderate physical exercise on postural control among 64-74 years old men. *Archives of Gerontology and Geriatrics*, 54, 279-283.

Stemplewski, R., Maciaszek, J., Tomczak, M., Szeklicki, R., Sadowska, D., & Osinski, W. (2013). Habitual physical activity as a determinant of the effect of moderate physical exercise on postural control in older men. *American Journal of Mens Health*. 7, 58-65.

Stenberg, J., Astrand, P. O., Ekblom, B., Royce, J. & Saltin, B. (1967). Hemodynamic response to work with different muscle groups, sitting and supine. *Journal of Applied Physiology*, 22, 61-70.

Stevens, J. A., & Sogolow, E. D. (2005). Gender differences for non-fatal unintentional fall related injuries among older adults. *Injury Prevention*, 11 (2), 115-119

Strang, A.J., Berg, W.P., Hieronymus, M., (2009). Fatigue-induced early onset of anticipatory postural adjustments in non-fatigued muscles: support for a centrally mediated adaptation. *Experimental Brain Research*, 197, 245–254.

Strang, A.J., Choi, H.J., Berg, W.P., (2008). The effect of exhausting aerobic exercise on the timing of anticipatory postural adjustments. *Journal of Sports Medicine and Physical Fitness*, 48, 9–16.

Surenkok, O., Kin-Isler, A., Aytar, A., Gültekin, Z., (2008). Effect of trunk-muscle fatigue and lactic acid accumulation on balance in healthy subjects. *Journal of Sports Rehabilitation*, 17, 380–386.

Susheela, A. K., & Walton, J. N. (1969). Note on the distribution of histochemical fibre types in some normal human muscles: A study on autopsy material. *Journal of the Neurological Sciences*, 8 (2), 201-207.

Taimela, S., Kankaanpää, M., & Luoto, S. (1999). The effect of lumbar fatigue on the ability to sense a change in lumbar position: a controlled study. *Spine*, 24 (13), 1322.

Tanaka, H., Monahan, K.D., Seals, D.R. (2001). Age-predicted maximal heart rate revisited. Journal of the American College of Cardiology, 37, 153-156.

Tanaka, H., De Souza, C.A., Jones, P., et al. (1997). Greater rate of decline in maximal aerobic capacity with age in physically active vs sedentary healthy women. Journal of Applied Physiology, 83, 1947–1953.

Tanaka, H., & Uetake, T. (2005). Characteristics of postural sway in older adults standing on a soft surface. *J. Hum. Ergol (Tokyo)*, 34, 35-40.

Tang, P. F., & Woollacott, M. H. (1998). Inefficient postural responses to unexpected slips during walking in older adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 53 (6), M471-M480.

Taylor, J. L., Butler, J. E., & Gandevia, S. C. (2000). Changes in muscle afferents, motoneurons and motor drive during muscle fatigue. *European Journal of Applied Physiology*, 83, 106-115.

Taylor, A. W., & Johnson, M. J. (2008). *Physiology of exercise and healthy aging*. Champaign: Human Kinetics

Tew, G., Nawaz, S., Zwierska, I., & Saxton, J.M. (2009). Limb-specific and cross-transfer effects of arm crank exercise training in patients with symptomatic peripheral arterial disease. *Clinical Science*, 117, 405-413.

Thomas, K. S., VanLunen, B. L., & Morrison, S. (2013). Changes in postural sway as a function of prolonged walking. *European Journal of Applied Physiology*, 113 (2), 497-508.

Tinetti, M. E. (1986). Performance-oriented assessment of mobility problems in elderly patients. *Journal of the American Geriatrics Society*, 34, 119-126.

Tinetti, M. E., Speechley, M., & Ginter, S. F. (1988). Risk factors for falls among elderly persons living in the community. *New England journal of medicine*, 319 (26), 1701-1707.

Toner, M. M., Sawka, M. N., Levine, L., & Pandolf, K. B. (1983). Cardiorespiratory responses to exercise distributed between the upper and lower body. *Journal of Applied Physiology*, 54, 1403-1407.

Tordi, N., Belli, A., Mougin, F., Rouillon, J. R. & Gimenez, M. (2001). Specific and transfer effects induced by arm or leg training. *International Journal of Sports Medicine,* 22, 517-524.

Treat-Jacobson, D., Bronas, U.G., and Leon, A.S. (2009). Efficacy of arm-ergometry versus treadmill exercise training to improve walking distance in patients with claudication. *Vascular Medicine*, 14 (3), 203-13

Tropp, H., & Odenrick, P. (1988). Postural control in single-limb stance. *Journal of Orthopaedic Research*, 6 (6), 833-839.

Tropp, H., Samuelsson, K., and Jorfeldt, L. (1997). Power output for wheelchair driving on a treadmill compared with arm crank ergometry. *British Journal of Sports Medicine*, 31(1),41-4.

Tucker, M. G., Kavanagh, J. J., Barrett, R. S., & Morrison, S. (2008). Age-related differences in postural reaction time and coordination during voluntary sway movements. *Human Movement Science*, 27 (5), 728-737.

Tulppo, M. P., Makikallio, T. H., Laukkanen, R. T. & Huikuri, H. V. (1999). Differences in autonomic modulation of heart rate during arm and leg exercise. *Clinical Physiology*, 19, 294-299.

Twist, C., Gleeson, N., Eston, R., (2008). The effects of plyometric exercise on unilateral balance performance. *Journal of Sports Science*, 26, 1073–1080.

Uchiyama, M., & Demura, S. (2008). Low visual acuity is associated with the decrease in postural sway. *The Tohoku Journal of Experimental Medicine*, 216 (3), 277-285.

Van Kan, G. A., Rolland, Y., Andrieu, S., Bauer, J., Beauchet, O., Bonnefoy, M., & Vellas, B. (2009). Gait speed at usual pace as a predictor of adverse outcomes in community-dwelling older people an International Academy on Nutrition and Aging (IANA) Task Force. *The Journal of Nutrition, Health & Aging*, 13 (10), 881-889

van Emmerik, R. E., & van Wegen, E. E. (2002). On the functional aspects of variability in postural control. *Exercise and Sport Sciences Reviews*, 30 (4), 177-183.

Van Wegen, E. E. H., Van Emmerik, R. E. A., & Riccio, G. E. (2002). Postural orientation: age-related changes in variability and time-to-boundary. *Human Movement Science*, 21 (1), 61-84.

Vianna, L. C., Oliveira, R. B., & Araujo, C. G. S. (2007). Age-related decline in handgrip strength differs according to gender. *The Journal of Strength & Conditioning Research*, 21 (4), 1310-1314.

Viccaro, L. J., Perera, S., & Studenski, S. A. (2011). Is timed up and go better than gait speed in predicting health, function, and falls in older adults? *Journal of the American Geriatrics Society*, *59* (5), 887-892.

Visser, J.E., Carpenter, M.G., van der, K.H., & Bloem, B.R. (2008). The clinical utility of posturography. *Clinical Neurophysiology*, 119, 2424–2436

Vissing K, Overgaard K, Nedergaard A, Fredsted A, Schjerling P. (2008). Effects of concentric and repeated eccentric exercise on muscle damage and calpain-calpastatin gene expression in human skeletal muscle. *European Journal of Applied Physiology*, 103, 323–3327.

Vokac, Z., Bell, H., Bautz-Holter, E. & Rodahl, K. (1975). Oxygen uptake/heart rate relationship in leg and arm exercise, sitting and standing. *Journal of Applied Physiology*, 39, 54-59.

Volianitis, S., Yoshiga, C. C., Nissen, P. & Secher, N. H. (2004). Effect of fitness on arm vascular and metabolic responses to upper body exercise. *American Journal of Physiology. Heart and Circulatory Physiology*, 286, 1736-1741.

Vuillerme N, Anziani B, Rougier P (2007). Trunk extensor muscles fatigue affects undisturbed postural control in young healthy adults. *Clinical Biomechanics*, 22, 489-494.

Vuillerme, N., & Boisgontier, M. (2008). Muscle fatigue degrades force sense at the ankle joint. *Gait & posture*, 28 (3), 521-524.

Vuillerme, N., & Hintzy, F. (2007). Effects of a 200 W-15 min cycling exercise on postural control during quiet standing in healthy young adults. Europe*an Journal of Applied Physiology*, 100, 169-175.

Vuillerme, N., Forestier, N., & Nougier, V. (2002). Attentional demands and postural sway: the effect of the calf muscles fatigue. *Medicine and Science in Sports and Exercise*, 34 (12), 1907-1912.

Vuillerme, N., Nougier, V., & Teasdale, N. (2002). Effects of lower limbs muscular fatigue on anticipatory postural adjustments during arm motions in humans. *Journal of Sports Medicine and Physical Fitness*, 42 (3), 289-294.

Vuillerme, N., Pinsault, N., & Vaillant, J. (2005). Postural control during quiet standing following cervical muscular fatigue: effects of changes in sensory inputs. *Neuroscience Letters*, 378 (3), 135-139.

Vuillerme, N., Sporbert, C., & Pinsault, N. (2009). Postural adaptation to unilateral hip muscle fatigue during human bipedal standing. *Gait & Posture*, 30 (1), 122-125.

Walker, R. D., Nawaz, S., Wilkinson, C. H., Saxton, J. M., Pockley, A. G., & Wood, R. F. (2000). Influence of upper-and lower-limb exercise training on cardiovascular function and walking distances in patients with intermittent claudication. *Journal of Vascular Surgery*, 31(4), 662-669.

Walsh, L. D., Hesse, C. W., Morgan, D. L., & Proske, U. (2004). Human forearm position sense after fatigue of elbow flexor muscles. *The Journal of Physiology*, 558(2), 705-715.

Watanabe, T., Owashi, K., Kanauchi, Y., Mura, N., Takahara, M., & Ogino, T. (2005). The short-term reliability of grip strength measurement and the effects of posture and grip span. *The Journal of Hand Surgery*, 30 (3), 603-609.

Wernick-Robinson, M., Krebs, D. E., & Giorgetti, M. M. (1999). Functional reach: does it really measure dynamic balance?. *Archives of Physical Medicine and Rehabilitation*, 80 (3), 262-269.

Whipp, B. J., & Wasserman, K. (1969). Efficiency of muscular work. *Journal of Applied Physiology*, 26 (5), 644-648.

Wiley, R. L., & Lind, A. R. (1975). Respiratory responses to simultaneous static and rhythmic exercises in humans. *Clinical Science and Molecular Medicine*, 49(5), 427-432

Williams, G. N., Higgins, M. J., & Lewek, M. D. (2002). Aging skeletal muscle: physiologic changes and the effects of training. *Physical Therapy*, 82 (1), 62-68

Windhurst, U. (2007). Muscle proprioceptive feedback and spinal networks. *Brain Research Bulletin*, 73, 155-202.

Winter, D.A., Prince, F., Franck, J.S., Powel, C., & Zabjek, K.F. (1996). Unified theory regarding A/P and M/L balance in quiet standing. Journal of Neurophysiology, 75, 2334–234.

Winter, D. A. (1995). Human balance and posture control during standing and walking. *Gait & Posture*, 3 (4), 193-214

Winter, D. A. (2009). *Biomechanics and motor control of human movement*. John Wiley & Sons

Winter, D. A., Patla, A. E., & Frank, J. S. (1990). Assessment of balance control in humans. *Med Prog Technol*, 16, 31-51.

Winter, D. A., Patla, A. E., Prince, F., Ishac, M., & Gielo-Perczak, K. (1998). Stiffness control of balance in quiet standing. *Journal of Neurophysiology*, 80 (3), 1211-1221.

Wojcik, L. A., Nussbaum, M. A., Lin, D., Shibata, P. A., & Madigan, M. L. (2011). Age and gender moderate the effects of localized muscle fatigue on lower extremity joint torques used during quiet stance. *Human Movement Science*, 30 (3), 574-583.

Wolf, B., Feys, H., De, W., Van der Meer, J., Noom, M., Aufdemkampe, G., et al. (2001). Effect of a physical therapeutic intervention for balance problems in the elderly: a single-blind, randomized, controlled multicentre trial. *Clinical Rehabilitation*, 15 (6), 624–36.

Wolf, S. L., Barnhart, H. X., Ellison, G. L., & Coogler, C. E. (1997). The effect of Tai Chi Quan and computerized balance training on postural stability in older subjects. *Physical Therapy*, 77 (4), 371-381.

Woodhull-McNeal, A. P. (1986). Activity in torso muscles during relaxed standing. *European Journal of Applied Physiology and Occupational Physiology*, 55(4), 419-424.

Woollacott, M.H., Schumway-Cook, A., Nashner, L.M., (1986). Aging and posture control: changes in sensory organisation and muscular coordination. *International Journal of Aging and Human Development*, 23, 97–114.

Woollacott, M. H. (2000). Editorial Systems Contributing to Balance Disorders in Older Adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 55 (8), M424-M428.

Yaggie, J.A., McGregor, S.J. (2002). Effects of isokinetic ankle fatigue on the maintenance of balance and postural limits. *Archives of Physical Medicine and Rehabilitation*, 83, 224–228.

Zhang, J. G., Ishikawa-Takata, K., Yamazaki, H., Morita, T., & Ohta, T. (2006). The effects of Tai Chi Chuan on physiological function and fear of falling in the less robust elderly: an intervention study for preventing falls. *Archives of Gerontology and Geriatrics*, 42 (2), 107-116.

Zhang, L. Q., & Rymer, W. Z. (2001). Reflex and intrinsic changes induced by fatigue of human elbow extensor muscles. *Journal of Neurophysiology*, 86 (3), 1086-1094.

Zok M, Mazzà C, Cappozzo A. (2008). Should the instructions issued to the subject in traditional static posturography be standardised? *Medical Engineering and Physics*, 30, 913–6.

Zwierska, I., Walker, R. D., Choksy, S. A., Male, J. S., Pockley, A. G., & Saxton, J. M. (2007). Relative tolerance to upper and lower limb aerobic exercise in patients with peripheral arterial disease. *European Journal of Vascular and Endovascular Surgery*, 31, 157-163.