Age-related differences in muscle morphology, strength and biomechanical function of the lumbar spine in healthy younger versus older men

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A thesis submitted in partial fulfilment of the University's requirements for the Degree of Doctor of Philosophy

Certificate of Ethical Approval

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Alexander Dallaway

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Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in an Elderly Community-dwelling Population: a biomechanical approach into the functional consequences of ageing lumbar paravertebral muscle morphology

This is to certify that the above named applicant has completed the Coventry University Ethical Approval process and their project has been confirmed and approved as Medium Risk

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Contents

List of Tables

List of Figures

List of Appendices

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Declaration and Summary of Publications

The PhD candidate (AD) declares that the work included in this thesis is his own work and has not been submitted for a previous degree. The research was conducted independently by AD. Parts of this thesis have been used in conference presentations and accepted for publication as follows:

Journal Articles

Dallaway, A., Kite, C., Griffen, C., Duncan, M., Tallis, J., Renshaw, D., and Hattersley, J. (2020) 'Age-Related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-Level Meta-Regression'. *Experimental Gerontology* 133, 110856

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Conference Poster Presentations

Dallaway, A., Duncan, M., Wilson, A., Wayte, S., Hutchinson, C., Weedall, A., Jones, T., Koning, B., Cooke, S., and Hattersley, J. (2018) 'Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in Older Adults'. *in UHCW 4th Annual Research Summit 2018*. held 2018 at Coventry, UK

Conference Abstracts

Weedall, A., **Dallaway, A**., Hattersley, J., Diokno, M., Hutchinson, C., Wilson, A., and Wayte, S. (2021) 'Is it possible to show differences in spinal muscles between younger and older healthy adult male volunteers using DTI?'. *in ISMRM & SMRT Annual Meeting & Exhibition 2021*. held 15-20 May 2021 at Vancouver, BC, Canada

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Thesis Abstract

Background Senescence of the musculoskeletal system is characterised by declines in muscle strength, muscle mass and physical function. These changes that accompany the ageing process were conceptualised into the condition known as sarcopenia, which is associated with adverse health outcomes and affects over 70% of older adults. Sarcopenia has been recently recognised as a geriatric syndrome; migrating from its original gerontological concept. As the proportion of older adults continues to rise across the world, sarcopenia presents an ever-increasing major health concern and socioeconomic burden. The research to date predominantly focuses on the appendicular muscles, despite the recognised importance of the lumbar musculature in maintaining physical function and independence in older age. Indeed, some researchers have suggested that the lumbar paravertebral muscles (LPMs) may be more susceptible to the effects of age-related sarcopenia than the appendicular muscles, although mechanisms for this phenomenon are ambiguous. With renewed interest, researchers are increasing efforts to understand sarcopenia of the spine which has led to the emerging concept of "spinal sarcopenia". Given the lack of research on this topic, it is of prime importance to investigate the effects of ageing on muscle morphology, strength and biomechanical function in the lumbar spine. Furthermore, it is important to understand the normal progression of age-related changes in this region to allow identification of sarcopenic and pathological deviations. Therefore, research should initially target healthy adults as undetermined phenotypes are likely hidden in the demographics of general populations. Extending our understanding of age-related changes in the LPMs will also provide guidance for effective clinical and public health intervention strategies to offset adverse health outcomes related to spinal sarcopenia in older adult populations.

Aim The main aim of the thesis was to explore age-related differences in lumbar spine specific measures of sarcopenia (i.e. muscle morphology, strength and biomechanical function) in healthy younger and older men. The secondary aim was to evaluate the interrelationships between muscle morphology, strength and biomechanical function in the lumbar spine alongside differences as a result of age.

Methods A range of methods was used due to the multidisciplinary nature of the research. Initially, a systematic review with meta-analysis was conducted to establish the relationship between ageing and degeneration of the lumbar musculature. Findings from the meta-regression were also used to inform methodological decisions regarding investigation of LPM morphology. Subsequently, quantitative MRI was performed to evaluate age-related volumetric and compositional differences in the LPMs. This

study was the first of three prospective observational studies to obtain primary data. Twelve healthy older men (67.3 ± 6.0 years) and 12 healthy younger men (24.7 ± 3.1 years) were included. Participants in the young group (YG) were matched to participants in the older group (OG) based on sex, ethnicity and physical activity (PA) level. To obtain strength data specific to the lumbar spine musculature, isokinetic dynamometry was applied to the trunk in the second experimental study. The OG and YG completed a protocol which included concentric and eccentric contractile modes as well as a wide range of angular velocities ($15°\cdot s^{-1}$ to $180°\cdot s^{-1}$). Finally, age-related differences in biomechanical function of the trunk during walking gait was explored using 3-D motion analysis. Statistical parametric mapping was used as a novel approach to determine phase-specific differences in kinematic and kinetic waveforms between the young and older age groups. As muscle measures are sensitive to lifestyle factors and health status, confounding variables such as PA level and physical disability were measured and controlled for where appropriate.

Results From the 34 studies (n = 6047) included in the meta-analysis, ageing was associated with atrophy (r = -0.255) and fat infiltration (r = 0.394) in the lumbar musculature. These degenerative changes also showed muscle, lumbar level and sex-specific responses. It was recommended that studies should use high-resolution imaging modalities to measure muscle volume across levels and across all muscles in the lumbar spine. Subsequently, the T2-weighted axial MRI images of the lumbar spine showed that increased fat infiltration was a global change across the lumbar musculature in older age. However, atrophy was muscle specific with age explaining 42% and 18% of the variance in quadratus lumborum and erector spinae muscle atrophy, respectively. Interestingly, PA level did not moderate the effect of age on muscle morphology degeneration (i.e. atrophy and fat infiltration). Concentric strength of the back muscles declined with age, which was more pronounced at greater movement speeds. However, loss of muscle volume and increase in fat infiltration was not able to explain age-related concentric strength loss in the trunk extensor muscles. It was likely that neuropathic processes with ageing were the cause of reduced concentric extensor strength in the OG. There was also an apparent preservation of eccentric trunk strength in older age, which was negatively associated with quadratus lumborum fat infiltration but not age. Regarding biomechanical function of the trunk during walking, the OG demonstrated reduced movement amplitudes in all planes of motion. However, reduced trunk movements in the coronal plane were likely a reflection of decreased range of pelvic obliquity motion. Few differences existed in trunk kinetics between the YG and OG, although the YG performed significantly more negative work in the coronal plane during the gait cycle (GC). This was likely due to greater lateral flexion excursions. Walking was on average 20% more functionally demanding on the trunk muscles in the OG compared to the YG, although this difference was not

statistically significant. There was also evidence of interplanar uncoupling of trunk motion in older age, which may increase the energetic demand of walking. Loss of muscle volume and increase in fat infiltration was unable to explain age-related differences in biomechanical trunk function.

Conclusion This thesis represents the first research to investigate lumbar spine specific measures of age-related sarcopenia. The dataset will provide a useful step in establishing normal features of muscle degeneration, strength loss and functional decline in the lumbar spine with ageing. This thesis will also help to establish the concept of spinal sarcopenia, which is an emerging field of interest in healthy ageing and musculoskeletal research. Furthermore, the findings within this thesis can be used in future research to design more effective targeted interventions aiming to improve physical function and health outcomes in older adults.

Highlights

- The first study to investigate age-related differences in muscle morphology, strength and biomechanical function of the lumbar spine
- The lumbar musculature degenerates with ageing
- High-resolution imaging modalities should be used to evaluate age-related morphological degeneration in the lumbar paravertebral muscles
- Age-related fat infiltration is a global change in the lumbar musculature whilst atrophy is muscle-specific
- Concentric strength of the trunk extensor muscles declines in older men (> 60 years) and the effect is greater with faster movements
- Eccentric trunk strength is somewhat preserved in older men
- In older age, trunk kinematics during gait are altered predominantly in the coronal and transverse planes
- Walking is more biomechanically demanding on the trunk musculature in older adults
- Decrements in lumbar muscle morphology do not appear to influence trunk strength loss and biomechanical function in older age
- The work presented in this thesis represents an extension to academic understanding of spinal sarcopenia

List of Abbreviations

3-D – Three-dimensional
ADL – Activity of daily living
ALM – Appendicular lean mass
ANOVA – Analysis of Variance
ANCOVA – Analysis of Covariance
BIA – Bioelectrical impedance analysis
BMI – Body mass index
BOS – Base of support
COG – Centre of gravity
COM – Centre of mass
CT – Computed tomography
ES – Erector spinae
FD – Functional demand
FOV – Field of view
GC – Gait cycle
GRF – Ground reaction force
IC – Initial contact
IPAQ – International Physical Activity Questionnaire
ISw – Initial swing
LBP – Low back pain
LLA – Lumbar lordotic angle
LPM – Lumbar paravertebral muscle
LR – Loading response
MANCOVA – Multivariate Analysis of Covariance
MF – Multifidus
MFI – Muscle-fat-infiltrate
MPA – Moderate physical activity
MRI – Magnetic resonance imaging
MS – Midstance
MVPA – Moderate to vigorous physical activity
NJM – Net joint moment
NMV – Normalised muscle volume
ODQ – Oswestry Low Back Pain Disability Questionnaire
ODQ-m – Modified Oswestry Low Back Pain Disability Questionnaire
OG – Older group
PA – Physical activity
PiG – Plug-in Gait
PS – Psoas
PSw – Pre-swing
QL – Quadratus lumborum
ROI – Region of interest
ROM – Range of motion
SPM – Statistical parametric mapping
TMC – Trunk Modular Component
TO – Toe off
TS – Terminal stance
TSw – Terminal swing
US – Ultrasound
VPA – Vigorous physical activity
YG – Young group

Chapter 1 Rationale and Narrative Review of the Literature

1.1 Introduction

The ancient Greeks were the first to document the loss of "flesh" and "vigour" as features of ageing; viewing ageing as an incurable, chronic and progressive disease (Narici and Maffulli, 2010). Galen of Pergamon (Green, 2012) proposed new ideas of ageing, viewing it as a condition midway between illness and health rather than a progressive disease. Although senescence of the musculoskeletal system continued to be explored over the centuries, Rosenberg (1989) was the first to assign a specific term to the age-related loss of muscle mass. Sarcopenia, derived from the Greek "sarx" meaning flesh and "penia" meaning a lack of, was the term proposed to describe the loss of lean mass with ageing independent of disease. Later, studies showed that muscle mass decreases at a rate of 1 – 2 % per year after the 6th decade of life (Doherty, 2003; Abellan Van Kan, 2009). The definition of sarcopenia has been refined over 30 years, encompassing elements of strength and function loss in addition to muscle mass loss (Cruz-Jentoft et al., 2019). Sarcopenia is a powerful risk factor for frailty and incidence risk of falls and fractures, loss of independence, weakness, functional impairment, physical disability and highly predictive of mortality in older populations (Roubenoff, 2000; Rantanen et al., 1999; Janssen, Heymsfield and Ross, 2002; Cawthon et al., 2014; Rizzoli et al., 2013). Sarcopenia has also been reported to increase the risk of concomitant diseases (Lu et al., 2013; Papalia et al., 2014) and ultimately lead to a poorer quality of life (Rizzoli et al., 2013). Age-related loss of muscle mass, strength and physical function is a major health concern and represents a substantial socioeconomic burden (Pinedo-Villanueva et al., 2019; Janssen et al., 2004). As the prevalence of sarcopenia, affecting up to 73% of older individuals (Sobestiansky, Michaelsson and Cederholm, 2019), and its impact has continued to increase due to ageing populations across the world (Department of Economic and Social Affairs Population Division, 2019), there has been a surge of interest in the condition (**Figure 1.1**).

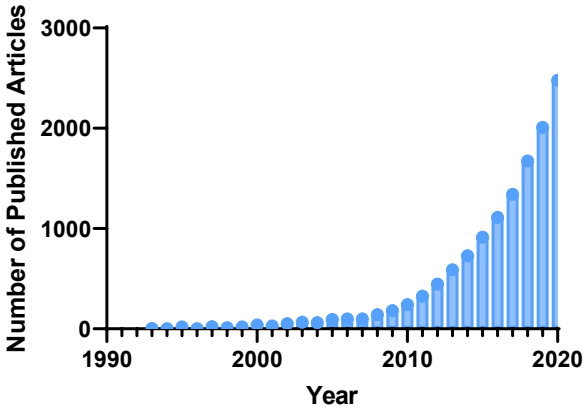


Figure 1.1 *Increasing number of publications on "sarcopenia" from inception to Jan 2021 using PubMed with the search term "sarcopenia"*

1.2 Sarcopenia Research in the Trunk

Sarcopenia migrated from its original gerontological concept to a clinical condition (Cruz-Jentoft et al., 2010b); recently being recognised as a geriatric syndrome (Gupta, 2019). This shift resulted in research groups and clinicians focusing more on the functional consequences of muscular deficits in older age and the impact of extrinsic factors such as nutrition, lifestyle and disease (Malafarina et al., 2012). For the purposes of this thesis, older age is defined as adults aged 60 years and above. During the seventh decade of life, the rate of muscle mass and strength loss increases marking 60 years of age as an important point when the musculoskeletal system undergoes involutional changes (von Haehling, Morley and Anker, 2010). Most of the research on sarcopenia focuses on appendicular muscle mass and basic measures of muscle strength and function. Indeed, the main working groups' diagnostic criteria include handgrip strength, gait speed (as a proxy for muscle function) and appendicular muscle mass index as primary assessments (Fielding et al., 2011; Studenski et al., 2014; Muscaritoli et al., 2010; Morley et al., 2011; Chen, L. K. et al., 2014; Cruz-Jentoft et al., 2019). Whilst assessment of the limbs is valuable, the importance of the spinal musculature in maintaining health and physical function in older age is being increasingly recognised. However, the concept of spinal sarcopenia is yet to be established.

According to Narici and Maffulli (2010), the postural muscles in the spine may be more susceptible to the effects of sarcopenia than the appendicular muscles. Knowledge on this topic is still limited by relatively few studies (approximately 1 % of published articles on sarcopenia concern the spinal musculature using PubMed searches from inception to Jan 2021) and a range of measurement techniques that precludes comparisons between them. Furthermore, research on sarcopenia of the trunk has tended to focus on populations with spinal diseases (Eguchi et al., 2017; Urrutia et al., 2018b), which makes it difficult to separate the independent effects of ageing from associated degenerative processes in the lumbar spine (Urrutia et al., 2018b). Analogous to the diagnostic criteria for sarcopenia, understanding the effects of age-related sarcopenia in the trunk is imperative in preventing immobility and maintaining physical independence in older age. Due to the paucity of studies investigating the lumbar musculature in healthy older adults, further research is warranted to understand the normal progression of age-related muscle degeneration, strength loss and functional decline in the trunk. Furthermore, the basic clinical assessments currently used to diagnose sarcopenia may be insufficient to detect subtle changes in muscle morphology or declines in physical function with ageing in the trunk. Therefore, more sophisticated measurements and analyses in the trunk may provide new insights into degenerative features of age-related sarcopenia.

1.3 Anatomy and Functional Roles of the Paravertebral Muscles in the Lumbar Spine

The trunk muscles play an important role in performing activities of daily living (ADLs), particularly the LPMs (**Figure 1.2**) which provide stability to the trunk (Crisco and Panjabi, 1991; Panjabi, 1992) and have a key function in the performance of lower limb tasks (Hicks et al., 2005a). Therefore, degeneration of the LPMs may be particularly detrimental to physical function in older age (Kita et al., 2013; Hicks et al., 2005b; Williams et al., 2017) given that the ability of the lumbar spine to withstand high loads is almost entirely attributable to the dynamic stabilising capacity of the trunk musculature (Cholewicki and McGill, 1996; Mcgill et al., 2003). Indeed, a cadaveric osteoligamentous spine will buckle under a compressive load of approximately 90 N (Crisco et al., 1992). Further assessment of strength and composition of the spinal musculature as well as physical function in older adults has been recommend due to substantial heterogeneity amongst studies (Hicks et al., 2005a; Granacher et al., 2013). To better understand the effect of ageing on LPM degeneration and loss of strength and physical function in the trunk, the intrinsic roles of the LPMs must first be established.

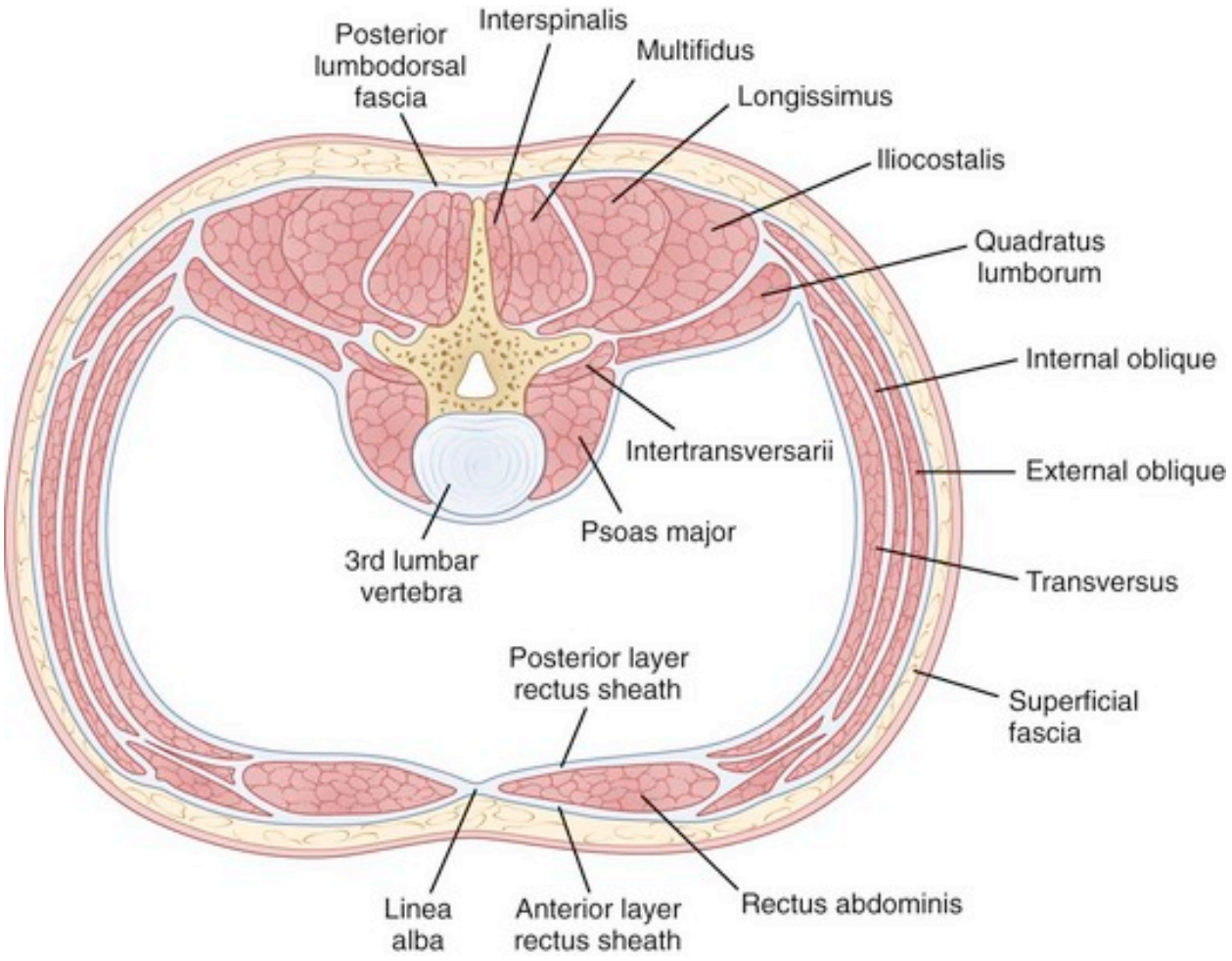


Figure 1.2 *Cross section of the musculature and fascia at the third lumbar level. The psoas, erector spinae (longissimus and iliocostalis), multifidus and quadratus lumborum muscles make up the lumbar paravertebral musculature. Image taken from Finneson (1980)*

1.3.1 Intrinsic Muscles of the Lumbar Spine

The muscles surrounding the lumbar spine are key to achieving spinal stability; withstanding forces encountered in everyday life (Barr, Griggs and Cadby, 2005), assisting in controlling movement and providing mechanical stability (Meakin et al., 2013). These muscles comprise of two main groups, the intrinsic muscles that are characterised by their intervertebral connections (Choi and Kim, 2012) and the extrinsic muscles that attach vertebrae to the limbs (Vasavada et al., 2011). The LPMs that are located deeply are also responsible for small movements that stabilise the spine (Ward et al., 2009; Cornwall, Stringer and Duxson, 2011; Hansen et al., 2006). Indeed, the actions of the MF account for more than two thirds of the stiffness of the spine (Wilke et al., 1995). Its short fibres and relatively large cross-sectional area (CSA) make it well suited to controlling intersegmental motion (Moseley, Hodges and Gandevia, 2002). The ability of the MF to produce large forces over a small operating range (Rosatelli, Ravichandiran and Agur, 2008) also demonstrates that it is biomechanically designed for stabilisation rather than movement (Ward et al., 2009). The MF itself is comprised of two types of fibre, deep and superficial (**Figure 1.3**). The deep fibres span two vertebral levels and function tonically, whereas the superficial fibres span three to five vertebrae and activate phasically (Macintosh and Bogduk, 1986; MacDonald, Lorimer Moseley and Hodges, 2006). Similar to the gross arrangement of the LPMs, the anatomical arrangement of the lumbar MF makes the deeper fibres more suited to stabilisation.

Figure 1.3 *Anatomy of the multifidus muscle. Image adapted from Jones (2020)*

The erector spinae (ES) in the lumbar spine is composed of the iliocostalis and longissimus muscles (**Figure 1.4**). It is a large muscle group that lies deep to the lumbodorsal fascia and attaches to an aponeurosis on the sacrum, iliac crest, and thoracolumbar spinous processes (Bogduk, 1980; Daggfeldt, Huang and Thorstensson, 2000). Its main function is extension of the spine (Danneels et al., 2000), although others have suggested that it primarily restricts excessive trunk movements during walking (Thorstensson et al., 1982). Since the ES has a large moment arm to bring about extension and lateral flexion of the spine (Lin et al., 2001) it is likely that it plays an important role in controlling sagittal and coronal plane trunk movements during walking (Masaki et al., 2016). Although the MF and ES have distinct primary roles (stabiliser and prime mover, respectively), their coordination elicits stability and optimum functioning of the trunk during dynamic tasks (Hicks et al., 2005a).

Figure 1.4 *Anatomy of the erector spinae. Image taken from Jones (2020)*

1.3.2 Extrinsic Muscles of the Lumbar Spine

The superficial muscles of the lumbar spine (**Figure 1.5**) generate torque for spinal motion and dissipate external forces acting on the spine (Bergmark, 1989). These muscles have also been shown to contribute to lumbar stability during load bearing movements (Kuukkanen and Mälkiä, 2000). The psoas (PS) and quadratus lumborum (QL) are frequently included in analyses of the lumbar

musculature, likely due to their relative size and the importance of their roles. The QL increases lumbar stiffness through its attachment to the transverse processes of the lumbar spine via the thoracolumbar fascia (Ebenbichler et al., 2001), assisting in lateral stabilisation of the spine (McGill, 2001). Electromyographical studies support this, suggesting that the dominant role of the QL is lumbar stabilisation (McGill, Juker and Kropf, 1996). However, others have suggested that the QL has a relatively modest action on the lumbar spine and its actual role in spinal biomechanics is yet to be determined (Phillips, Mercer and Bogduk, 2008). The PS has the largest CSA of any muscle in the lower levels of the lumbar spine (McGill, Patt and Norman, 1988). Its morphology, uniform fascicles lengths (Bogduk, Pearcy and Hadfield, 1992), together with electromyographical evidence suggests that the primary function of the PS is flexion at the hip (Bogduk, Pearcy and Hadfield, 1992; Juker et al., 1998). However, the PS also has the potential to flex the lumbar spine laterally and increase spinal stability through generating compressive forces (Santaguida and McGill, 1995).

Figure 1.5 *Anatomy of the superficial lumbar muscles. Image taken from Jones (2020)*

1.4 Literature Review of Studies

It has been recognised that more research is needed to fully understand the effect of age on muscle morphology, strength and physical function in the lumbar spine (Crawford et al., 2016c; Kalichman, Carmeli and Been, 2017; Granacher et al., 2013). To the author's knowledge, only one study has investigated age-related changes in LPM morphology, trunk strength and physical function together

in healthy adults (Shahtahmassebi et al., 2017). Other studies have included muscle morphology with either trunk strength or physical function measures (Williams et al., 2017; Hicks et al., 2005a; Schlaeger et al., 2019; Ikezoe et al., 2015; Lee et al., 2012) and not always in healthy populations (Williams et al., 2017; Lee et al., 2012). Discounting any of the three features of age-related sarcopenia could result in an incomplete understanding of the impact ageing has on the lumbar spine. Indeed, age-related strength loss can occur independent of morphological changes in skeletal muscle (Ochala et al., 2007) and decrements in strength may not necessarily result in functional declines in older age (Roubenoff, 2003). An approach that considers the interdependency between these with the age-effect is needed.

Only two studies (Lee et al., 2012; Schlaeger et al., 2019) have investigated the relationships between ageing, muscle morphology of the lumbar spine and trunk strength. Despite highly disparate methods, age was poorly associated with morphological degeneration and strength measures in both studies. However, reduced muscle CSA was related to declining trunk strength. Schlaeger and colleagues (2019) suggested that fat infiltration within the ES and PS improved the prediction of paraspinal muscle strength more than muscle size alone, whilst Lee et al. (2012) observed no associations between paraspinal and PS fat composition and trunk strength. This disparity is likely the result of sampling and methodological variance. Lee et al. (2012) included patients with low back pain (LBP) whilst Schlaeger et al's (2019) study was composed of healthy men and women. Furthermore, fat infiltration was qualitatively graded by Lee et al. (2012) compared to proton density fat fraction being quantitatively derived by Schlaeger et al. (2019). Strength measures also differed between the studies. Lee and colleagues (2012) used isokinetic dynamometry to measure peak trunk flexor and extensor concentric strength at $60°\cdot s^{-1}$ whilst the other study (Schlaeger et al., 2019) measured trunk strength isometrically, which limits the information that can be obtained about the trunk's dynamic force producing capacity through a range of motion (ROM). These studies also discounted physical function measures which may have revealed important relationships with age, trunk strength and muscle morphology. The considerable heterogeneity between these studies precludes conclusions from being drawn, warranting further research on this topic.

Other studies have focused on the relationships between ageing, muscle degeneration and physical function decline (Williams et al., 2017; Hicks et al., 2005a; Ikezoe et al., 2015). However, the same issues complicate our understanding as these studies use a variety of tests and in different populations. These studies were also limited by not measuring trunk muscle strength. Indeed, physical function tests are more varied than strength assessments and often provide composite measures which makes it difficult to isolate where decline in physical function is most apparent. Similar to the other studies (Lee et al., 2012; Schlaeger et al., 2019), Williams and colleagues (2017) found that age

had little effect on degeneration of the trunk musculature. Lumbar spine muscle mass was also poorly associated with physical function, although intramuscular fat infiltration was able to identify physical function impairments amongst older adult cancer patients (Williams et al., 2017). These findings were consistent with Hicks et al. (2005a), who reported that trunk muscle composition, and not trunk muscle area, is associated with reduced functional capacity in a large cohort of well-functioning older adults. However, older institutionalised women have shown a reduction in walking ability with muscle atrophy of the ES, MF and PS muscles (Ikezoe et al., 2015).

Whilst there appears to be a consensus that ageing muscle morphology in the lumbar spine is related to declining physical function, measurement of physical function varied greatly. Amongst these studies (Williams et al., 2017; Hicks et al., 2005a; Ikezoe et al., 2015) physical function tests included clinical assessments (e.g. Timed Up-and-Go test), performance batteries (e.g. The Health ABC Physical Performance Battery), self-reported physical health and falls and maximum walking speed. The efficacy of these assessments regarding trunk function is questionable. Whilst the performance of functional movements relies on the mechanical function of the spine (Cholewicki and McGill, 1996) and the engagement of its stabilising muscles (Hicks et al., 2005a), it is difficult to distinguish the effect of regional changes in muscle composition on functional outcomes (i.e. does degeneration of the spinal musculature affect physical function?). Concerns have also arisen over the presence of a 'ceiling effect' in such assessments (Frost et al., 2005; Sayers et al., 2006). It has been suggested that functional decline in active, independent individuals may not be detectable, which will give individuals a false sense of good health (Puthoff, 2008). These performance tests may also lack the sensitivity to identify sarcopenia or LBP as risk factors for functional decline (Eggermont et al., 2014; Cawthon, 2015). Previous studies have focused on the peripheral musculature, however, given that the LPMs play an important role in measured functional changes (Cholewicki and McGill, 1996; Hicks et al., 2005a) and stabilise the trunk during everyday activities (Panjabi, 1992; Crisco and Panjabi, 1991), it is pertinent to explore physical function in relation to the trunk using sophisticated methods to understand the age-effect specific to this region. Biomechanical analysis is able to derive kinematic and kinetic data of the trunk and could extend our knowledge of physical function in older adults.

Shahtahmassebi et al. (2017) investigated the associations between trunk muscle morphology, strength and function with ageing in a healthy population. Advancing age was significantly and negatively correlated with functional outcomes but not with trunk flexion/extension strength nor MF atrophy. Shahtahmassebi et al. (2017) also found that trunk muscle strength was more consistently associated with functional performance than trunk muscle morphology after accounting for age. This may be explained by the disproportionately faster rate of strength loss compared to muscle atrophy with ageing (Delmonico et al., 2009). However, these findings must be taken with caution as there

were potential limitations associated with each of the primary data collection methods. Firstly, ultrasound was used to image the MF muscle, which has known limitations due to its low resolution and inability to differentiate between muscle and fat tissue (Hides, Richardson and Jull, 1995; Pressler et al., 2006; Wallwork et al., 2009). Strength was assessed using an isokinetic dynamometer which is considered the gold-standard approach (Dvir and Müller, 2019; Dvir, 2004), although the mode of contraction was isometric. Isometric assessment provides useful information about maximal strength producing capacity of the trunk (Roth et al., 2017; De Blaiser et al., 2018). However, torque measured at discrete joint angles may not accurately reflect dynamic muscle function (Rousanoglou and Boudolos, 2008). During gait, the trunk musculature is constantly changing between states of concentric and eccentric activation (White and McNair, 2002). Therefore, it is likely that Shahtahmassebi et al's (2017) findings regarding trunk strength and physical function do not reflect real-world scenarios. Finally, physical function was examined using conventional performance batteries and clinical assessments such as the Six Minute Walk Test, the 30-second Chair Stand Test and the Sitting and Rising Test. These tests do not isolate the effects of ageing on the spine, rather they evaluate whole-body physical function. Therefore, the observed associations between trunk muscle strength and physical function are not wholly reflective of the ageing process in the spine. Given that this study represents the entire body of literature investigating age-related decrements in the lumbar musculature, trunk strength and physical function and their interdependencies, further research should be conducted using more sophisticated methods and outcomes specific to the lumbar spine.

1.5 Current Research and Advancements

Age-related sarcopenia of the LPMs has received renewed interest, even stimulating the conceptualisation of spinal sarcopenia (Kuo et al., 2020; Kim et al., 2019). Initially this started with researchers using non-invasive imaging to characterise age-related muscle degeneration in the lumbar spine (Bukvić et al., 2019; Englesbe et al., 2010; Golse et al., 2017). There are now efforts to understand how ageing affects LPM morphology, trunk strength and physical function (Kim et al., 2019; Shahtahmassebi et al., 2017) using the same conceptual framework as sarcopenia (Cruz-Jentoft et al., 2019). However, conventional sarcopenia indices are not sensitive enough or appropriate to explore age-related decline in muscle morphology, strength and function in the lumbar spine. Therefore, it is necessary to investigate age-related sarcopenia in the lumbar spine using measures and indices that are specific to the region. Furthering understanding will provide numerous benefits including identification of spinal pathologies, application of evidence-based training protocols

targeting the lumbar musculature that are contractile mode and muscle specific, and elucidation on whether sarcopenia is systemic or site-specific in older adults.

1.6 Aims of the Thesis

The **primary** aim of the thesis was to explore age-related differences in lumbar spine specific measures of sarcopenia (i.e. muscle morphology, strength and physical function) comparing healthy younger and older men.

The **secondary** aim was to evaluate the interrelationships between muscle morphology, strength and function in the lumbar spine alongside differences as a result of age .

Each experimental chapter focused on a single component using specific measures to provide an in-depth investigation into age-related differences in the lumbar spine. Subsequent chapters incorporate measures from previous chapters to draw together interdependencies between muscle morphology, strength and function as well as moderating effects such as PA.

A thesis map is presented throughout the thesis, providing the “Problem Statements” and “Aims” at the start of each chapter and the “Key Findings” and “Implications” at the end of each chapter.

1.7 Thesis Structure

This thesis explores normal age-related differences in muscle morphology, strength and function in the lumbar spine. The diagnostic criteria for sarcopenia was used as a conceptual framework and applied to the lumbar spine. This approach is in line with other research groups aiming to characterise ‘spinal sarcopenia’, which is poorly understood at present. This also provided justification for studying age-related differences in a healthy population; normal changes in older adults must be established before sarcopenic and pathological deviations can be identified. Due to the multidisciplinary nature of the research and range of data collection methods used to assess muscle morphology, strength and function of the lumbar spine, this thesis is presented in a manuscript format comprised of self-contained chapters. A key decision in laying out the thesis in this manner was including literature reviews within their corresponding chapters. Whilst chapters were theoretically and conceptually related, the literature relevant to each chapter generally did not acknowledge the other facets of age-related sarcopenia (i.e. muscle morphology, muscle strength and physical function). Therefore, including literature reviews within chapters was an effort to focus the reader’s attention to specific topics whilst making the development of ideas clearer throughout the thesis. Chapters were inextricably linked and relationships between them were also explored, however, it was felt that the

diverse nature of the research areas were better represented as separate literature reviews within corresponding chapters.

A chapter was also dedicated to confounding factors as age-related changes in musculoskeletal outcomes are sensitive to PA level, functional disability status and whole-body composition. Overall, the thesis was structured to highlight individual components of the research whilst sequentially building upon previous chapters by considering moderating effects and interrelationships. Therefore, chapters five (muscle morphology), six (strength) and seven (physical function) reflect the conceptual framework of sarcopenia but specific to the lumbar spine. The information gained in each experimental chapter was fed forward into the following chapter as indicated in the flow chart below.

CHAPTER 1

Chapter 1
- The first chapter provides a rationale for the research and introduces key concepts that frame the thesis. A narrative review of the literature is also included, focusing on studies that have investigated age-related degeneration of the lumbar musculature, trunk strength and physical function.

Chapter 2
- Chapter 2 describes the study design and research methodology in the thesis. Population characteristics were presented in addition to the recruitment and matching procedures.

Chapter 3
- Potential confounding factors were explored in this chapter. Methods were outlined regarding physical activity measurement, functional disability assessment, whole body composition analysis and handgrip strength. Covariates were identified and retained for future analyses.

Chapter 4
- A systematic review with meta-analysis is presented in chapter 4. The purpose of this chapter was to establish the age-effect on muscle degeneration in the lumbar spine. In addition, methodological covariates were investigated to assess whether they moderated the relationship between ageing and muscle degeneration. The outcomes from the meta-analysis were used to develop the primary data collection methods, specifically in chapter 5.

Chapter 5
- Age-related atrophy and compositional changes of the four main muscles in the lumbar spine were investigated in this chapter. An MRI protocol was developed and image analysis techniques refined to quantify muscle volume and muscle fat infiltration. The influence of physical activity on muscle outcomes was also explored. The moderating effects of muscle atrophy and fat infiltration were explored in subsequent chapters.

Chapter 6
- The effect of age on trunk strength was explored. Dynamic trunk strength was measured using isokinetic dynamometry. A range of contractile testing modes was used to explore whether contraction type and movement speed affected trunk extensor and flexor strength with ageing. Relationships with muscle morphology and physical activity were also evaluated to ascertain whether strength was influenced by muscle degeneration and lower engagement with vigorous physical activity in old age.

Chapter 7
- In chapter 7, biomechanical differences in trunk function during walking gait were investigated between the younger and older participants. Spaciotemporal, kinematic and kinetic variables were calculated. In addition, functional demand of the trunk was calculated using strength data from chapter 6. This provided a measure of the biomechanical challenge posed by walking in the trunk. Rather than focusing solely on discrete data to investigate the effect of age, a novel statistical analysis technique was used to identify whether age-related differences were phase specific. Relationships between kinematic variables and muscle morphology were also explored to determine if loss of trunk function was due to degenerative muscle morphology.

Chapter 8
- The final chapter draws the findings together and makes general conclusions. The significant contribution of the research is highlighted and the wider implications are also considered. Practical applications and areas of future research are recommended to build on the work in this thesis.

Chapter 2 Methodology and General Methods

2.1 Research Design

The research design was cross-sectional, quantitative, and prospective, utilising comparative analyses between an OG and YG. The purpose of this methodological design was to investigate differences in LPM morphology, physical function and strength between age groups (old vs young) as well as analyse the interrelationships between them. Younger (18-30 years) and older (60-80 years) participants were chosen to represent a sufficient time period in which it would be reasonable to observe changes in skeletal muscle morphology, physical function and strength. Previous studies investigating age-related differences in muscle morphology and movement biomechanics have adopted a similar approach (Mian et al., 2007; Reeves et al., 2009; Ikezoe et al., 2012; Samuel et al., 2012; Crawford et al., 2016b). Muscle mass and strength decline after the age of 30 years then accelerate past the age of 60 years (Keller and Engelhardt, 2013; Hughes et al., 2002). Comparing age group extremes provides a useful reference point in which the onset of any observed responses between the ages of 30 and 60 years could be explored in the future.

2.2 Sample Size

The sample size for each group (n = 12) was determined based on power calculations and on the balance of additional data collection efforts and the ability to detect significant effects within the boundaries of normal distribution. According to Desmond and Glover (2002), a group of 12 participants is sufficient to ensure 80% power at $\alpha = 0.05$ at the single voxel level in the use of MRI. From a statistical perspective, an increase in sample size past 12 participants may not offer any additional benefit in increasing statistical significance for a medium to large constant effect size or reducing the 95% confidence interval limit (Birkett and Day, 1994). Power calculations were also performed in G* Power (Version 3.1.9.2) based on the means and standard deviations of previous studies. Firstly, studies comparing LPM volume and fat infiltration in young and older adults were sought (Valentin, Licka and Elliott, 2015; Crawford et al., 2016a; Meakin et al., 2013). According to the power calculation estimate, a sample size of seven to ten participants per group was sufficient for the MRI study based on large observed effect sizes, equal allocation rates, an α rate of 0.05 and β rate of 0.2. It should be noted that not all observed age effects were large; differences in some outcome measures between young and OGs in the aforementioned studies were small (Cohen's $d < 0.2$) which yielded much larger sample size estimates. For the trunk strength study, a priori sample size estimation was performed using means and standard deviations from a previous study (Danneskiold-

Samsøe et al., 2009) assessing isokinetic trunk torque in a similar population of healthy younger and older men. A sample size of eight to 12 participants per group was calculated for large observed effect sizes ($\alpha = 0.05$, $\beta = 0.2$). Resource limitations were also considered in determining an appropriate sample size. Therefore, a total sample size of n = 24 represented the maximum number of participants that could be feasibly included in the study whilst providing sufficient power based on calculations from previous studies.

2.3 Participants

2.3.1 Recruitment

Healthy volunteers were recruited between November 2018 and June 2019. There were two reasons for recruiting healthy participants. Firstly, much of the research focuses on diseased populations which has confounded our understanding of normal changes with ageing in the lumbar spine musculature. Therefore, there is a need to establish the effect of healthy ageing. Secondly, selecting participants based on health status is important as undetermined phenotypes are likely hidden in the demographics of general populations. Potential participants were recruited using the following strategy:

- Access a research database at Coventry University of older individuals living in Coventry and where participants have consented to data being made available for other projects.
- Approach participants who are already participating in the "Faster, higher, older: multidisciplinary research into the effects of age-related sarcopenia" project.
- Through targeted advertisement and poster placement (**Appendix a**) (e.g. social clubs, community groups, sports and leisure clubs, libraries, churches, newspapers/magazines, Coventry University)
- Through social media (e.g. Twitter)
- Public engagement (e.g. local communities, open days at UHCW, Coventry University)

Volunteers who were eligible to participate in the study were sent a letter of invitation with the Participant Information Sheet (**Appendix b**) and consent form (**Appendix c**). Following a short interval (minimum of two days), potential participants were contacted by email or phone so that any questions about participation they had could be answered. Prior to the first measurement, participants had another opportunity to have any questions they had about the study or any of the documents they completed addressed. All participants were made aware of the option to withdraw at any time from the study, without prejudice or consequence, and for any reason.

2.3.2 Eligibility

To be eligible to participate in the study and assigned to the OG an individual had to satisfy the following inclusion criteria:

Inclusion criteria (OG)

- Aged between 60 years and 80 years
- Body mass index (BMI) in the range of 18.5 $kg \cdot m^{-2}$ to 30 $kg \cdot m^{-2}$
- Generally healthy (free from disease, musculoskeletal injury or functional impairment, e.g. stroke)
- Able to provide informed consent and follow study procedures

To be eligible to participate in the study in the YG an individual had to satisfy the following inclusion criteria:

Inclusion criteria (YG)

- Aged between 18 years and 30 years
- BMI in the range of 18.5 $kg \cdot m^{-2}$ to 30 $kg \cdot m^{-2}$
- Generally healthy (free from disease, musculoskeletal injury or functional impairment, e.g. stroke)
- Able to provide informed consent and follow study procedures

Participants were excluded if any of the following conditions were met:

Exclusion criteria (Both groups)

- Unable to undergo MRI based on the MRI screening questionnaire
- Current smokers, or ex-smokers ceasing < 6 months ago (assessed on the Health and lifestyle questionnaire)
- Daily consumption of alcohol (assessed on the Health and lifestyle questionnaire)
- Existing or past medical history of vascular disease, cancer, diabetes, neurological disorders, kidney, pulmonary, digestive (Coeliac disease), thyroidal disease, osteoporosis or history of falls
- Neuromuscular disorders/injuries, unable to live independently or physical impairments that limit 'normal' physical function

To mitigate the impact of biological variation across the life course, women were excluded. Women undergo distinct biological events during their life, namely the menopause. During the menopause, there is a sharp decrease in hormonal status (e.g. oestrogen levels), bone remodelling and muscle mass and strength (Maltais, Desroches and Dionne, 2009). As men do not experience such an extreme biological event, rather experiencing gradual senescence with age, including both men and women in the analyses would introduce a source of experimental error. Furthermore, controlling hormonal status on musculoskeletal outcomes was outside the remit of this thesis meaning that including only women would also introduce experimental error. As a first step, understanding degeneration of the

lumbar musculature in healthy men provides a baseline that can be used as a comparator in future studies seeking to elucidate inter-sex differences.

To minimise temporal effects (i.e. time of day and seasonal variations), matched participants underwent testing concurrently rather than one of the age groups being tested before the other. PA levels are known to vary with seasonality (Tucker and Gilliland, 2007), therefore, efforts were made to ensure matched participants from the OG and YG were tested within the same time period. Furthermore, tests were conducted at approximately the same time of day for each participant which reduced the effect of daily biorhythms.

The flow chart in **Figure 2.1** illustrates the number of volunteers approached, the number of volunteers who met the inclusion criteria and the number of participants included in the study.

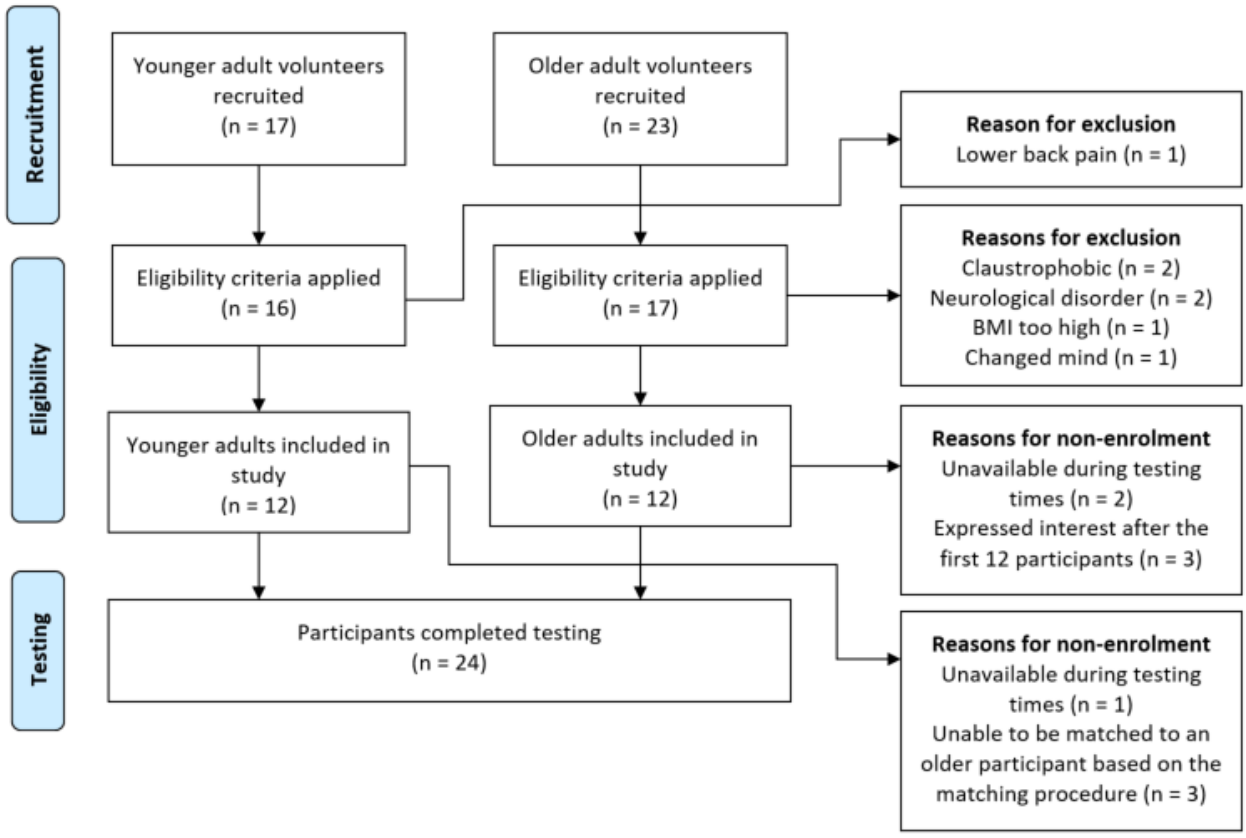


Figure 2.1 *Flow-diagram showing the recruitment and inclusion process, completion number and reasons for exclusion and non-enrolment*

2.3.3 Matching Procedure

In the literature, comparative studies investigating age-related differences in muscle morphology, physical function and strength have provided limited information regarding matching procedure for

older and younger adults. There are a number of potential variables that could moderate morphological, strength and function-based measures of skeletal muscle as a function of age. PA level, sex and ethnicity are well-known to moderate skeletal muscle mass and strength (Goodpaster et al., 2008, 2006; Delmonico et al., 2009). Therefore, for each participant stratified into the OG, a younger participant was matched based on these variables in a one-to-one participant matching procedure. Participant characteristics are presented in **Table 2.1**

2.3.3.1 Physical Activity Matching Procedure

Questionnaires are frequently used to obtain self-reported measures of PA. Although subjective reporting, recall and response bias, can affect the validity and reliability of such methods, they are practical, low cost and have a low participant burden (Shephard, 2003; Prince et al., 2008). The International Physical Activity Questionnaire (IPAQ) – long and short form have demonstrated good concurrent and construct validity for assessing PA levels when self-administered in healthy adults (Craig et al., 2003; Hagströmer, Oja and Sjöström, 2007). The brevity of the IPAQ-short form and its ability to provide an overall total PA estimate makes it particularly advantageous when seeking to establish PA levels in a population (Bauman et al., 2009). To reflect changes in PA patterns with ageing, a modified IPAQ was developed for older adult respondents known as the IPAQ-E. This modified version, suitable for over 65's, has shown a high level of concurrent validity with the IPAQ and is able to classify older adults into PA categories (Hurtig-Wennlf, Hagstrmer and Olsson, 2010). However, recent research has questioned the suitability of the IPAQ for assessing habitual physical behaviour in older adults (Ryan et al., 2018).

Participants in the YG completed the self-administered IPAQ – Short Form English (IPAQ) (**Appendix d**) whilst participants in the OG completed the IPAQ – Short Form English - elderly (IPAQ-E) (**Appendix e**). Responses were converted into a categorical score following the IPAQ scoring protocol (IPAQ Research Committee, 2005). The three categories for PA were: low, moderate and high. Younger participants were matched to an older participant based on their categorical IPAQ score. Eight older participants were categorised as highly active and four were categorised as moderately active. Eight highly active and four moderately active younger volunteers were therefore matched accordingly. Age group mean PA scores (met·min/week) were also compared (independent t-test, $\alpha < 0.05$) once all participants were matched to determine whether PA levels were statistically different between the OG and YG. The OG (4662 ± 2133 met·min/week) were more active than the YG (4235 ± 2868) but not significantly ($p > .05$).

Table 2.1 *Participant Characteristics*

	Younger group (n = 12)	Older group (n = 12)	T test
Demographics			
Age (years)	24.7 ± 3.1	67.3 ± 6.0	t (22) = -21.8, p < .001
Age range (years)	19 – 30	61 – 81	
Ethnicity (% white)	100	100	
Sex (% male)	100	100	
Anthropometrics and body composition			
Height (m)	1.78 ± 0.1	1.74 ± 0.1	t (22) = 1.2, p = .227
Mass (kg)	76.4 ± 11.2	79.2 ± 10.8	t (22) = -0.6, p = .550
BMI ($kg \cdot m^{-2}$)	24.1 ± 2.2	26.0 ± 2.7	t (22) = -1.9, p = .066
Whole-body lean mass (kg)	59.7 ± 6.9	57.4 ± 6.3	t (22) = 0.8, p = .410
Trunk lean mass (kg)	31.3 ± 3.9	33.3 ± 3.1	t (22) = -1.4, p = .180
Appendicular lean mass (kg)	28.3 ± 3.1	24.1 ± 3.3	t (22) = 3.3, p = .003
Whole-body fat mass (kg)	13.6 ± 4.9	19.2 ± 5.4	t (22) = -2.6, p = .016
Physical limitation/disability			
Modified Oswestry Low Back Pain Disability Questionnaire (%)	2.2 ± 2.3	2.2 ± 3.5	t (22) = 0.0, p = 1.000
Physical activity status			
IPAQ (low : moderate : high)	0 : 4 : 8	0 : 4 : 8	
IPAQ score (met·min/week)	4235 ± 2868	4762 ± 2133	t (22) = -0.5, p = .615
Moderate physical activity (MPA) (average hours per day)	3.9 ± 0.8	4.2 ± 1.1	t (22) = -0.7, p = .514
Moderate-to-vigorous physical activity (MVPA) (average hours per day)	6.6 ± 1.4	6.3 ± 1.5	t (22) = 0.5, p = .604
Vigorous physical activity (VPA) (average hours per day)	2.6 ± 0.6	2.1 ± 0.6	t (22) = 2.4, p = .027
Muscle function			
Dominant handgrip strength (kg)	45.0 ± 7.5	37.4 ± 9.1	t (22) = 2.2, p = .037
Non-dominant handgrip strength (kg)	42.8 ± 5.3	36.3 ± 7.9	t (22) = 2.4, p = .027

Note: bold text denotes a significant difference

2.4 Institutional Ethical Approval and GafREC

Ethical approval for this study was obtained from the Coventry University Ethics Committee (P70399) on 13th September 2018. University Hospitals Coventry and Warwickshire (UHCW) was used as a site for data collection. Governance arrangements for Research Ethics Committees (GafREC) documentation was completed and approved by UHCW NHS Trust so that data collection could take place on site (**Appendix f**). Written informed consent was obtained from all participants prior to entering the study. Upon enrolment, all participants completed the Health and Lifestyle Questionnaire (**Appendix g**). Prior to each testing session that required any PA, participants completed a Pre-test Health Screening Questionnaire. Participants also completed an MRI Safety Questionnaire (**Appendix**

h) and MRI Consent Form (**Appendix i**) prior to undergoing their MRI scan. This was checked and counter-signed by the Research Radiographer at UHCW.

2.5 Research Setting

Data collection was carried out across two different research facilities. Participants attended the Human Performance Laboratory at Coventry University on two separate occasions and UHCW on two separate occasions, within a four-week data collection period. Therefore, each participant's data were collected within a four-week period. This was to ensure data were obtained within as minimal time period as possible whilst allowing sufficient recovery between successive test sessions. An example data collection overview is illustrated in **Figure 2.2**.

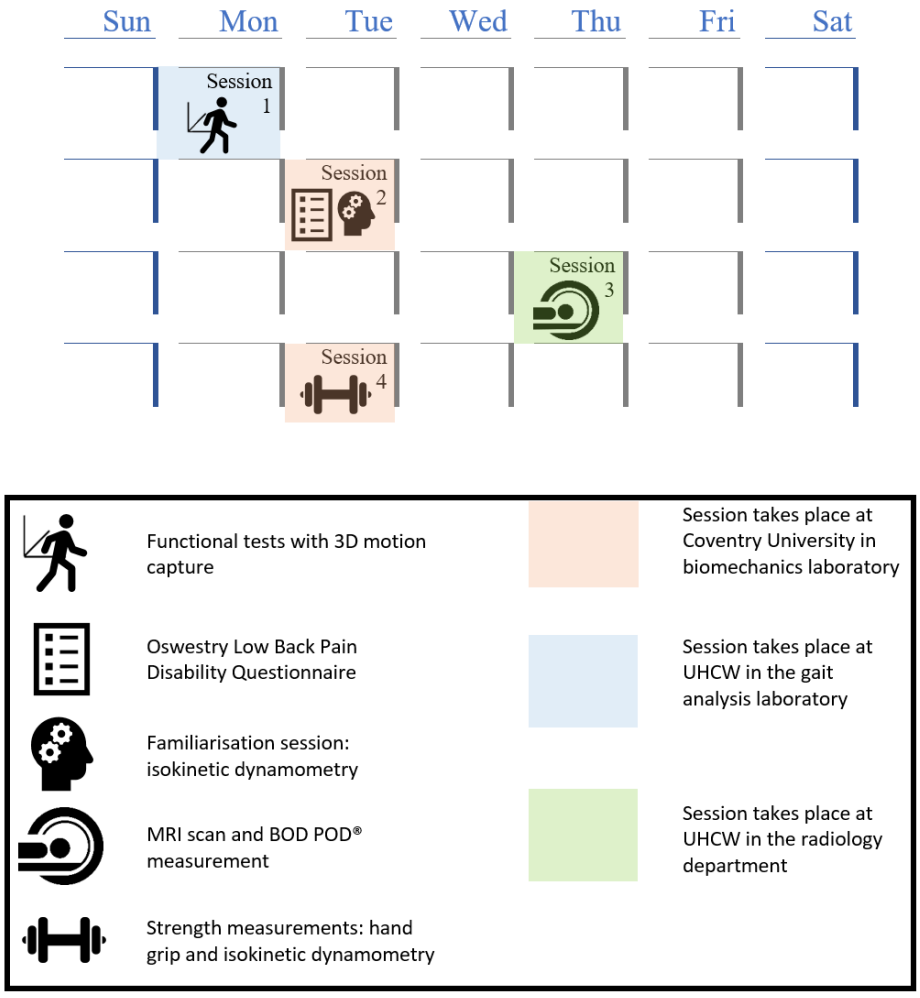


Figure 2.2 *Example of order of testing and familiarisation sessions*

Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function

Chapter Abstract

Background Age-related changes in skeletal muscle morphology, strength and biomechanical function are highly sensitive to the moderating effects of LBP, whole-body composition and PA. Handgrip strength is also associated with functional capacity in older men. It is important to control the effect of these covariates on muscle related outcomes to separate them from the effects of ageing.

Methods Whole-body composition, handgrip strength, LBP and PA level were measured in twelve healthy older (67.3 ± 6.0 years) and young (24.7 ± 3.1 years) men. Participants completed the Modified Oswestry Low Back Pain Disability Questionnaire (ODQ-m) and underwent bioelectrical impedance analysis. A handgrip dynamometer was used to measure handgrip strength and PA was analysed using wrist-worn accelerometers. Independent samples T-tests were used to compare differences between the YG and OG. Significance was set at 0.05 for all statistical tests. Cohen's d values were calculated to assess the size of the difference between age groups.

Results The YG had significantly more appendicular lean mass (ALM) ($p = .003$) and significantly less whole-body fat mass ($p = .016$) than the OG. Dominant and non-dominant handgrip strength were significantly greater in the YG than the OG ($p = .037$ and $p = .027$, respectively). The OG spent significantly less time in vigorous physical activity (VPA) intensities than the YG ($p = .027$). The ODQ-m scores were not significantly different between the YG and OG.

Conclusions The large and significant difference in VPA between the YG and OG indicates that it should be controlled for when determining age-related differences in musculoskeletal measures. Whole-body fat mass and ALM should also be considered potential covariates. Low ODQ-m scores suggested that there was no evidence of LBP causing functional disabilities, and the high level of similarity between groups indicated that LBP was not likely to be a confounding factor. Given the large difference in handgrip strength between the OG and YG, it is an interesting exploratory factor that may associate with musculoskeletal outcomes of the lumbar spine and should be explored.

Key words: physical activity; accelerometer; handgrip strength; whole-body composition; low back pain; ageing

Table 3.1 *Thesis Map*

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	
		Implications	
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression		Aim	
		Key findings	
		Implications	
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men		Aim	
		Key findings	
		Implications	
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	

3.1 Introduction

Muscle morphology of the lumbar spine, trunk strength and biomechanical function are highly sensitive to the moderating effects of LBP, whole-body composition and PA (Teichtahl et al., 2015b; Kalichman et al., 2010; Gabr and Eweda, 2019; Hicks et al., 2005b; Morie et al., 2010). Handgrip strength is also associated with functional capacity in older men (Desrosiers et al., 1995). Furthermore, the prevalence of LBP has been shown to increase with ageing (Dionne, Dunn and Croft, 2006), older adults exhibit increased adiposity (Ponti et al., 2020), handgrip strength generally declines in older age (Samuel et al., 2012) and PA levels reduce (Milanović et al., 2013). It is likely that there is a substantial amount of covariance between these variables and muscle related outcomes with advancing age. Therefore, it is important to assess and control the moderating effect of LBP, whole-body composition and PA on muscle related outcomes to separate them from the effects of ageing.

3.1.1 Low Back Pain and Physical Disability

The Oswestry Low Back Pain Disability Questionnaire (ODQ) is a valuable tool for evaluating an individual's functional status regarding LBP and has emerged as the most widely recommended condition specific outcome measure for spinal disorders (Fairbank and Pynsent, 2000). A recent modification by Fritz and Irrgang (2001) replaced the sex life section with a section concerning employment and home-making ability. Fritz and Irrgang (2001) made this change because the sex life item was frequently left blank or was not appropriate or applicable (Hicks and Manal, 2009; Mousavi et al., 2006; Fritz and Irrgang, 2001). The Modified ODQ (ODQ-m) has shown to have superior measurement properties compared to other back pain disability questionnaires and higher test-retest reliability over a 4-week period (Fritz and Irrgang, 2001).

3.1.2 Whole-Body Composition

Bioelectrical impedance analysis (BIA) offers a less expensive, less time consuming, safer and more easily performed method of determining body composition compared to the reference and criterion standard Dual-energy x-ray absorptiometry (DEXA) (Lukaski, 1987; Kyle, 2004; Verney et al., 2016). BIA has shown to be an acceptable and reproducible alternative amongst various populations, including healthy younger and older adults (Esco et al., 2015; Verney et al., 2015; Wang et al., 2013), although factors such as obesity (Verney et al., 2016) and ethnicity (Jakicic, Wing and Lang, 1998) may affect its validity and reliability. Efforts were made in this research to form two homogenous groups, varying only in age, to separate the age-effect from confounding factors such as ethnicity (see **2.3.3**). More recently, segmental BIA has been developed to attenuate discrepancies between resistance and trunk mass (Kyle, 2004; Foster and Lukaski, 1996). This technique is particularly useful in determining appendicular skeletal muscle mass (De Lorenzo and Andreoli, 2003).

3.1.3 Handgrip Strength

Handgrip strength is one of three criteria used in the diagnosis of sarcopenia (Cruz-Jentoft et al., 2019, 2010a; Fielding et al., 2011). Handgrip strength is a good predictor of physical performance, current health status (Desrosiers et al., 1995) and disability in older men (Giampaoli et al., 1999). However, others have suggested that handgrip strength may not be the most suitable measure to predict physical function (Samuel et al., 2012; Liu et al., 2016). Despite this, handgrip strength testing remains commonplace due to its simple and non-invasive application. In a large study (n = 6089) of 45-68 year

old men, handgrip strength was highly predictive of functional limitations and disability 25 years later (Rantanen et al., 1999), illustrating its efficacy in musculoskeletal ageing research. Handgrip strength therefore remains clinically relevant and is time and labour efficient as a measure of strength for use in older adult populations.

3.1.4 Physical Activity Status

PA contributes to healthy ageing (Nelson et al., 2007; Nilsson, Wåhlin-Larsson and Kadi, 2017), conversely sedentary behaviour has been associated with adverse health outcomes (Thorp et al., 2011). PA typically reduces with ageing (Milanović et al., 2013), which may increase age-related decrements in muscular force production (Bassey, 1998; Skelton et al., 1994; Hunter, Thompson and Adams, 2001; Ferreira et al., 2012; Goodpaster et al., 2008; Rolland et al., 2004), muscle morphology (Goodpaster et al., 2008; Fragala et al., 2014; Cartee et al., 2016) and physical function (Skelton et al., 1994; Hunter, Thompson and Adams, 2001; Haider et al., 2016; Ferreira et al., 2012; Fragala et al., 2014). Subjective and objective methods have been used to assess habitual PA, although the literature relating to older adults is scarce (Copeland and Esliger, 2009). Whilst frequently used, self-reported questionnaires have many limitations including recall bias, socially desirable responses (Sallis and Saelens, 2000) and the influence of mood state (Rikli, 2000). Accelerometery-based measurement of PA eliminates many of the subjective challenges associated with questionnaires (Copeland and Esliger, 2009) and is appropriate for use in older adult populations (Murphy, 2009).

Actigraph accelerometers are commonly used in PA research (Gorman et al., 2014; Sasaki, John and Freedson, 2011; Holmquist et al., 2017; Nawrocka, Mynarski and Cholewa, 2017). Accelerations, measured in three individual orthogonal planes, are processed to form a single composite vector magnitude which provides a measure of PA intensity. The tri-axial GT3X model has been validated and cut points determined for moderate, hard and very hard PA intensities (Sasaki, John and Freedson, 2011). However, cut points are not standardised, and studies have suggested different thresholds for PA intensities depending on the age, gender and health status of the population (Keadle et al., 2014; Santos-Lozano et al., 2013; Troiano et al., 2008; Freedson, Pober and Janz, 2005; Sandroff et al., 2014; Trost et al., 2012; Freedson, Melanson and Sirard, 1998). Interestingly, no cut points have been determined for wrist-worn accelerometers in older adults. This is surprising given that wrist-worn accelerometers may be able to capture activities that are commonly performed in older adult populations which typically involve less centre of mass (COM) movement, such as gardening and household cleaning.

PA estimates may also be affected by the location of the accelerometer on the body. Studies assessing PA in older populations have typically placed the accelerometer on the hip, although there is a lack of consensus on placement sites (Migueles et al., 2017). Some studies advocate hip placement (Ellis et al., 2014; Chen et al., 2003), whilst others have found comparable performance between hip and wrist-worn accelerometers (Ozemek et al., 2014; Zhang et al., 2012; Kamada et al., 2016). Contrary to these studies, Staudenmayer et al. (2015) demonstrated that wrist placement yielded greater accuracy for PA classification and Choi et al. (2012) found wrist placement on the dominant hand was more sensitive in detecting non-wear-time. Others support placement at the wrist indicating that it is robust enough for daily PA monitoring (Zhang et al., 2012) and shows superior wear-time compliance compared to hip-worn accelerometers (Kamada et al., 2016; Fairclough et al., 2016). Despite the benefits and detriments of different device locations, there appears to be a negligible difference between wrist-worn and hip-worn Actigraph accelerometers when determining PA in free-living older adults (Kamada et al., 2016).

3.1.5 Aims, Objectives and Hypotheses

The aim of this chapter was to establish whether there were significant differences in PA level, whole-body composition, handgrip strength and functional disability between the OG and YG. Variables with significant age-related differences would be used as potential covariates in the analyses of subsequent experimental studies. In order to accomplish the aim, objectives were to:

Table 3.2 *Objectives and hypotheses for chapter 3*

	Objective	Null Hypothesis
1	Measure LBP-related functional disability using participant's responses to the ODQ-m	ODQ-m scores will not be significantly different between the OG and YG
2	Measure whole-body fat mass and whole-body, trunk and appendicular lean mass using BIA	a) The OG will not have significantly greater whole-body fat mass than the YG b) The OG will not have significantly less whole-body, appendicular and trunk lean mass than the YG
3	Measure dominant and non-dominant handgrip strength using a handgrip dynamometer	Dominant and non-dominant handgrip strength will not be significantly weaker in the OG compared to the YG
4	Calculate the average daily time spent in moderate, moderate-to-vigorous and vigorous PA intensities using accelerometers	Average daily time spent in moderate, moderate-to-vigorous and vigorous PA intensities will not significantly differ between the OG and YG

3.2 Methods

Participants completed a range of assessments for variables that were considered potential covariates or exploratory factors. The ODQ-m, BIA and handgrip strength were completed in that order in a single session at Coventry University. Following this session, participants were given an accelerometer to wear continuously for one week.

3.2.1 ODQ-m Protocol

Participants completed the self-administered ODQ-m (**Figure 3.1**). ODQ-m scoring instructions were followed. The questionnaire consists of ten items addressing different aspects of function. Each item is scored out of a possible score of five with higher values representing greater disability. The total score was then divided by the total possible score of 50 and expressed as a percentage. The interpretation of scores was as follows: 0 – 20% indicated minimal disability, 21 – 40% indicated moderate disability, 41 – 60% indicated severe disability, 61 – 80% indicated that LBP impinges on all aspects of the participant's life and greater than 81% indicated that the participant was bed-bound (Alcántara-Bumbiedro et al., 2006).

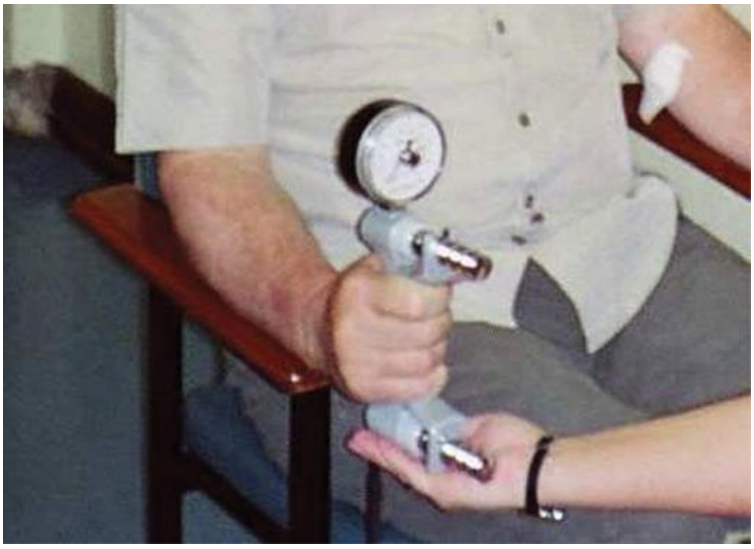
Figure 3.1 *Example of the Modified Oswestry Low Back Pain Disability Questionnaire*

3.2.2 Bio-electrical Impedance Analysis Protocol

The Tanita MC-780 (Tanita Corp, Tokyo, Japan) is a multi-frequency segmental body composition analyser with reasonable sensitivity and specificity to diagnose low levels of appendicular lean mass (ALM) in community-dwelling older adults (Verreijen et al., 2018). Participants were asked to fast four hours prior to the BIA assessment as recommended by the National Institute of Health (1996). Some research suggests that there is a relatively minor impact of food and drink consumption on BIA measures (Androutsos et al., 2015). However, others have shown that 20 minutes after eating, percent body fat significantly increases and remains elevated for 60 minutes postprandial (Dixon, Masteller and Andreacci, 2013). Participants wore light clothing and stood on the scale platform barefoot whilst the device was in standard mode. Once body mass had been determined by the scale, participants gripped the hand-grip electrodes in both hands and held them alongside their body throughout the measurement. This was consistent with previous BIA protocols for the Tanita MC-780 (Malczyk et al., 2016; Iizuka et al., 2015; Verney et al., 2015). Segmental fat and fat-free masses were recorded. ALM and trunk lean mass were reported separately.

3.2.3 Handgrip Strength Protocol

Handgrip strength was assessed using a handgrip dynamometer (Takei 5401, Takei Scientific Instruments Co Ltd, Japan). The Southampton protocol (Roberts et al., 2011) was adhered to (**Figure 3.2**) based on the recommendations of Schaap et al. (2016). Participants completed three maximal effort trials on each hand. The order of trials alternated between the dominant and non-dominant hand, with 30 seconds rest between each trial. This allowed 60 seconds recovery between trials of the same hand. According to Mathiowetz (1990), 60 seconds rest between trials is a sufficient to attenuate the cumulative effects of fatigue. The highest score achieved for each hand across the three trials was used for final analysis.



Posture	Subject seated, same chair for every measurement
Arm position	Forearms rested on the arms of the chair
Wrist position	Wrist just over the end of the arm of the chair, in a neutral position, thumb facing upwards
Lower extremity position	Feet flat on the floor
Encouragement	'I want you to squeeze as hard as you can for as long as you can until I say stop. squeeze, squeeze, squeeze, stop' (when the needle stops rising)
Number of trials	Three trials on each side, alternating sides
Score to use	Maximal grip score from all six trials used

Figure 3.2 *Southampton protocol for hand grip strength test (Roberts et al., 2011)*

3.2.4 Physical Activity Monitoring Protocol

Participants wore an Actigraph GT9X accelerometer (Pensacola, FL, US) on their dominant wrist for seven consecutive days (Copeland and Esliger, 2009; Ryan et al., 2018). The small and lightweight devices are minimally intrusive and record accelerations between ± 8 G at a rate of 30 - 100 Hz, which is suitable for data capture of ADLs (Bouten et al., 1997). A 90 Hz sampling frequency was chosen based on the recommendations of Migueles et al. (2017) when using manufacturer signal processing methods. According to Migueles and colleagues (2017), placement of the accelerometer should be chosen based on reliability, validity and compliance. The decision for wrist-worn accelerometers was therefore based on maximising wear-time compliance and data fidelity. Hip placement potentially lacks the sensitivity to capture less traditional modes of PA more commonly performed in older adults (Sallis et al., 1986; Lawlor et al., 2002; Walsh et al., 2001; Shephard, 2003). Furthermore, wrist-worn accelerometers are able to identify walking and running (the dominant moderate and vigorous activities in most adults) with a 98% or greater accuracy (Zhang et al., 2012) and are better at predicting activities with significant arm movement (Ellis et al., 2014). Finally, the dominant wrist was preferred due to greater PA classification accuracy than the non-dominant wrist, albeit a negligible difference (Zhang et al., 2012).

3.2.4.1 Accelerometer Data Analysis

Data were processed using dedicated software (Actilife, version 6.13, Pensacola, FL, US). To ensure data were representative of PA performed in a typical day and week, wear-time criteria were

established. To be valid, data must have been obtained for a minimum of four days including one weekend day and at least 10 hours of awake time during these days (Trost, Mciver and Pate, 2005; Migueles et al., 2017; Hagströmer, Oja and Sjöström, 2007; Ashe et al., 2008; Parker, Strath and Swartz, 2008; Kang et al., 2009; Ham and Ainsworth, 2010; Clark et al., 2011). Valid data were divided into 1 second epochs. Although no studies have investigated the influence of epoch length on accelerometer outcomes (Migueles et al., 2017), unpublished data suggest that shorter epochs (1 second vs 60 seconds) are more sensitive in detecting time spent in MVPA (Migueles et al., 2017).

Cut-points for moderate PA (MPA) and vigorous PA (VPA) have not been established for wrist-worn accelerometers in healthy older adult populations. Indeed, estimation of PA intensities are particularly difficult in older adults (Santos-Lozano et al., 2013). Colley and Tremblay (2011) derived moderate and vigorous intensity cut-points of 1535 and 3960 counts/minute, respectively. However, this was achieved using 60 second epochs in a younger adult sample. Copeland and Esliger (2009) defined MVPA cut-off as 1041 counts/minute in healthy older adults using hip-worn accelerometers. A similar cut-off of 1031 counts/minute for moderate intensity activities was identified by (Diaz et al., 2018) for wrist-worn accelerometers in healthy adults. Cut-off values for MPA (1031 counts/minute) and VPA (3589 counts/minute) intensities were chosen based on the recommendations of Diaz et al. (2018) due to their high sensitivity and specificity. Average time spent per day in MPA, MVPA and VPA intensities were calculated and used in the statistical analysis.

3.2.5 Statistical Analysis

Independent samples T-tests were performed to assess statistically significant differences between the young and older age groups. Alpha was set at 0.05. Effect sizes (Cohen's d) were also calculated to estimate the magnitude of the difference between age groups. Effect size estimates were considered small (d = 0.2), medium (d = 0.5) or large (d = 0.8) (Cohen, 2013). Data are presented as means with standard deviations (mean ± SD) unless otherwise stated. All data were normally distributed (Shapiro-Wilk test, $p > .05$) and homogeneous variances were assumed (Levene's test, $p > .05$).

3.3 Results

3.3.1 Modified Oswestry Low Back Pain Disability Questionnaire

The ODQ-m scores were not significantly different between the young (2.2 ± 2.3%) and old (2.2 ± 3.5%) groups ($t(22) = 0.00$, $p = 1.0$).

3.3.2 Whole-Body Composition

The YG had significantly more ($t(22) = -3.28$, $p = .003$) ALM (28.3 ± 3.1 kg) and significantly less ($t(22) = 2.62$, $p = .016$) whole-body fat mass (13.6 ± 4.9 kg) than the OG (24.1 ± 3.3 kg and 19.2 ± 5.4 kg, respectively). The magnitudes of these differences were large (Cohen's $d > 1.1$). No statistical differences were revealed for whole-body lean mass and trunk lean mass between the groups.

3.3.3 Handgrip Strength

Handgrip strength was greater in the YG than the OG. The YG's dominant handgrip strength (45.0 ± 7.5 kg) was significantly greater ($t(22) = -2.22$, $p = .037$) than the OG's (37.4 ± 9.1 kg). The YG's non-dominant hand (42.8 ± 5.3 kg) was also significantly stronger ($t(22) = -2.38$, $p = .027$) compared to the OG's (36.3 ± 7.9 kg). Effect sizes were large for both comparisons (Cohen's $d > 0.9$).

3.3.4 Physical Activity Level

There was a significant difference between age groups for average time spent in VPA ($t(22) = -2.371$, $p = .027$, Cohen's $d = 0.97$). The OG spent significantly less time in VPA intensities (2.07 ± 0.59 hours/day) than the YG (2.64 ± 0.59 hours/day) (**Figure 3.3**). Differences between age groups for time spent in MVPA ($p = .60$) and MPA intensities ($p = .51$) were not statistically significant.

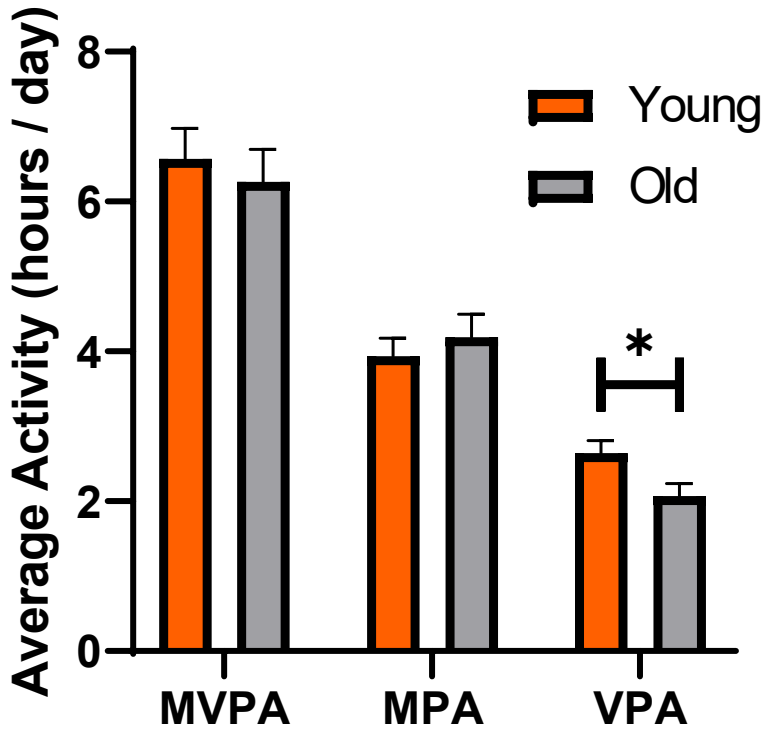


Figure 3.3 *Physical activity level presented as means with SEM. MVPA = moderate to vigorous physical activity, MPA = moderate physical activity, VPA = vigorous physical activity. * independent T-test: $p < .05$*

3.4 Summary

The current findings are consistent with other research showing a decline in VPA with ageing (Westerterp, 2000). Despite the known benefits of regular PA in reducing the risk of developing cardiovascular and metabolic diseases, obesity, falls and musculoskeletal disorders (Hamer, Lavoie and Bacon, 2014; Villareal et al., 2011; Gulsvik et al., 2012), participation in PAs remains low amongst older adults (McPhee et al., 2016). Although the current results show no significant difference in MPA with age, there is a dose-response relationship to suggest that more vigorous activities bring about greater health benefits (Swain and Franklin, 2006; Bruce, Fries and Hubert, 2008; Ebrahim, 2000; Kim, Adamson and Ebrahim, 2013; Wannamethee et al., 2005). This is particularly important for attenuating the detrimental effects of age-related sarcopenia. Higher intensity activities have been shown to be superior to lower intensities for improving strength in older adults (Steib, Schoene and Pfeifer, 2010). Therefore, VPA may combat the effects of sarcopenia (Steib, Schoene and Pfeifer, 2010)

by stimulating positive adaptations in muscle morphology (St-Jean-Pelletier et al., 2017) and improving strength and physical function in older adults (Chahal, Lee and Luo, 2014; Pau et al., 2014); where low-moderate intensities are less effective (Hamer, Lavoie and Bacon, 2014). However, it should be noted that the relationship between PA intensity and health outcomes is not always linear and indeed positive. This has been observed in endocrine function in older people, where high resistance training led to functional improvements whilst lower intensity exercises were more advantageous in terms of endocrine adaptations (Onambélé-Pearson, Breen and Stewart, 2010a). Furthermore, the addition of nutrition supplementation may diminish any additional benefit to metabolic adaptation, endocrine function and indeed strength improvements that high intensity PA may have over lower intensities (Onambélé-Pearson, Breen and Stewart, 2010b). Older adults may also be unable to cope with the demands of higher intensity PA, which may increase the risk of falls and musculoskeletal injury (Tiedemann et al., 2011) as well as increase the inflammatory response (Della Gatta et al., 2014).

It should also be noted that PA was measured by the dynamic acceleration of the accelerometer. Whilst this is widely accepted as the gold-standard approach to objectively measure habitual PA levels (Migueles et al., 2017), it does not consider the metabolic demands of the tasks being performed. This may confound comparisons between the OG and YG as walking has been shown to be more metabolically demanding in older men than younger men, which cannot be explained by changes in mechanical work (Mian et al., 2006). Despite limitations associated with the data collection methods, the difference in VPA between age groups must be accounted for when investigating age-related changes in muscle morphology, strength and function.

Given the large difference in handgrip strength between the OG and YG, it is an interesting exploratory factor that may associate with musculoskeletal outcomes of the lumbar spine. It is a well-known indicator of mobility and health status in older men (Desrosiers et al., 1995; Giampaoli et al., 1999), therefore, the association between handgrip strength and trunk function should be explored. It should be noted that the OG's scores for the dominant (37.4 ± 9.1 kg) and non-dominant (36.3 ± 7.9 kg) hands were high with respect to the sarcopenia cut-off value of 27 kg (Cruz-Jentoft et al., 2019). This indicates that the OG participants were high-functioning and not sarcopenic based on diagnostic criteria (Cruz-Jentoft et al., 2019). Therefore, caution should be taken when generalising findings in this thesis as participants were not representative of the wider population, rather a high-functioning and active sub-group within the general population.

The influence of whole-body composition should also be considered given the large differences in ALM and whole-body fat mass between age groups. When determining age-related changes in

musculoskeletal measures (i.e. muscle volume atrophy, fat infiltration, peak torque generation), whole-body fat mass and ALM should be considered potential covariates. In this sample, functional disability is unlikely to influence muscle morphology, strength and physical function. Low ODQ-m scores suggested that there was no evidence of LBP causing functional disabilities, and the high level of similarity between groups indicated that LBP was not likely to be a confounding factor.

Table 3.3 *Thesis Map*

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression		Aim	
		Key findings	
		Implications	
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	

Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men	Aim
	Key findings
	Implications
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men	Aim
	Key findings
	Implications

Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression

The work from this chapter has been published in a peer-reviewed journal.

Dallaway, A., Kite, C., Griffen, C., Duncan, M., Tallis, J., Renshaw, D., and Hattersley, J. (2020) 'Age-related degeneration of the lumbar paravertebral muscles: Systematic review and three-level meta-regression'. *Experimental Gerontology* 133, 110856

Chapter Abstract

Background Morphological changes of the lumbar spine muscles are not well characterised with ageing. To further the understanding of age-related degeneration of the lumbar spine musculature, normative morphological changes that occur within the paravertebral muscles must first be established.

Methods A systematic review and meta-regressions were conducted adhering to PRISMA guidelines. Searches for published and unpublished data were completed in June 2019.

Results Searches returned 4781 articles. 34 articles were included in the quantitative analysis. Three-level meta-analyses showed age-related atrophy (r = -0.26; 95% CI: -0.33, -0.17) and fat infiltration (r = 0.39; 95% CI: 0.28, 0.50) in the lumbar paravertebral muscles. Degenerative changes were muscle-specific and men (r = -0.32; 95% CI: -0.61, 0.01) exhibited significantly greater muscle atrophy than women (r = -0.24; 95% CI: -0.47, 0.03). Imaging modality, specifically ultrasound, also influenced age-related muscle atrophy. Measurements taken across all lumbar levels revealed the greatest fat infiltration with ageing (r = 0.58, 95% CI: 0.35, 0.74). Moderators explained a large proportion of between-study variance in true effects for muscle atrophy (72.6%) and fat infiltration (79.8%) models.

Conclusions Lumbar paravertebral muscles undergo age-related degeneration in healthy adults with muscle, lumbar level and sex-specific responses. Future studies should use high-resolution imaging modalities to quantify muscle atrophy and fat infiltration.

Key words: back muscles, lumbosacral region, sarcopenia, muscle degeneration, healthy ageing

CHAPTER 4

Table 4.1 *Thesis Map*

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and	Aim	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology

	composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Key findings Implications
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men		Aim Key findings Implications
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men		Aim Key findings Implications
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men		Aim Key findings Implications

CHAPTER 4

4.1 Introduction

Age-related degeneration of skeletal muscle is characterised by intramuscular fat infiltration and a loss of muscle tissue (Doherty, 2001; Delmonico et al., 2009; Cruz-Jentoft et al., 2010a; McGregor, Cameron-Smith and Poppitt, 2014). These, together with the concomitant loss of muscle force generation (Frontera et al., 2000; Kent-Braun and Ng, 2000; Doherty, 2001), are associated with poor functional outcomes as well as increased risk of morbidity and mortality (Baumgartner et al., 1998; Roubenoff and Hughes, 2000; Sayer et al., 2005; Gale et al., 2007; Cruz-Jentoft et al., 2010a; Landi et al., 2012; Arango-Lopera et al., 2013; Landi et al., 2013; Beaudart et al., 2017). Sarcopenia encompasses the interrelationships between deteriorating muscle morphology, physical function and strength (Cruz-Jentoft and Sayer, 2019). Adverse outcomes associated with sarcopenia are a major health concern and socioeconomic burden, resulting in estimated excess annual healthcare costs of £2.5b in the United Kingdom (Pinedo-Villanueva et al., 2019) and $18.5b in the United States (Janssen et al., 2004). Research on sarcopenia has predominantly focused on the systemic loss of muscle and its impact on physical function (Batsis et al., 2013; Bahat et al., 2016). However, a systemic approach to understand sarcopenia may not be appropriate due to the muscle and location-specific nature of its progression (Candow and Chilibeck, 2005; Abe et al., 2014). Whilst studies have examined degeneration of the appendicular muscles (von Haehling, Morley and Anker, 2010; Müller et al., 2014; Cawthon et al., 2015; Woo and Leung, 2016) there is a paucity of available research focusing on age-related changes in the trunk musculature. This has been acknowledged by other researchers (Crawford et al., 2016c; Kalichman, Carmeli and Been, 2017) despite the importance of paravertebral muscles in the maintenance of spinal health and physical function being increasingly recognised (Hicks et al., 2005b; Goubert et al., 2016; Kalichman, Carmeli and Been, 2017; Crawford et al., 2019a). Although age is known to influence paravertebral muscle morphology and attempts have been made to characterise degeneration of the paravertebral muscles with the natural ageing process (Meakin et al., 2013; Fortin et al., 2014; Valentin, Licka and Elliott, 2015; Crawford et al., 2016a; Kalichman, Carmeli and Been, 2017; Lee et al., 2017; Shahidi et al., 2017b; Burian et al., 2018) the phenomenon is not fully understood.

The paravertebral muscles (i.e. MF, ES, PS and QL) all contribute to the stability of the lumbar spine (Santaguida and McGill, 1995; McGill, 2001; Barr, Griggs and Cadby, 2005); although the anatomy and biomechanics of the MF demonstrate that it is the most suited to this role (Macintosh and Bogduk, 1986; Moseley, Hodges and Gandevia, 2002; MacDonald, Lorimer Moseley and Hodges, 2006; Ward et al., 2009). The larger more superficial muscles surrounding the lumbar region function primarily as torque generators for spinal movement. The PS acts primarily as a flexor muscle of the hip (Bogduk, Pearcy and Hadfield, 1992), the ES function primarily as extensor muscles (Potvin, McGill and Norman,

1991) and the QL brings about lateral flexion although its role in spinal biomechanics is undetermined (Phillips, Mercer and Bogduk, 2008). Senescence of the LPMs may have greater functional consequences compared to the appendicular muscles (Hicks et al., 2005a; Eguchi et al., 2017). However, whereas efforts have been made to reach consensus of a reference standard for the measurement of appendicular muscle mass in sarcopenia (Buckinx et al., 2018; Cruz-Jentoft et al., 2019), such efforts have yet to translate to measurements of muscle morphology in the lumbar spine, resulting in disparate methods amongst studies.

Relatively few studies have measured the morphology of all the four main LPMs. Indeed, previous systematic reviews focusing on paravertebral muscle degeneration have investigated the morphology of the MF and ES without examining the PS and QL (Hebert et al., 2009; Fortin and Macedo, 2013). Given the different functions of the LPMs and their potential for localised degeneration in diseased and healthy populations (Ploumis et al., 2011; Min et al., 2013; Crawford et al., 2016c; Baracos, 2017), normative features are of interest for each individual muscle surrounding the lumbar spine. Furthermore, there has been limited investigation into both muscle size and fat composition of the paravertebral muscles. These measurements have been typically performed at a single representative slice in the lumbar region (Parkkola, Rytokoski and Kormano, 1993; Gibbons et al., 1997; Watson, McPherson and Starr, 2008; Ikezoe et al., 2012; Yoshizumi et al., 2014; Hiepe et al., 2015; Frost and Brown, 2016; Hamaguchi et al., 2016; Kim et al., 2017; Burian et al., 2018; Ebadi et al., 2018; Hedermann et al., 2018; Kalafateli et al., 2018; Maltais et al., 2018; Rahmani et al., 2019) resulting in CSAs despite volumetric information being preferable due to its greater association with muscle function (Boom et al., 2008). Inconsistent imaging modalities and image analysis techniques across studies, as well as different measures representing muscle size and quality, also confound comparisons between studies.

The considerable variation in methodological factors across studies makes comparing findings difficult, which has hampered our understanding of changes in lumbar muscle morphology with ageing. A necessary step to better understand this age-related phenomenon is to conduct a systematic review and meta-analysis. To the authors' knowledge, a quantitative analysis of the research on this topic has not been performed to date. Therefore, bringing together the evidence and accounting for methodological differences will establish a reference for normal age-related degenerative features of LPM morphology and provide recommendations for future studies. Furthermore, findings from the moderators included in the meta-regression model will provide practical guidance for collecting primary data relating to LPM morphology.

4.1.1 Aims, Objectives and Hypotheses

The aim of this study was to perform a quantitative analysis of the literature to establish the relationship between normal ageing and LPM degeneration. A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of LPM morphology. In order to accomplish these aims specific objectives were to:

Table 4.2 *Objectives and hypotheses for chapter 4*

	Objective	Hypotheses
1	Perform a scoping search of the literature on normative degeneration of the lumbar musculature	n/a
2	Develop a protocol and search strategy using appropriate terms	n/a
3	Perform a systematic review of the available literature	n/a
4	Meta-analyse available data to establish the effect of age on atrophy and fat infiltration in the lumbar spine muscles	a) the LPMs will not atrophy with ageing b) the LPMs will not increase in fat content with ageing
5	Use appropriate methods to account for statistical dependency amongst effect sizes	a) heterogeneity within studies will not be significant b) heterogeneity between studies will not be significant
6	Identify key moderators that are responsible for methodological variation amongst studies	a) imaging modality will not moderate the relationship between ageing and LPM degeneration b) LPM degeneration will not be significantly different between sexes c) age-related LPM degeneration will not be muscle-specific d) lumbar level will not moderate the relationship between ageing and LPM degeneration

4.2 Materials and Methods

4.2.1 Protocol and Registration

This systematic review was registered on the Prospero International Prospective Register of Systematic Reviews (CRD42018093157) and is reported based on the guidelines of the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) statement (Moher et al., 2009).

4.2.2 Search Methods for Identification of Studies

To assess the relationship between healthy ageing and changes in muscle morphology, data were sought from eligible studies. **Table 4.3** presents the eligibility criteria for inclusion in this systematic review. Although it can be questioned how baseline data from experimental studies may represent age-related muscle degeneration, in the current study baseline data were treated as cross-sectional observations and deemed eligible provided the inclusion criteria were met. To meet the inclusion criteria for exposure, studies had to show ageing as a generally healthy process, stating that participants were healthy, physically independent and free from disease likely to affect paravertebral muscle morphology (e.g. spondylolisthesis, low back pain, stroke and cancer). This was not exhaustive as shown by the MeSH description for “healthy ageing”, and due to the lack of consensus on a definition for healthy ageing (Peel, Bartlett and McClure, 2004). If a study reported disease cases within an otherwise healthy sample, data were sought for the healthy participants only. If the health status of participants was unclear or ambiguous, confirmation was sought from the author(s).

Table 4.3 *Eligibility criteria for including studies in this systematic review*

Inclusion criteria:

1. Study design: observational and baseline data from experimental studies
2. Population: healthy sample including adults older than 40 years of age with an age range of at least ten years. If age is a dichotomous variable, older group's mean age must be greater than 40 years and at least 10 years greater than the younger group's mean age. Longitudinal studies must have a minimum follow-up of 10 years and the sample's mean age must be greater than 40 years at follow-up.
3. Exposure: healthy ageing
4. Comparator: not required. If present, comparison group must meet the inclusion criteria for exposure and have a mean age more than ten years younger than the older group's mean age
5. Expected outcomes: quantitative measures of muscle size (atrophy) or quality (fat infiltration); Imaging modality – magnetic resonance imaging (MRI), computerised tomography (CT) or ultrasound; Lumbar level(s) of measurement – L1-L5/S1; Muscles measured – measurements include psoas, erector spinae, quadratus lumborum and or multifidus

Exclusion criteria:

1. Study design: case series, case reports, preclinical studies, reviews and meta-analyses
2. Population: sample contains no participants aged over 40 years. If age is a dichotomous variable, older group's mean age equal to or less than 40 years or within ten years of the comparison group's age. Longitudinal studies' follow-up period is less than ten years or sample's mean age equal to or less than 40 years at follow-up
3. Exposure: evidence of disease or impairment that is likely to affect lumbar paravertebral muscle morphology
4. Comparator: if reported, comparison group shows evidence of disease or impairment, or has a mean age within ten years of the older group's mean age
5. Outcomes: semi-quantitative and qualitative measures of muscle size or quality; Imaging modality – use of imaging modality other than MRI, CT or ultrasound; Lumbar level(s) of measurement – does not include measurements with L1-L5/S1; Muscles measured – measurements do not include psoas, erector spinae, quadratus lumborum and or multifidus

4.2.3 Information Sources and Data Extraction

A search strategy was developed by (AD) for PubMed (**Table 4.4**), which was adapted to the syntax and appropriate subject headings of the other databases. The databases searched were MEDLINE and CINAHL (via EBSCOhost), PubMed, The Cochrane Central Register of Controlled Trials (CENTRAL), and EMBASE (via OvidSP). No study design, date or participant demographic restrictions were imposed on the search to ensure literature saturation. An English language restriction was used due to resource limitations. Final searches were completed June 1st, 2019. After initial searches were completed and

duplicate records removed (AD), titles and abstracts were screened independently and in duplicate by AD and Griffen, C. (CG) against the eligibility criteria. Unpublished data and grey literature were sought through handsearching reference lists of included articles and searching electronic grey literature databases (i.e. OpenGrey) to ensure a more comprehensive search strategy and reduce the possibility of publication bias (Paez, 2017). Articles not excluded based on title and abstract and deemed relevant progressed to full-text review. Full-text eligibility screening was completed independently by two reviewers (AD, CG) and reasons for exclusion were provided. Disagreements on eligibility were resolved by discussion. Whilst it was planned that unresolved disagreements would be arbitrated independently by a third reviewer, Hattersley, J. (JH), this was never exercised due to the reviewers reaching consensus in all discussions. Where studies were described in multiple publications, the publication with the most comprehensive data was used as the primary reference, excluding the others if the same data were presented. Where multiple publications from the same study but different data were retrieved, all relevant publications were included. If data could not be obtained from the full-text or if clarification was required, authors were contacted by one reviewer (AD). If sufficient data could not be obtained for a study, the study was excluded. Two reviewers (AD, CG) extracted data independently from eligible studies on: study design; sample and comparator information [sample size, gender, mean age, age range, mean body mass index (BMI), ethnicity, additional information about the setting, definition of health status]; imaging modality; image analysis outcome measures; lumbar level(s) and paravertebral muscle(s) measured; study results including statistical findings and overall conclusions.

Table 4.4 *PubMed search strategy*

#1 Paraspinal muscles MeSH Terms
#2 Paraspinal musc* Title/Abstract
#3 Back muscles MeSH Terms
#4 Back musc* Title/Abstract
#5 Multifidus Title/Abstract
#6 Lumbar multifidus Title/Abstract
#7 Lumbar musc* Title/Abstract
#8 Trunk musc* Title/Abstract
#9 Paravertebral musc* Title/Abstract
#10 #1 OR #2 OR #3 OR #4 OR #5 OR #6 OR #7 OR #8 OR #9
#11 Aged MeSH Terms
#12 Aged Title/Abstract
#13 Age Title/Abstract
#14 Aging MeSH Terms
#15 Aging Title/Abstract
#16 Ageing Title/Abstract
#17 Elderly MeSH Terms

#18 Elderly Title/Abstract
#19 Older adult* Title/Abstract
#20 #11 OR #12 OR #13 OR #14 OR #15 OR #16 OR #17 OR #18 OR #19
#21 Atroph* Title/Abstract
#22 Muscular atrophy MeSH Terms
#23 Spinal Muscular Atrophy MeSH Terms
#24 Degenerat* Title/Abstract
#25 Morpho* Title/Abstract
#26 Morphology MeSH Terms
#27 Size Title/Abstract
#28 Attenuation Title/Abstract
#29 Infiltration Title/Abstract
#30 Replacement Title/Abstract
#31 Sarcopen* Title/Abstract
#32 Sarcopenia MeSH Terms
#33 #21 OR #22 OR #23 OR #24 OR #25 OR #26 OR #27 OR #28 OR #29 OR #30 OR #31 OR #32
#34 #10 AND #20 AND #33
#35 Animals MeSH Major Topic NOT Humans MeSH Major Topic
#36 #34 NOT #35

4.2.4 Assessment of Risk of Bias in Individual Studies and Study Quality

Risk of bias was assessed independently by two reviewers (AD, CK) at the study level using the National Institutes of Health (NIH) Study Quality Assessment Tools. Reviewers used the study rating tools to rate the quality of the study as good, fair or poor. The Risk of Bias Assessment Tool for Nonrandomised Studies (RoBANS) (Park et al., 2011; Kim, S. Y. et al., 2013) was also used (AD, CK) to independently assess risk of bias at the outcome level. A judgement of “low”, “high” or “unclear” was assigned to each question for all included studies. If ratings using the NIH Study Quality Assessment tool or judgements using the RoBANS tool differed between reviewers, reviewers discussed the study in an effort to reach consensus, otherwise a third reviewer (JH) arbitrated disagreements not due to assessor error.

4.2.5 Synthesis of Results and Statistical Methods

Standardised effect sizes were used in the meta-analysis due to studies using different measurement scales. Pearson’s product-moment correlation coefficient (r) was the principal summary measure. For studies reporting ageing as a continuous variable, correlations (r) were transformed into Fisher’s z units (z’) to approximate normally distributed data. Data were excluded from the meta-analysis when studies used non-parametric statistical tests as non-parametric data violate the meta-analytical

assumption of normal distribution. For studies that reported ageing as a dichotomous variable, the standardised mean difference (Cohen's d) was calculated. Cohen's d values were then converted into Fisher's z units (Borenstein et al., 2009; Polanin and Snilstveit, 2016). To account for the large variability in spinal-level measurements and different slice orientations, evaluations were categorised into high (L1-L2), mid (L2/3-L3/4), low (L4-L5/S1) and all (combined measurements across high, mid and low levels) lumbar levels. If a study contributed multiple effect sizes, differing only by lumbar level measurements, they were aggregated into appropriate categories. For example, if a study measured PS CSA at the L1 and L2, these two effect sizes were aggregated to provide one effect size at the "high" level.

The "metaSEM" package (Cheung, 2014a) was used in the RStudio (RStudio Team (2015). RStudio: Integrated Development for R. RStudio, Inc., Boston, MA URL http://www.rstudio.com/. Version 1.1.463) environment to perform three-level meta-analyses. Level 1 referred to participants within studies, level 2 (within-study variance) referred to interdependent effects within studies, and level 3 (between-study variance) referred to the studies themselves. This approach allowed informative differences to be fully explored between outcomes whilst accounting for statistical dependency due to studies contributing multiple effect sizes. The three-level meta-analytical model was also adopted as the dependency between effect-sizes was unknown (Cheung, 2014a, 2014b). Due to the complexity of the data obtained, traditional meta-analytical methods were not appropriate and would have likely artificially reduced variance within and between studies (Cheung and Chan, 2008). Moderators were included in the models to assess their influence on the effect size estimate and to investigate the amount of between-study variance in true effects that could be explained by their inclusion. Categorical moderators included:

1. sex: female*, male;
2. muscle: psoas*, erector spinae, multifidus, quadratus lumborum, combined paraspinals (erector spinae + multifidus), combined paravertebral muscles (all four muscles);
3. level: all*, high, mid, low; and
4. imaging modality: CT*, MRI, ultrasound.

Asterisks denote the reference category. Dummy codes were created for categorical moderators for entry into the meta-regression models.

In addition, age (mean and range) and mean BMI were included as continuous. Continuous covariates were centred, but not standardised, to increase numerical stability. Additionally, random-effects meta-analyses, with effects aggregated within studies, were performed using the R-package "metaphor" (Viechtbauer, 2010) to estimate the robustness of the three-level meta-analyses.

Moderator coefficients and summary effects (z') were transformed back to correlation coefficients (r) with their 95% confidence intervals. Before performing any meta-analyses, a Baujat plot was visually inspected to identify and remove effects that excessively contributed to heterogeneity and the overall result (Baujat et al., 2002). For muscle size, three effect-sizes (Aboufazeli et al., 2018; Hedermann et al., 2018), and for fat infiltration, two effect-sizes (Masaki et al., 2015; Frost and Brown, 2016) lay away from the majority and were deemed outliers. Sensitivity analyses were performed to explore how the main findings were affected by the removal of studies that: a) did not explicitly state that their sample were healthy and with a normal BMI (18.5-24.9) and b) were rated as fair or poor quality based on the NIH quality assessment tools.

4.2.6 Investigation of Heterogeneity and Explained Variance in True Effects

Heterogeneity within (level 2) and between studies (level 3) was evaluated using the Chi-squared test and I^2 statistic. The I^2 statistic describes the percentage of variability in the point estimates that is due to heterogeneity rather than sampling error (Deeks, 2011). Interpretation of heterogeneity followed Deeks' (2011) suggestion that 0-40% might not be important, 30-60% may represent moderate heterogeneity, 50-90% may represent substantial heterogeneity and 75-90% considerable heterogeneity. The percentage of variance in true effects (R^2) explained by the inclusion of moderators was calculated (Konstantopoulos and Hedges, 2009).

4.2.7 Assessment of Risk of Bias Across Studies

To explore publication bias potential, asymmetry was inspected visually using funnel plots and statistically using Egger's regression intercept test (Egger et al., 1997), for which there were a sufficient number of studies. Sutton et al. (2000) suggest that five studies is usually too few to allow the detection of an asymmetric funnel. Duval and Tweedie's trim and fill test (Duval and Tweedie, 2000) was performed if publication bias was indicated, providing a revised summary point estimate adjusted for publication bias.

4.3 Results

4.3.1 Study Selection

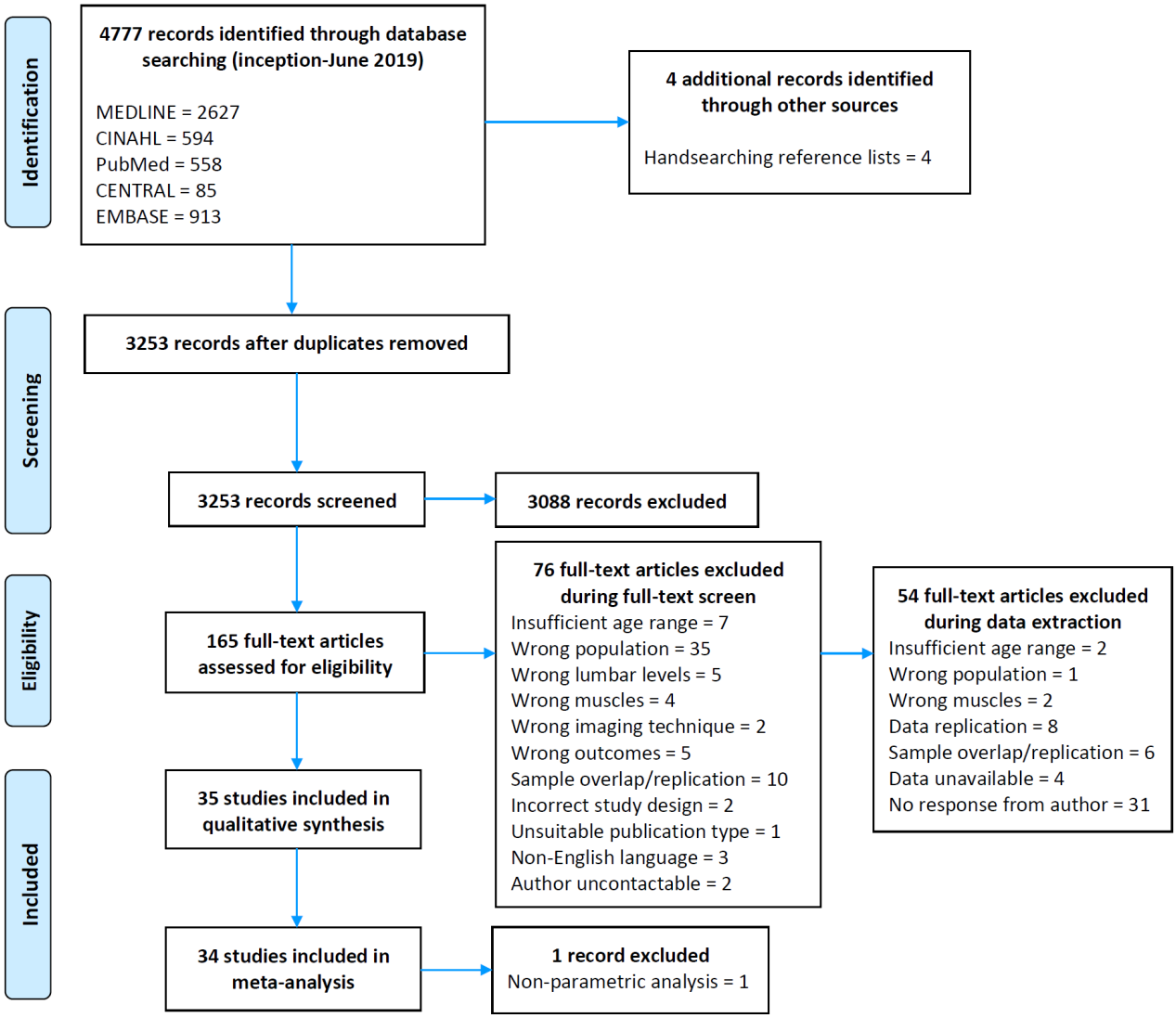


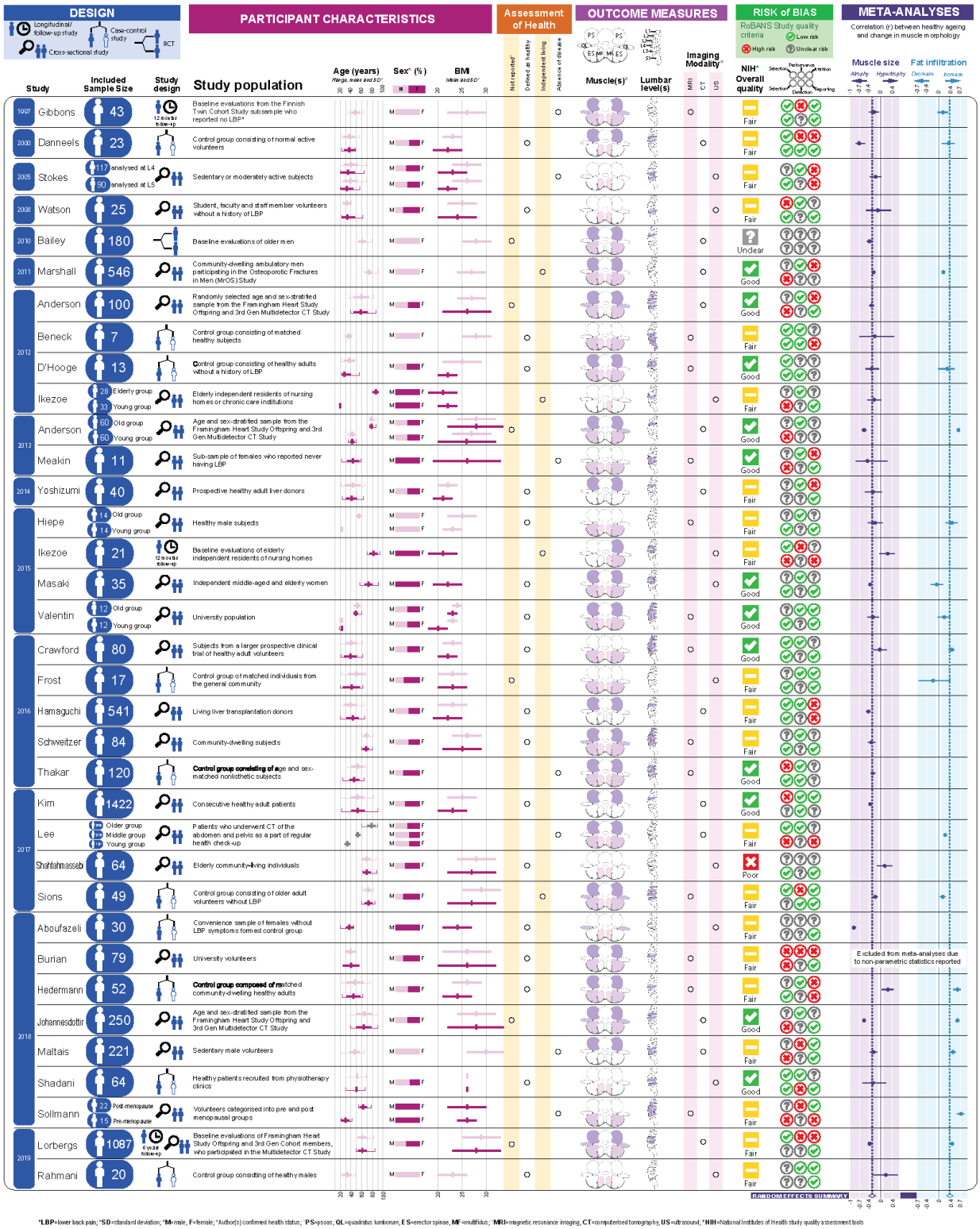
Figure 4.1 *PRISMA flow diagram depicting the selection process for studies*

The flow diagram (**Figure 4.1**) presents the study selection process applied in this meta-analysis. Of the 35 studies (Gibbons et al., 1997; Danneels et al., 2000; Stokes, Rankin and Newham, 2005; Crawford et al., 2016a; Lee et al., 2017; Burian et al., 2018; Watson, McPherson and Starr, 2008; Bailey et al., 2010; Marshall et al., 2011; Anderson et al., 2012; Beneck and Kulig, 2012; D'Hooge et al., 2012; Ikezoe et al., 2012; Anderson et al., 2013; Meakin et al., 2013; Yoshizumi et al., 2014; Hiepe et al., 2015; Ikezoe et al., 2015; Masaki et al., 2015; Valentin, Licka and Elliott, 2015; Frost and Brown, 2016; Hamaguchi et al., 2016; Schweitzer et al., 2016; Thakar et al., 2016; Kim et al., 2017; Shahtahmassebi et al., 2017; Sions et al., 2017b; Aboufazeli et al., 2018; Hedermann et al., 2018; Johannesdottir et al.,

2018; Maltais et al., 2018; Shadani et al., 2018; Sollmann et al., 2018; Lorbergs et al., 2019; Rahmani et al., 2019) included in the qualitative synthesis, 32 reported outcomes for muscle size (n = 5523) and 18 studies reported outcomes for muscle quality (fat infiltration) (n = 3471). These studies comprised of one randomised controlled trial whilst all others adopted observational study designs. Of these, 10 studies originated from North America, 12 from European countries, 11 from Asian countries and 2 from Australia. For studies where separate data by sex were obtained, studies reporting on muscle atrophy with ageing as a continuous variable involved 2860 male and 2430 female participants. Studies comparing muscle size between older and younger groups involved 50 males and 64 females in the older group and 50 males and 69 females in the younger group. For studies reporting on muscle fat infiltration with ageing as a continuous variable, 1615 males and 997 female participants were included. Studies comparing muscle fat infiltration between older and younger groups involved 171 males and 186 females in the older group and 293 males and 209 females in the younger group. Across all studies, age ranged from 18 to 94 years for women, whilst for men age ranged from 18 to 92 years. Women's mean BMI was lower than men's and ranged from 20.5 to 28.0, whereas men exhibited a mean range of 22.2 to 30.4, discounting younger comparison groups. Further details on study design, population characteristics, assessment of health, outcome measures and study quality are presented as a graphical overview in **Table 4.5** for each included study. For the three-level meta-analytical model on age-related muscle atrophy, 29 studies were included giving 144 correlation coefficients. For the three-level model on age-related fat infiltration, 16 studies encompassing 92 correlation coefficients were included.

Table 4.5 *Graphical overview of study characteristics.*

Visual Summary **Graphical overview** comparing participant characteristics, evidence quality, risk of bias and findings of the included studies.



*LBP=lower back pain; SD=standard deviation; M=male, F=female; Author(s) confirmed health status; PS=psoas, QL=quadratus lumborum, ES=erector spinae, MF=multifidus; MRI=magnetic resonance imaging, CT=computerised tomography, US=ultrasound; NIH=National Institutes of Health study quality assessment tools

Note: Unable to obtain sufficient image quality in MS Word; please refer to ***Appendix j****.*

4.3.2 Assessment of Risk of Bias in Included Studies

A risk of bias summary is presented in **Figure 4.2** with the reviewers' judgements on overall study quality and on each domain included in the graphical overview of study characteristics (**Table 4.5**).

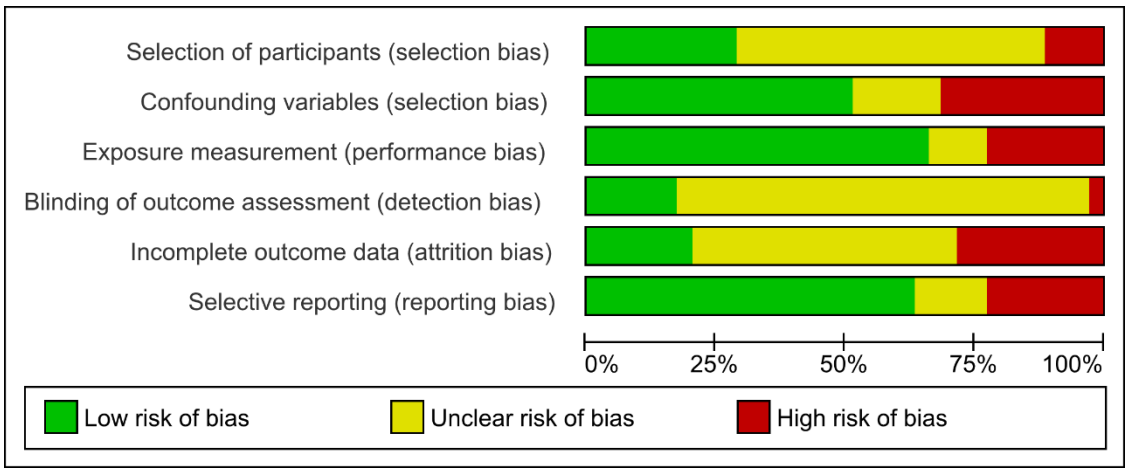


Figure 4.2 *Risk of bias summary: review of authors' judgements on each item from the Risk of Bias Assessment Tool for Nonrandomised Studies (RoBANS) presented as percentages across all included studies*

4.3.3 Overall Summary

Random-effects meta-analyses were performed where each study contributed one effect size. The correlation (with its 95% Wald CI's) between healthy ageing and change in LPM size was estimated at r = -0.25 (-0.33, -0.18, $p < .001$). For change in intramuscular fat infiltration with ageing, the overall correlation was r = 0.38 (95% CI: 0.27, 0.49, $p < .001$). These correlations were similar to those obtained from the three-level meta-analyses (**Table 4.6**). To assess the robustness of the three-level models, the null hypothesis: $\tau^2_{(3)} = 0$ was tested. Likelihood-ratio tests for the muscle size model (-2LL (df_1) = 54.6, $p < .001$) and muscle quality model (-2LL (df_1) = 56.3, $p < .001$) demonstrated that the three-level models were statistically better than the two-level models.

Table 4.6 *Three-level meta-analysis models for age-related muscle atrophy and fat infiltration in the lumbar paravertebral muscles*

	No. of studies	No. of effects	Effect size (r)	95% CI	*p*
Three-level muscle atrophy model					
Intercept	29	144	-0.255	-0.333, -0.169	< 0.001
Model summary	Level 2: $\tau^2_{(2)}$ = 0.004 (SE = 0.002), p = .05, I^2 = 6.60% (95% LBCI 1.8% 17.5%) Level 3: $\tau^2_{(3)}$ = 0.039 (SE = 0.014), p < .01, I^2 = 73.97% (95% LBCI 56.3% 86.0%) Q(df_{143}) = 367.44, p < .001, -2LL(df_{141}) = -26.29				
Three-level fat infiltration model					
Intercept	16	92	0.394	0.278, 0.499	< 0.001
Model summary	Level 2: $\tau^2_{(2)}$ = 0.006 (SE = 0.002), p < .05, I^2 = 7.54% (95% LBCI 2.4% 20.0%) Level 3: $\tau^2_{(3)}$ = 0.059 (SE = 0.025), p < .05, I^2 = 79.84% (95% LBCI 61.7% 91.1%) Q(df_{91}) = 411.96, p < .001, -2LL(df_{89}) = -11.06				

CI = Wald confidence intervals; LBCI = likelihood-based confidence intervals; -2LL = -2 log likelihood

For the random-effects meta-analyses, examination of the I^2 statistic suggested a considerable level of heterogeneity (muscle size model: I^2 = 94%, Q(df_{29}) = 223.5, p < .001; muscle quality model: I^2 = 98%, Q(df_{16}) = 464.1, p < .001). To explore potential reasons for heterogeneity, a sub-group analysis was performed by grouping study sample effect sizes by sex. One study analysing muscle size (Yoshizumi et al., 2014) and one analysing muscle fat infiltration (Lee et al., 2017) combined sexes in their analysis; these studies were removed from further analyses. For the muscle atrophy model, the random-effects meta-analysis produced summary effects of r = -0.22 (95% CI: -0.31, -0.13, p < .001) for females and r = -0.32 (95% CI: -0.40, -0.23, p < .001) for males, which were similar to those obtained in the three-level meta-regression model (**Table 4.7**). For the fat infiltration model, the random-effects meta-analysis produced summary effects of r = 0.42 (95% CI: 0.25, 0.57, p < .001) for females, and r = 0.44 (95% CI: 0.31, 0.55, p < .001) for males. However, these correlations were considerably less than those obtained from the three-level meta-regression (**Table 4.8**).

Substantial heterogeneity was still apparent in both muscle atrophy (females I^2 = 71%, Q(df_{21}) = 48.9, p < .001; males I^2 = 74%, Q(df_{24}) = 65.3, p < .001) and fat infiltration (females I^2 = 82%, Q(df_{11}) = 46.9, p < .001; males I^2 = 82%, Q(df_{12}) = 71.4, p < .001) random-effect meta-analyses when sub-grouped for sex. The three-level models also revealed greater variance between studies (level 3) than within studies (level 2) (**Table 4.6**), which was supported by rejection of the null hypothesis: $\tau^2_{(2)} = \tau^2_{(3)}$ for both muscle size (-2LL (df_1) = 15.3, p < .001) and muscle quality (-2LL (df_1) = 14.0, p < .001) models. This indicates moderators are more likely to exist between than within studies. Therefore, potential

moderators (Sex, Muscle, Level, Imaging technique, Mean sample BMI, Mean sample age, Sample age range) were examined using meta-regression to further explore reasons for between-study variance.

4.3.4 Three-level Meta-regression Models

The potential differentiating role of moderators on the overall relationship between changes in muscle morphology and ageing in healthy older adults were evaluated using three-level meta-regression. After controlling for other potential covariates that may influence the relationship between the change in muscle size and ageing, males (r = -0.32) differed significantly ($p < .01$) with females (r = -0.24). Muscle as a group was approaching significance ($p = .06$), whilst the ES (r = -0.32) and QL (r = -0.33) were significant individual muscle moderators ($p = .01$ and $p < .01$, respectively). There was a significant moderation with the average correlation obtained in studies using ultrasound (r = 0.08, $p < .001$); this was reflected in imaging modality reaching significance as a group ($p < .01$). Moderators with their regression coefficients are presented in **Table 4.7**. The inclusion of all moderators explained 72.6% of between-study variance in true effects (**Figure 4.3**), although the significant moderators alone explained 63.5%.

Table 4.7 *Three-level meta-regression model estimating the moderating effects of Sex (Female = reference category), Muscle (Psoas = reference category), Level (All levels = reference category), Imaging technique (CT = reference category), BMI, mean age, and age range on the relationship between change in paravertebral muscle size and ageing*

Moderator	β	SE	95% CI	*P*
Intercept*	-0.24	0.14	-0.47, 0.03	.02
Male**	-0.08	0.03	-0.14, -0.02	< .01
Muscle (ΔLL (df_5) = 10.54, p = .06)				
Erector spinae*	-0.08	0.03	-0.14, -0.02	.01
Multifidus	-0.03	0.06	-0.14, 0.08	.55
Quadratus lumborum**	-0.09	0.03	-0.15, -0.03	< .01
Paraspinals	0.05	0.14	-0.09, 0.19	.50
Combined paravertebrals	-0.18	0.07	-0.44, 0.10	.20
Level (ΔLL (df_3) = 1.79, p = .62)				
High levels	0.00	0.14	-0.27, 0.28	.98
Mid levels	-0.07	0.13	-0.32, 0.18	.59
Low levels	-0.05	0.13	-0.30, 0.21	.70
Imaging modality** (ΔLL (df_2) = 9.91, p < .01)				
MRI	0.13	0.08	-0.02, 0.28	.09
Ultrasound***	0.32	0.10	0.14, 0.48	< .001
BMI	0.00	0.03	-0.06, 0.07	.95
Mean age	0.02	0.03	-0.04, 0.09	.49
Age range	-0.03	0.03	-0.09, 0.03	.28
Level-2 variance	0.002	0.001	-0.001, 0.005	.23
Level-3 variance	0.011	0.006	-0.002, 0.023	.09

# of studies = 29, k = 144 correlation coefficients, Q(143) = 367.44; p < .001, -2LL(127) = -64.58, *< .05, **< .01, ***< .001

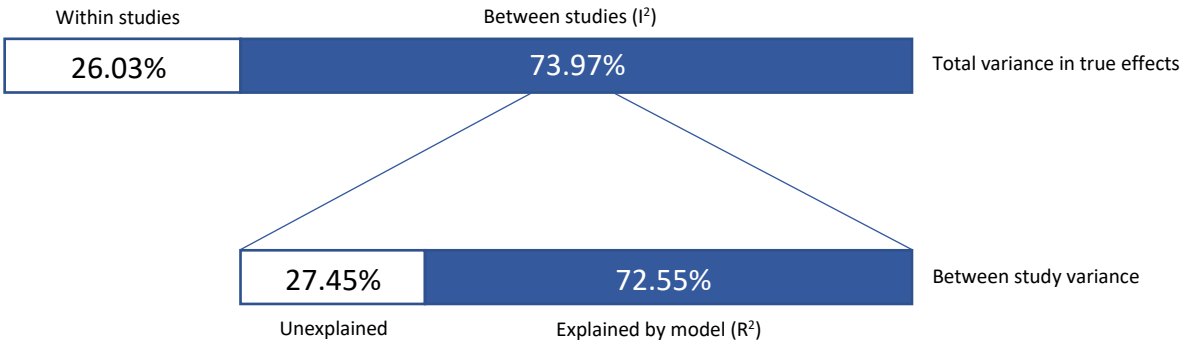


Figure 4.3 *Visual representation of the amount of between-study variance in true effects explained by moderators for the relationship between muscle atrophy and ageing*

The overall relationship between intramuscular fat infiltration and ageing was moderated by the selection of paravertebral muscle (p < .001), after controlling for other potential covariates. Within this group, the ES (r = 0.73), QL (r = 0.68) and paraspinals (r = 0.82) were significant individual

moderators (p < .001). Level of lumbar measurement also made a significant difference in the estimated correlation between fat infiltration and ageing (p = .03). Measurements at the high (r = 0.12, p = .002), mid (r = 0.24, p = .012) and low (r = 0.22, p = .009) lumbar levels differed significantly with measurements taken across all lumbar levels (r = 0.58). BMI was close to having a significant moderating effect on overall relationship (β = 0.10, p = .08). Age range of the sample was however significant and had an even greater influence on the relationship between fat infiltration and ageing (β = 0.16, p < .001). Moderators with their regression coefficients are presented in **Table 4.8**. The inclusion of all moderators explained 79.8% of between-study variance in true effects (**Figure 4.4**), although the significant moderators alone explained 65.8%.

Table 4.8 *Three-level meta-regression estimating the moderating effects of Sex (Female = reference category), Muscle (Psoas = reference category), Level (All levels = reference category), Imaging technique (CT = reference category), BMI, mean age, and age range on the relationship between change in paravertebral muscle fat infiltration and ageing*

Moderator	β	SE	95% CI	*P*
Intercept***	0.58	0.15	0.35, 0.74	< .001
Male	-0.05	0.03	-0.11, 0.02	.18
Muscle*** (ΔLL (df_4) = 29.59, p < .001)				
Erector spinae***	0.15	0.03	0.09, 0.20	< .001
Multifidus	0.08	0.08	-0.08, 0.23	.33
Quadratus lumborum***	0.10	0.03	0.05, 0.16	< .001
Paraspinals***	0.24	0.06	0.13, 0.34	< .001
Level* (ΔLL (df_3) = 8.85, p = .03)				
High levels**	-0.46	0.16	-0.67, -0.18	.002
Mid levels*	-0.34	0.14	-0.56, -0.08	.012
Low levels**	-0.36	0.14	-0.58, -0.09	.009
Imaging modality (ΔLL (df_2) = 1.40, p = .50)				
MRI	-0.04	0.10	-0.23, 0.15	.66
Ultrasound	-0.24	0.20	-0.57, 0.15	.23
BMI	0.10	0.06	-0.01, 0.22	.08
Mean age	-0.05	0.06	-0.16, 0.06	.36
Age range***	0.16	0.04	0.08, 0.25	< .001
Level-2 variance	$1e^{-10}$	0.001	-0.002, 0.002	.99
Level-3 variance	0.012	0.007	-0.001, 0.025	.08

# of studies = 16, k = 92 correlation coefficients, Q(91) = 411.96, p < .001, -2LL(76) = -64.83,
*< .05, **< .01, ***< .001

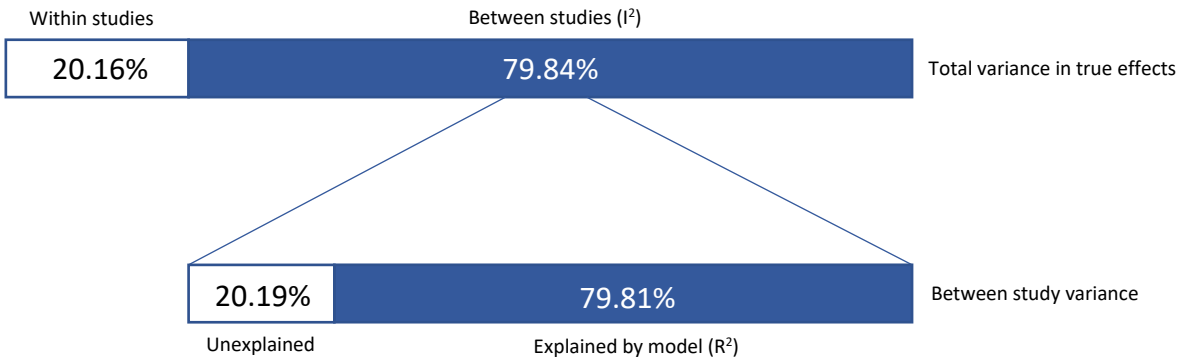


Figure 4.4 *Visual representation of the amount of between-study variance in true effects explained by moderators for the relationship between fat infiltration and ageing*

4.3.5 Risk of Bias across Studies

For studies assessing muscle size, Egger's regression intercept test (z = 1.41, *p* = .16) indicated that publication bias was not present. For studies assessing muscle quality (fat infiltration), visual inspection of the funnel plot (**Figure 4.5**) suggested potential evidence of publication bias, which was consistent with the Egger's regression intercept test (z = -2.03, *p* = .04). Due to the asymmetry detected in the funnel plot, Duval and Tweedie's trim and fill test estimated six studies should be added to the right of the mean, which would yield an adjusted point estimate of r = 0.49 (95% CI: 0.39, 0.58) for the relationship between fat infiltration and ageing.

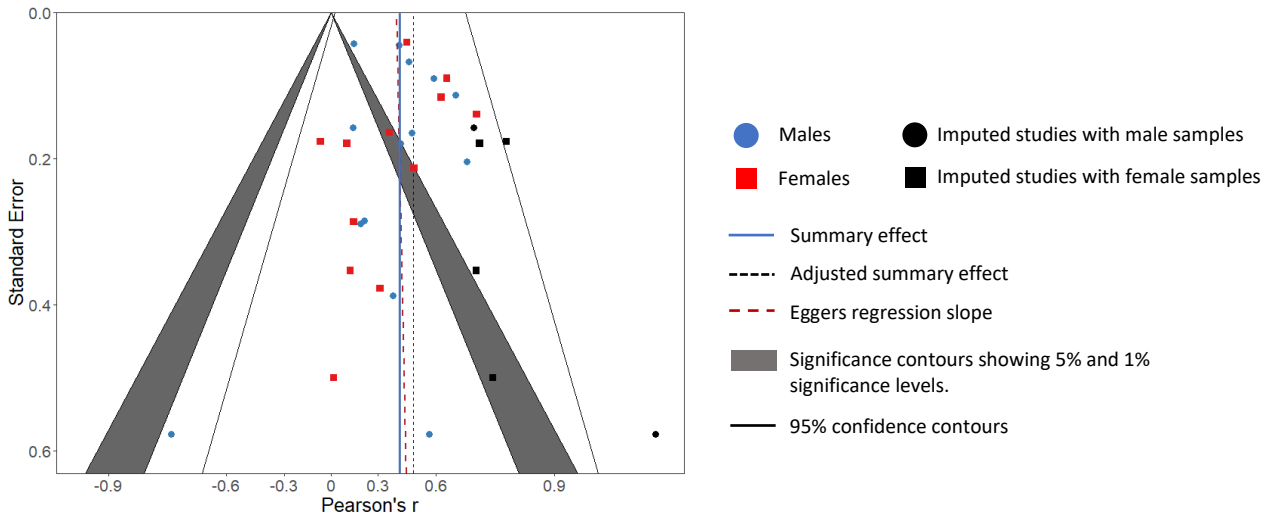


Figure 4.5 *Contour enhanced funnel plot to illustrate potential publication bias for studies assessing muscle quality (fat infiltration)*

4.3.6 Sensitivity Analyses

For the relationship between muscle atrophy and ageing, removing studies that did not explicitly state that their sample were "healthy" and had a BMI outside of 18.5-24.9 yielded a lower overall correlation (r = -0.23, 95% CI: -0.36, -0.10, $p < .001$), no significant moderators and lower between-study variance (I^2 = 67%) when compared to the original three-level meta-analysis. The inclusion of moderators explained 100% of the between-study variance in true effects. Removal of fair and poor-quality studies yielded a greater overall effect (r = -0.35, 95% CI: -0.47, -0.22, $p < .001$); significant moderators included sex ($p < .02$) and mean age of the sample ($p < .05$). Between-study variance was also lower in this model (I^2 = 65%) compared to the original three-level meta-analysis, and the inclusion of moderators resulted in an R^2 value of 77%.

For the relationship between fat infiltration and ageing, removing studies that did not explicitly state that their sample were "healthy" or had a BMI outside of 18.5-24.9 yielded a lower overall correlation (r = 0.32, 95% CI: 0.21, 0.43, $p < .001$). No moderators were significant in this model, although between-study variance was substantially lower (I^2 = 17%) compared to the original three-level meta-analysis. Removal of fair and poor-quality studies slightly lowered the overall correlation (r = 0.39, 95% CI: 0.18, 0.56, $p < .001$). Males ($p < .02$), paraspinal muscles ($p < .01$), low lumbar levels ($p < .05$), ultrasound ($p < .02$), BMI ($p < .01$) and mean age of the sample ($p < .01$) all significantly moderated the relationship between fat infiltration and ageing in the adjusted three-level meta-regression. Muscle ($p < .02$) was also retained as a significant moderator group. Between-study variance was lower in this model (I^2 = 72%) compared to the original three-level meta-analysis. Both sensitivity analyses resulted in 100% of the between-study variance in true effects being explained by the inclusion of moderators.

4.4 Discussion

This is the first study to present a systematic review and meta-analysis on age-related degeneration of the LPMs in healthy adults. Given the inconsistent methods and equivocal nature of findings on this topic, this work provides up-to-date evidence on normal age-related changes in the muscles surrounding the lumbar spine and constitutes an important contribution to the literature base to date. The current findings show that the LPMs undergo degenerative morphological changes as part of healthy ageing in older adults, with increases in fat infiltration more effectual than reductions in muscle size. This suggests that fat infiltration may be a better indicator of age-related decline in the lumbar musculature than muscle atrophy. Indeed, given the predominance of type I fibres in the LPMs (Ng et al., 1998; Mannion et al., 1997; Parkkola et al., 1993; Kimura, 2002; Sirca and Kostevc, 1985) and that type I fibres tend to accumulate fat deposits with age (Gueugneau et al., 2015; Choi et al., 2016) whilst fast-twitch fibres typically exhibit greater atrophy with age (Gueugneau et al., 2015; Novotny, Warren and Hamrick, 2015; Lexell, Taylor and Sjöström, 1988), it is unsurprising that fat infiltration was the more apparent degenerative feature in the lumbar musculature. Although the findings in this review can be explained by established mechanisms that contribute to the development and morphological expressions of age-related sarcopenia (Larsson, Grimby and Karlsson, 1979; Klitgaard et al., 1990; Doherty, 2003; Vettor et al., 2009; von Haehling, Morley and Anker, 2010; Bougea et al., 2016), confidence in the findings is diminished somewhat by the substantial variance between studies. However, disparate methods and population characteristics amongst studies included in this review were able to explain a large proportion of variance and shed light on which factors play a pivotal role in moderating the age-related changes in LPM morphology.

4.4.1 Sex Differences in Muscle Atrophy

The relationship between muscle atrophy and ageing differed significantly between males and females but not for fat infiltration. Males exhibited greater LPM atrophy with ageing compared to females. Males possess greater muscle mass than females, therefore having greater potential for atrophy with age (Janssen et al., 2000). However, this may be overly simplistic and not reflect the complex sex-specific mechanisms that drive decrements in muscle morphology associated with sarcopenia (Payette et al., 2003; Kirchengast and Huber, 2009; Maggio, Lauretani and Ceda, 2013). Lifestyle factors, such as PA, may influence the sex-specific loss of muscle size. Given that PA reduces with ageing equally among men and women (Milanović et al., 2013) and PA has been shown to attenuate the loss of lower limb muscle volume in men but not women (Rivera et al., 2016), it is possible that males also experience greater age-related muscle atrophy in other muscles such as those located in the posterior trunk. However, paravertebral muscle size is relatively independent of PA level (Dasarathy and Merli,

2016; Fortin et al., 2014). A more likely explanation concerns the sex-specific muscle fibre phenotypes of the lumbar musculature. Males possess a greater proportion of type II muscle fibres in the ES than women (Mannion et al., 2000). Whilst type I fibres are more affected by inactivity and denervation-induced atrophy, type II fibres are more susceptible to the effects of ageing (Wang and Pessin, 2013). Therefore, a greater proportion of type II fibres may predispose men to greater paravertebral muscle atrophy. This is reflected in the current findings where men exhibited greater age-related atrophy than women. However, intramuscular fat infiltration does not appear to be a sex-specific degenerative feature of the LPMs.

4.4.2 Muscle-specific Degenerative Responses

Muscle as a group significantly moderated the relationship between fat infiltration and ageing and was approaching significance for moderating the relationship between muscle atrophy and ageing. These findings indicate that there is a muscle-specific response in the lumbar musculature. Therefore, selection of LPMs may be important when evaluating age-related muscle degeneration in the lumbar spine. Atrophy and fat infiltration of the ES and QL showed significantly greater effects with ageing compared to the reference muscle (PS). The correlation between ageing and fat infiltration in the paraspinals was also significantly greater. Indeed, the paraspinals yielded the greatest estimate of fat infiltration amongst the LPMs, but exhibited the least amount of age-related atrophy, albeit without reaching significance. There are perhaps two main reasons to explain these findings. The first concerns how the paraspinal muscles' region of interest (ROI) is defined. Measurements of the paraspinals are sometimes preferred due to the difficulty in discerning the ES and MF muscle boundaries (Lee et al., 2012). The fascial line between the muscles is used to distinguish the medial border of the ES (Crawford et al., 2017). However, this non-muscular tissue, typically included in paraspinal muscle measurements (Ropponen, Videman and Battié, 2008; Lee et al., 2012; Gungor et al., 2015; Schlaeger et al., 2019), may overestimate fat infiltration especially when fat tissue under the lumbosacral plane has been excluded from the ROI (Crawford et al., 2017; Berry et al., 2018). With advancing age, a redistribution of fat and increase in non-contractile tissue between muscles is observed (Addison et al., 2014). Therefore, it is likely that the greater amount of fat infiltration in the paraspinals is an overestimation and in part caused by the inclusion of age-related increases in non-muscle tissue between the MF and ES. This approach may also explain why atrophy is seemingly attenuated in the paraspinals. Increases in non-contractile tissue size between the MF and ES may mask age-related muscular atrophy of the paraspinals. However, fat infiltration has been shown to exceed the loss of lean tissue; indicating that intramuscular adipose tissue does not simply replace the space left by muscle atrophy (Manini et al., 2007). As the paraspinal muscles (ES and MF) are composed mainly of

slow-twitch fibres (JØrgensen, Nicholaisen and Kato, 1993; Rantanen, Rissanen and Kalimo, 1994; Mannion et al., 1997) that are more vulnerable to fat accretion than atrophy (Gueugneau et al., 2015; Choi et al., 2016), this may also explain why paraspinal muscle size is relatively spared in comparison to compositional changes.

The second explanation concerns functional decline with ageing and low PA status in older adults and their effects on muscle morphology. The paraspinal muscles' function, to provide postural support of the lumbar spine and actuate gross trunk movements (Crisco and Panjabi, 1991), may decline with ageing (McGill, Yingling and Peach, 1999; Singh, Bailey and Lee, 2011). PA also significantly decreases in older age (Morse et al., 2004), which results in the accumulation of intramuscular fat (Goodpaster et al., 2008; Marcus et al., 2010; Leskinen et al., 2013). Therefore, diminished age-related muscle function coupled with physical inactivity is likely to result in atrophy and fat accretion in the LPMs (Ikezoe et al., 2012; Teichtahl et al., 2015a). Indeed, skeletal muscle undergoes adaptive reductive remodelling in response to both physical inactivity (Paddon-Jones et al., 2006; Fortney, Schneider and Greenleaf, 2011) and ageing (Rogers and Evans, 1993; Roubenoff and Hughes, 2000; Kalichman, Carmeli and Been, 2017) and given their inter-relationship it is unsurprising that older adults are susceptible to muscle disuse atrophy (Wall, Dirks and Van Loon, 2013). Narici and Maffulli (2010) have suggested that postural muscles are particularly affected by age-related sarcopenia, although this claim warrants further investigation. Deterioration of the paravertebral muscles is likely due to reduced axial loading, as a result of physical inactivity in older age preferentially affecting the antigravity muscles (Ikezoe et al., 2012). Given that paravertebral muscles are predominantly composed of type I muscle fibres (Ng et al., 1998; Mannion et al., 1997; Parkkola et al., 1993; Kimura, 2002; Sirca and Kostevc, 1985) that are suited to prolonged tonic activity (Schiaffino and Reggiani, 2011; Crawford et al., 2016b) and this fibre type is susceptible to inactivity atrophy (Wang and Pessin, 2013), less engagement with PA is a likely mechanism for muscle atrophy in older age.

Muscles such as the paravertebrals may therefore be more vulnerable to degenerative changes in older age. Indeed, the lumbar musculature is more susceptible to progressive fat infiltration with ageing than the leg muscles (Dahlqvist et al., 2015). However, lower limb muscles appear to experience greater atrophy than the back muscles (Abe et al., 2014; LeBlanc et al., 1992). This suggests that the postural function of the LPMs may attenuate the loss of muscle size, although degenerative changes are still apparent in muscle composition. The results of the current review suggest that the ES and QL, which is frequently overlooked, experience the greatest degenerative changes amongst the lumbar musculature with normal ageing. These muscles in particular should be evaluated when determining age-related changes in the lumbar spine. However, researchers should look to include all

of the paravertebral muscles in such evaluations and obtain information on each muscle separately, as degenerative changes are muscle-specific and it is unlikely that any one muscle is representative.

4.4.3 Lumbar Level-dependent Fat Infiltration

The relationship between ageing and fat infiltration in older adults was significantly moderated by the lumbar level at which paravertebral muscles were measured. Studies evaluating muscles across all lumbar levels showed the greatest degenerative changes with ageing. However, the findings in this review indicate that age-related atrophy of the LPMs in healthy older adults is not influenced by the moderating effect of lumbar level. This reveals important methodological considerations as the level of measurement may not be significant for assessing age-related changes in muscle size, but it is of importance for assessing age-related changes in muscle quality. Assessing fat infiltration across all lumbar levels provided the greatest effect size estimate, whilst measurements taken at the high levels (L1-L2) provided the most conservative estimates of muscle quality change (increased fat infiltration) with ageing. Measurements at the mid (L2/3-L3/4) and low (L4-L5/S1) lumbar levels yielded similar small to moderate effect sizes. More importantly, the current findings infer that measurements at the high, mid or low lumbar levels are not representative of the muscle across the whole lumbar region. This finding is supported by the recommendations of Crawford et al. (2017), who suggest that a multi-slice approach across all lumbar levels is superior to determine fat proportion within paravertebral muscle. Although more time-consuming, multi-slice approaches show clear benefits compared to more expedient single slice measurements, primarily as fat infiltration and size measurements at a single slice are not representative of the whole lumbar spine (Urrutia et al., 2018a). Furthermore, volumetric measures are preferable as they are more meaningful functionally (Boom et al., 2008) and potentially minimise errors associated with postural variations during scanning (Meakin et al., 2013).

4.4.4 Influence of Imaging Modality on Muscle Atrophy

Age-related muscle atrophy was significantly influenced by imaging modality, specifically ultrasound. The summary effect for ultrasound studies showed that muscle size increased with age in contrast to the summary effect for studies utilising MRI and CT. This finding suggests that pseudo-hypertrophy is more likely to be reported when using ultrasound to measure LPM size. It also contradicts expectations and raises questions about ultrasound as an accurate imaging modality to measure paravertebral muscle atrophy in generally healthy older adults. Discrepant findings between ultrasound and CT/MRI studies may be due to the evaluation of muscle size when defining the ROI. Typically, MRI and CT evaluations do not consider fat infiltration as part of the muscle, whereas

ultrasound measurements tend to involve the entire muscle including fat. Not considering the amount of fat infiltration within the muscle could mask reductions in muscle size and lead to spurious results (Elliott et al., 2008). Despite its limitations, the inclusion of fat within the ROI may provide a somewhat useful gross measure of muscle degeneration, whilst excluding regions of fat may demonstrate a more specific measure of muscle quality and potentially degenerative features within the muscle boundaries (Berry et al., 2018). Indeed, skeletal muscle measures derived from ultrasound are less able to distinguish intramuscular fat from muscle and accurate definition of the LPMs' boundaries is challenging (Hides, Richardson and Jull, 1995; Pressler et al., 2006; Wallwork et al., 2009). MRI and CT provide high resolution images of soft tissues (Hyun et al., 2016). Compared to ultrasound, the superior soft tissue contrast of MRI/CT, particularly MRI (Hu et al., 2011), is thought to improve the visualisation of fascial boundaries (Upadhyay and Toms, 2015). Furthermore, the generally low resolution of ultrasonic images can make discernment of tissue types difficult (Hides, Richardson and Jull, 1995). This is particularly troublesome when investigating the deep muscles in the pelvis and trunk; sound is reflected or absorbed by superficial tissue layers which results in deeper muscles lacking sufficient resolution (Pillen, 2010).

Another limitation associated with ultrasound concerns the operator's ability to standardise pressure applied by the transducer to the scan site (Lukaski, 1987). Muscle thickness, as well as subcutaneous adipose tissue, may be affected by excessive pressure (Abe et al., 1994). Therefore, avoiding excessive pressure whilst following a strict imaging protocol is paramount to achieving more accurate measures of muscle morphology when using ultrasound (Dupont et al., 2001). Finally, ultrasound typically has a limited field of view (FOV) (Sions, Teyhen and Hicks, 2017), unlike MRI and CT which are capable of imaging the entire lumbar musculature whilst retaining sufficient resolution. Increasing the FOV to capture more of the lumbar musculature may compromise image quality for ultrasound, compounding the limitations stated above. Although imaging modality did not significantly influence the relationship between fat infiltration and ageing, ultrasound again exhibited marked differences with CT and MRI. Therefore, overestimation of muscle size was most likely due to the inclusion of non-contractile tissue (Sions, Teyhen and Hicks, 2017), whilst the underestimation of fat infiltration was likely a consequence of echo intensity diminishing in deeper muscles of the trunk (Pillen, 2010). Despite ultrasound being acknowledged as a lower cost and portable alternative to assess skeletal muscle morphology in clinical and community settings (Stringer and Wilson, 2018; Mourtzakis and Wischmeyer, 2014), the current findings indicate that studies should ideally use MRI or CT to evaluate age-related atrophy in the LPMs. However, recent advances in ultrasound technology indicate that this modality may be more clinically relevant and applicable in research settings going forward (Romero-Morales et al., 2021). A recent study has shown that panoramic ultrasound imaging is a valid tool for monitoring muscle atrophy in

the lower limbs (Scott et al., 2017). Furthermore, high spatial resolution and image quality is now achievable using ultra-high frequency ultrasonography, which has the potential to generate novel and innovative uses in musculoskeletal imaging (Izzetti et al., 2020).

4.4.5 Influence of BMI, Mean Age and Age Range on Muscle Degeneration

The continuous covariates did not have a moderating effect on the relationship between ageing and muscle atrophy. However, age range of the sample significantly influenced the relationship between fat infiltration and ageing, and mean BMI of the sample was approaching significance. It seems intuitive that an increase in BMI would increase the amount of fat infiltration in the LPMs with ageing. Since increases in BMI are largely attributed to increases in whole-body adiposity (Gallagher et al., 1996), it is likely that the amount of fat infiltrating the paravertebral muscles would also increase. Increasing age range also increased the effect of fat infiltration with ageing. Simply put, as age range increases for a population of healthy older adults, greater degenerative changes in the paravertebral muscles can be observed. This is reflected in longitudinal observations as small time periods (e.g. 12 months) are likely to highlight only modest age-related changes in muscle morphology (Gibbons et al., 1997; Ikezoe et al., 2015), whereas longer periods (e.g. 15 years) have the potential to exhibit greater changes (Fortin et al., 2014), specifically in fat infiltration.

4.4.6 Sensitivity Analyses

The sensitivity analyses showed that older adults, who are explicitly defined as healthy with a normal BMI, undergo less muscular degeneration in the lumbar region with normal ageing. Furthermore, all moderators were non-significant, suggesting that study level covariates are unable to moderate the relationship between ageing and LPM degeneration in this population. Although samples included from the eligible studies were generally healthy, free from disease and without physical limitations, only 51% of these studies explicitly stated that their sample were healthy in the article. Based on information within articles and correspondence from authors, it is unlikely that the health status of samples between studies differed greatly. However, the sensitivity analyses suggest that older adult participants selected for health (i.e. explicitly defined as healthy with a normal BMI), exhibit less degeneration within the lumbar musculature. Similar discrepancies are seen in the degeneration of the LPMs between healthy and diseased populations (Kalichman, Carmeli and Been, 2017). Although the samples included in the current review were not from diseased populations, the subtle differences in the definition of health status had a clear influence on age-related muscle atrophy and fat infiltration. Removal of ‘poor’ and ‘fair’ quality studies showed that greater atrophy was apparent with

ageing. This suggests that good quality studies, most likely through better outcome measurement, are able to detect greater changes in paravertebral muscle size. Caution should be taken with this interpretation due to substantial between-study heterogeneity (I^2 = 72%), although differences in methodologies and study population characteristics can explain all of the variance (R^2 = 100%).

4.4.7 Clinical and Practical Applications

Measurement of muscle morphology is performed as part of sarcopenia diagnostic criteria (Studenski et al., 2014; Chen, L. K. et al., 2014; Fielding et al., 2011; Cruz-Jentoft et al., 2019). Whilst appendicular skeletal muscle mass is typically measured (Correa-de-Araujo, 2017; Tosato et al., 2017), lumbar (L3) muscle CSA derived from CT or MRI offers a promising alternative (Gu et al., 2018; Schweitzer et al., 2015; Shen et al., 2004; Golse et al., 2017). However, the current results indicate that measurements derived from a single slice are not representative of the entire lumbar musculature. Volumetric measures across the lumbar are recommended; however, time costs involved in such an approach may not be suited to clinical settings. The choice of muscles should also be considered when investigating changes in lumbar muscle morphology. Whilst analysing each muscle in the lumbar spine would provide the most comprehensive assessment, the ES and QL should be included in measurements as they show significant atrophy and fat infiltration with ageing. Based on the results of this review, MRI and CT are recommended over the use of ultrasound to measure changes in muscle quality and size with ageing. Indeed, MRI and CT are considered gold standard modalities for non-invasive assessment of muscle size (Cesari et al., 2012; Olsen, Qi and Park, 2005). However, their use is limited in primary care settings by their availability, costs, radiation dosage (CT), inapplicability to persons with older generation implanted medical devices that are not MRI compatible (i.e. ferromagnetic), and requirement for highly specialised operators (Beaudart et al., 2016; Correa-de-Araujo, 2017). Despite these barriers, the use of high resolution imaging modalities to assess muscle degeneration is expected to become more commonplace in clinical practice (Cruz-Jentoft et al., 2019). Perhaps the greatest advantage of high-resolution imaging modalities is their ability to provide accurate estimates of muscle quality (McGregor, Cameron-Smith and Poppitt, 2014). However, this review found numerous measures used in the literature, which makes transference into clinical practice difficult due to a lack of consensus.

The importance of muscle fat composition as a key determinant of muscle function is being increasingly recognised and changes in muscle quality may precede those in muscle size with ageing (McGregor, Cameron-Smith and Poppitt, 2014; Correa-de-Araujo et al., 2017; Shahidi et al., 2017b; Anderson et al., 2016). Therefore, specific measures of fat infiltration to estimate changes in paravertebral muscle quality may be particularly useful in clinical settings. The findings of this study suggest that any fat infiltration measures derived from high resolution imaging modalities are suitable,

although further research is needed to determine the optimal approach for future research and clinical applications. Applying the results from the regression analyses in this review to clinical practice could add to the current clinical perspective. As a measure of magnitude for age-related degeneration in the LPMs, correlation coefficients obtained from clinical assessment data can be compared to the current findings, which may enable identification of abnormal degenerative changes with ageing. However, caution should be exercised due to the wide confidence intervals.

4.4.8 Limitations

Although this review was systematically rigorous, there were limitations that should be acknowledged. Firstly, data were collated from observational studies and baseline evaluations from experimental studies. Despite being the best available source, observational studies are considered to produce lower quality evidence than experimental studies and include a greater potential risk of bias. Furthermore, all data were collated from cross-sectional observations, making it difficult to ascertain the exact nature of age-related changes in muscle morphology. There is a need for more longitudinal studies directly investigating normative changes in lumbar paravertebral muscle morphology over longer time periods (>10 years).

Ill-defined and inconsistent definitions of health status also limited the ability to compare studies. Although 18 studies (51%) explicitly stated that their sample were healthy, many of these studies provided insufficient detail on what constituted as 'healthy'. Furthermore, whilst some studies considered matched controls representative of healthy individuals, caution should be taken with this approach as undetermined phenotypes are likely hidden in the demographics (Määttä et al., 2015). Insufficient selection of participants based on their health status and lack of reporting clarity were substantial limitations. A standardised definition should be adopted to allow comparison between healthy populations as well as with diseased populations. Such advances would provide a reference to facilitate understanding of spinal disease progression and pain-related expressions of muscle degeneration. Despite large sample sizes present in some studies, it is likely that many of the included data were not sufficiently powered to detect meaningful changes to muscle morphology with ageing; only six studies (17%) provided sample size justifications.

It should be recognised that the English language restriction may have also limited the number of articles that were returned, although it was unlikely to result in systematic bias (Morrison et al., 2012). Data regarding PA were scarce from the studies included in this review. Although exercise is known to affect muscle morphology (Konopka et al., 2018; Ikenaga et al., 2017; Belavý, Gast and Felsenberg, 2017; Stec et al., 2017; Janssen et al., 2016; Manini et al., 2007), changes in paravertebral muscle

morphology are relatively independent of PA (Fortin et al., 2014; Dahlqvist et al., 2015; Dasarathy and Merli, 2016). Future studies should consider PA level when evaluating age-related degeneration of the lumbar musculature.

4.5 Conclusion

This systematic review, for the first time, draws together the extant literature relating to age-related changes in the lumbar musculature. The findings are based on older adults free of diseases or impairments that likely affect paravertebral muscle morphology. This is a necessary first step in furthering our understanding of normative expressions of ageing muscle as well as providing recommendations to establish continuity amongst protocols in future studies. The findings in this review indicate that the paravertebral muscles undergo degenerative changes (atrophy and fat infiltration) with normal ageing. Future studies investigating muscle morphology in the lumbar spine should consider the sex and age range of their sample, look to use MRI/CT to image the paravertebral muscles and analyse all the individual muscles across the entire lumbar region. However, these methodological decisions should not be uniform, rather based on the morphological outcome of interest. In summary, this review will provide a reference for normal age-related changes observed in LPM morphology, which may enable identification of pathological deviations. Furthermore, the practical applications of this meta-analysis will provide guidance to future studies investigating age-related degeneration in the lumbar musculature.

Table 4.9 *Thesis Map*

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance

	to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men		Aim	
		Key findings	
		Implications	
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	

Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men

The work from this chapter has been accepted for publication in a peer-reviewed journal.

Dallaway, A., Hattersley, J., Diokno, M., Tallis, J., Renshaw, D., Wilson, A., Wayte, S., Weedall, A., and Duncan, M. (in press) 'Age-related Degeneration of Lumbar Muscle Morphology in Healthy Younger versus Older Men'. *The Aging Male*

Chapter Abstract

Background The lumbar paravertebral muscles are important in maintaining health and mobility in older age. Despite this, no studies have quantified age-related differences in muscle volume and fat infiltration for all of the main paravertebral muscles. Given the difference in their functional roles, investigating morphological changes with age are of interest for each of these muscles. Therefore, the main aim of this study was to evaluate age-related differences in lumbar paravertebral muscle morphology in healthy young and older adult men.

Methods T2-weighted axial MRI of the lumbar spine were obtained for twelve healthy older (67.3 ± 6.0 years) and young (24.7 ± 3.1 years) men. Normalised muscle volume (NMV) and muscle-fat-infiltrate (MFI) were determined bilaterally for the psoas (PS), quadratus lumborum (QL), erector spinae (ES) and multifidus (MF). MANOVA was used to compare NMV and MFI between age groups. Follow-up ANOVA compared NMV and MFI for each muscle between age groups, with PA as a covariate. Stepwise regression was used to explore the association between muscle morphology.

Results NMV of the ES and QL were significantly lower in the OG ($p = .040$ and $p < .001$, respectively). MFI across all muscles was significantly greater in the OG ($p<.001$). PA did not moderate the relationship between age and muscle degeneration. Non-dominant handgrip strength was associated with NMV ($p = .003$).

Conclusions Age-related atrophy is muscle-specific in the lumbar spine; changes in lumbar musculature is independent of PA, handgrip strength may reflect morphological changes in the postural muscles with age. This study supports establishing effective targeted exercise interventions in the lumbar musculature

Key Words: paravertebral muscles, MRI, segmentation, morphology, fat infiltration, atrophy

Table 5.1 Thesis Map

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance
		Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle	• Studies investigating muscle degeneration with ageing have typically focused on the appendicular muscles	Aims	• To investigate age-related differences in LPM morphology • A secondary aim was to investigate the age-response on fat infiltration and volume of the different lumbar

Morphology in Healthy Younger versus Older Men	• There is increasing recognition for the importance of the lumbar paravertebral muscles in maintaining health and mobility in older age • Few studies have characterised features of age-related degeneration in the lumbar musculature • Few studies have provided volumetric information on all of the paravertebral muscles using high-resolution imaging modalities		muscles (i.e. multifidus, erector spinae, quadratus lumborum and psoas) • An additional aim was to explore other predictors of lumbar paravertebral muscle degeneration
		Key findings	
		Implications	
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men		Aim	
		Key findings	
		Implications	
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	

denotes links to previous chapters. Links to chapter 3 – 1) VPA was included as a potential covariate; 2) The moderating effects of body composition measures and handgrip strength were explored with respect to muscle morphology degeneration. Links to chapter 4 – Methodological decisions for imaging the LPMs were based on the findings from the meta-analysis

CHAPTER 5

5.1 Introduction

Sarcopenia is a major health concern (Marcus et al., 2010; Landi et al., 2013; Beaudart et al., 2017) and socioeconomic burden, responsible for considerable healthcare expenditure in the United Kingdom (Pinedo-Villanueva et al., 2019) and United States (Janssen et al., 2004). With worldwide increases in the number of older adults, the challenges posed by sarcopenia are increasingly great at patient, societal and clinical levels (Dodds and Sayer, 2016; Beaudart et al., 2014).

Diagnostic criteria for sarcopenia typically include measurement of appendicular muscle mass (Cruz-Jentoft et al., 2019; Chen, L. K. et al., 2014; Fielding et al., 2011; Studenski et al., 2014; Morley et al., 2011; Muscaritoli et al., 2010), however, there is increasing evidence highlighting the value of measuring paravertebral muscle degeneration (atrophy and fat infiltration) (Dahlqvist et al., 2017; Hicks et al., 2005a; Crawford et al., 2019b; Shahtahmassebi et al., 2017; Sions et al., 2017a; Fortin et al., 2015). Narici and Maffulli (2010) suggest that the postural muscles may be more susceptible to the effects of age-related sarcopenia than the appendicular muscles. This suggestion is supported as the lumbar musculature is more susceptible to progressive fat infiltration with ageing than the lower limbs (Dahlqvist et al., 2015). Degeneration of the lumbar musculature has attracted interest in recent years, even stimulating ideas of spinal sarcopenia (Debiane et al., 2015; Kim et al., 2019). This focus is likely due to the importance of the paravertebral muscles in the maintenance of spinal health (Eguchi et al., 2017; Hicks et al., 2005b; Sions et al., 2017a), postural support, falls prevention, and assisting with trunk movements during ADLs (Suri et al., 2009; Granacher et al., 2013; Hicks et al., 2005a; Barr, Griggs and Cadby, 2005; Meakin et al., 2013; Crisco and Panjabi, 1991). Therefore, degradation of these muscles may be particularly detrimental to biomechanical and physical function, evidenced by adverse health outcomes in older age (Kita et al., 2013; Hicks et al., 2005b; Katzman et al., 2012; Williams et al., 2017).

Studies have shown that lower back pain and pathology modifies the size and composition of the LPMs (Beneck and Kulig, 2012; Danneels et al., 2000; Chen, Y. Y. et al., 2014; Kalichman et al., 2010, 2016; Teichtahl et al., 2015b). However, the extent of muscle atrophy and fat infiltration is confounded by physiological declines associated with normal ageing (Hicks et al., 2005a; Kalichman et al., 2010; Le Cara et al., 2014; Shahidi et al., 2017b; Valentin, Licka and Elliott, 2015; Hebert et al., 2014). Few studies have directly investigated the effects of healthy ageing on muscle size and fat infiltration in the lumbar spine (Valentin, Licka and Elliott, 2015; Crawford et al., 2016a; Fortin et al., 2014). Furthermore, the range of approaches used to evaluate age-related changes in LPM morphology makes comparing findings difficult (see **36)**.

Attenuating atrophy and fat infiltration in the LPMs is important to maintain quality of life and offset adverse health outcomes in older age (Kader, Wardlaw and Smith, 2000; Ekin, Yıldız and Mutlu, 2016; Ikezoe et al., 2015; Fortin et al., 2015; Kalichman, Carmeli and Been, 2017; Anderson et al., 2016). It is widely acknowledged that further investigation is needed to extend our understanding of age-related degeneration in the lumbar musculature (Crawford et al., 2016c; Kalichman, Carmeli and Been, 2017; Dahlqvist et al., 2015; Valentin, Licka and Elliott, 2015; Shahidi et al., 2017b), particularly in healthy volunteers as undetermined phenotypes are likely hidden in the demographics of general populations (Määttä et al., 2015; Crawford et al., 2019b). To the author's knowledge, no study to date has included volumetric and fat infiltration measures for the PS, QL, ES and MF in relation to healthy ageing. Investigating age-related differences in muscle volume and fat infiltration in the lumbar spine of healthy adults will provide further evidence for future comparative studies to identify pathological deviations. Furthermore, understanding this information may elucidate mechanisms related to functional decline and provide much needed evidence for effective targeted interventions in the lumbar spine.

5.1.1 Age-related Differences in Lumbar Paravertebral Muscle Morphology

Different outcome measures amongst studies has hampered our understanding of muscle atrophy in the lumbar spine. Factors such as the inclusion or exclusion of non-contractile tissue within the muscle's ROI has also contributed to discrepant findings. Studies accounting for non-contractile tissue have shown that the LPMs atrophy with ageing (Valentin, Licka and Elliott, 2015; Crawford et al., 2016a; Fortin et al., 2014), contrary to studies measuring total muscle CSA without consideration of fat infiltration (D'Hooge et al., 2012). Discounting fat infiltration may mask atrophic changes resulting in an apparent preservation of muscle size with ageing (Elliott et al., 2008). In **36** it was shown that men experience LPM atrophy. However, many of the studies included in the analysis used cross-sectional measures and sampling variance was high. Furthermore, no study included all of the main LPMs and definitions of health were inconsistent. It is therefore difficult to assert with any degree of confidence that healthy men undergo atrophic changes in the lumbar musculature. A study addressing these issues is needed to establish the extent of LPM atrophy in healthy older men.

Fat infiltration within skeletal muscle (myosteatosis) is a degenerative feature signalling a decline in muscle structure and quality (Mitchell et al., 2012; Elliott et al., 2013). Indeed, strength decrements often exceed the loss of muscle mass (Delmonico et al., 2009; Hughes et al., 2002; Narici and Maffulli, 2010), which suggests that intrinsic factors contribute to this decline. Alongside a shift towards a slower phenotype, due to a preferential loss of fast motor units (Campbell, McComas and Petito, 1973;

Evans and Lexell, 1995), fat deposits infiltrate skeletal muscle with ageing (Marcus et al., 2010). Rather than being characterised as an atrophic change however, intramuscular fat infiltration is seen as a feature of structural remodelling (Hodges et al., 2015). Studies have shown that increased fat infiltration in the lumbar musculature is associated with reduced muscle strength (Goodpaster et al., 2001), poor physical function (Hicks et al., 2005a) and impaired mobility (Goodpaster et al., 2006), as well as degenerative spinal features (Kalichman et al., 2010; Teichtahl et al., 2016, 2015b). Whilst methodological differences, inconsistent results (Fortin and Macedo, 2013; Fortin et al., 2014; Fortin, Yuan and Battié, 2013) and disease (Kalichman et al., 2010; Takayama et al., 2016; Lorbergs et al., 2019; Shahidi et al., 2017a; Masaki et al., 2016; Bayat et al., 2019) preclude any conclusions on compositional changes within the lumbar musculature from being drawn, the evidence suggests that fat infiltration in the LPMs is a normal feature of ageing (Shahidi et al., 2017b; Crawford et al., 2016a; Valentin, Licka and Elliott, 2015; Lee et al., 2017).

The LPMs may be particularly susceptible to fat infiltration with advancing age. Given that these muscles are mainly composed of type I fibres (Ng et al., 1998; Mannion et al., 1997; Parkkola et al., 1993; Kimura, 2002; Sirca and Kostevc, 1985) which have a propensity to accumulate more intramyocellular lipid with ageing than type II fibres (Gueugneau et al., 2015; Choi et al., 2016), it is unsurprising that fat infiltration is particularly apparent in the lumbar musculature. Furthermore, changes in muscle fat content have been suggested to precede changes in muscle size with ageing (Ismail et al., 2015; Watanabe et al., 2013; Goodpaster et al., 2006; McGregor, Cameron-Smith and Poppitt, 2014). This highlights the importance of fat infiltration measures as an early indicator of degenerative muscle in older age. Such measures, sensitive to small changes in muscle morphology, may enable preventative interventions to be implemented earlier and maintain healthy muscle in older adult populations (McGregor, Cameron-Smith and Poppitt, 2014).

5.1.1.1 Muscle-specific Changes in Paravertebral Muscle Morphology

The LPMs most frequently measured in the literature are the MF, ES, QL and PS (**Figure 5.1**). Given their different functions and propensity for localised degeneration in healthy and diseased populations (Ploumis et al., 2011; Min et al., 2013; Crawford et al., 2016c; Baracos, 2017; Sollmann et al., 2020), age-related differences are of interest for each of these muscles. Of the LPMs, the MF (Crawford, Elliott and Volken, 2017; Crawford et al., 2016c, 2016a; Marshall et al., 2011; Valentin, Licka and Elliott, 2015) and ES (Crawford, Elliott and Volken, 2017; Fortin et al., 2014; Lee et al., 2017; Fortin et al., 2016; Anderson et al., 2013; Valentin, Licka and Elliott, 2015) appear most vulnerable to age-related degeneration. However, it should be noted that fat infiltration, rather than atrophy, appears

to be the main degenerative feature of the MF (Valentin, Licka and Elliott, 2015; Crawford et al., 2016a). Other studies have shown that the QL exhibits the greatest decrements with ageing (Sions et al., 2017b; Johannesdottir et al., 2018), whilst the size and quality of the PS seems to be somewhat spared (Crawford, Elliott and Volken, 2017; Lee et al., 2017). This is possibly due to its involvement in hip flexion and the fact that adults rely more on hip power for walking in older age (DeVita and Hortobagyi, 2000; Neptune, Zajac and Kautz, 2004). It is not fully understood which mechanisms increase or attenuate the effects of ageing on degenerative muscle morphology in the lumbar spine. Indeed, the lumbar musculature is more susceptible to progressive fat infiltration with ageing than the lower limbs (Dahlqvist et al., 2015). Although, lower limb muscles appear to undergo greater atrophic changes than the back muscles (Abe et al., 2014; LeBlanc et al., 1992). This suggests that the postural function of the LPMs may attenuate the loss of muscle size, although degenerative changes are realised in muscle composition.

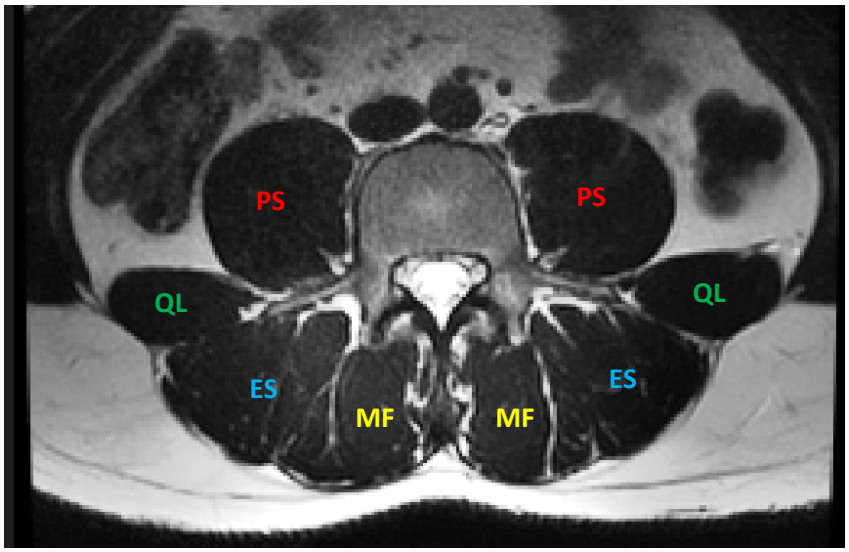


Figure 5.1 *Axial MRI image depicting the psoas (PS), quadratus lumborum (QL), erector spinae (ES) and multifidus (MF) muscles.*

5.1.1.2 Moderating Effect of Physical Activity and Lumbar Curvature

Lifestyle factors, such as PA, may also modify the relationship between ageing and muscle degeneration. Inactivity is a well-known factor that contributes to sarcopenia (Steffl et al., 2017), due to alterations in the rates of protein synthesis and degradation (Evans, 2010). Paravertebral muscle size has been associated with PA levels (Peltonen et al., 1997; Gibbons et al., 1998), although not unequivocally (Teichtahl et al., 2015a). Indeed, some studies report that changes in LPM morphology are relatively independent of PA (Fortin et al., 2014; Dahlqvist et al., 2015; Dasarathy and Merli, 2016).

In the author's view, PA level has been generally overlooked as a potential covariate in the literature. Anatomical variations in lumbar sagittal curvature (Meakin et al., 2013) may also influence age-related changes in LPM morphology and their moderating effect should be considered. Understanding which muscles are most affected by the deleterious effects of ageing and the influence of PA and anatomical variations, may provide better guidance in identifying where exercise programmes and resistance-based training interventions are best targeted.

5.1.2 Methodological Considerations

In the previous **chapter**, it was suggested that the evaluation of LPM morphology requires specialist imaging tools such as MRI or CT. Although US has been frequently used to assess muscle morphology in the lumbar spine (Stokes, Rankin and Newham, 2005; Watson, McPherson and Starr, 2008; Ikezoe et al., 2012, 2015; Masaki et al., 2016; Frost and Brown, 2016; Aboufazeli et al., 2018; Rahmani et al., 2019; Shahtahmassebi et al., 2017; Shadani et al., 2018; Hides, Richardson and Jull, 1995), its limitations may cause erroneous measurements. Compared to MRI/CT, ultrasonic measures are less able to distinguish intramuscular fat from muscle tissue which also makes defining boundaries of individual LPMs challenging (Hides, Richardson and Jull, 1995; Pressler et al., 2006; Wallwork et al., 2009). MRI is the gold standard for examining the paravertebral musculature (Crawford et al., 2019a). The superior soft tissue contrast and high resolution images of MRI/CT, particularly MRI (Hyun et al., 2016; Hu et al., 2011; Crawford et al., 2019a) improves the visualisation of fascial boundaries (Upadhyay and Toms, 2015), which may increase accuracy when identifying deep muscles in the lumbar spine. Methods for identifying muscle ROI with MRI may still vary and influence measures of paravertebral muscle degeneration. Inclusion of fat between the iliocostalis lumborum and longissimus thoracis and lateral to the iliocostalis, interposed between the epimyseal border and the fascial plane, may increase estimates for muscle size and fat infiltration (Berry et al., 2018). However, the decision to include this fatty ROI does not render muscle size and fat infiltration measures useless. Rather, such measures are indicative of a gross measure of muscle degeneration whereas excluding this fatty ROI may provide more specific measures of degeneration (Berry et al., 2018).

Another factor that may cause disparity amongst studies is the choice of MRI sequence. T1- and T2-weighted images are widely used to analyse soft tissue morphology in the lumbar spine (Fortin et al., 2014; Hebert et al., 2014; Shahidi et al., 2017a, 2017b; Beneck and Kulig, 2012; D'Hooge et al., 2012; Valentin, Licka and Elliott, 2015; Gibbons et al., 1997). Quantification techniques are achieved by distinguishing regions (pixels / voxels) of fat from contractile tissue within a selected muscle's ROI. Quantitative techniques are more accurate and have shown greater reliability than qualitative grading

(Abbott et al., 2018) and semi-quantitative methods (Mhuiris et al., 2016). However, accurately defining muscles' boundaries is paramount to achieving high accuracy. High contrast between contractile and non-contractile tissue in T2-weighted images (Bloem et al., 2018) is thought to improve visualisation of muscle boundaries (Upadhyay and Toms, 2015). This may increase accuracy when identifying a muscle and defining its ROI. Visualisation of fat infiltration within the muscle is also improved, although there is no consensus regarding the choice of T1 or T2 sequences (Upadhyay and Toms, 2015). Furthermore, T1- and T2-weighted images may be too simplified to precisely discriminate different structures of similar signal intensity (Bloem et al., 2018). Multi-echo acquisitions have been recently used to determine fat infiltration within the LPMs (Crawford, Elliott and Volken, 2017; Crawford et al., 2019b). Such techniques are purported to be superior in soft tissue analysis (Fischer et al., 2013; Reeder, Hu and Sirlin, 2012; Yoo et al., 2015; Ma et al., 2004), although errors may still occur due to field inhomogeneities (Crawford et al., 2017).

Finally, it is important to consider the advantages and disadvantages of volumetric versus cross-sectional measures. Assessing muscle volume, rather than CSA or thickness, is advantageous due to its greater association with muscle function and strength (Akagi et al., 2009; Blazevich et al., 2009; Boom et al., 2008), minimisation of errors associated with postural variations during scanning (Meakin et al., 2013) and more accurate representation of the entire lumbar musculature (Urrutia et al., 2018a). Despite the benefits of volumetric measures, CSA measurements remain more commonplace in the literature, likely due to expediency (Urrutia et al., 2018a) and being representative of a whole muscle in estimating volume and fat infiltration (Hogrel et al., 2015). However, this has been contested by other researchers focusing on the lumbar musculature (Urrutia et al., 2018b). Crawford et al. (2017) suggest that a multi-slice approach across all lumbar levels is superior to determine fat proportion within paravertebral muscle. Furthermore, contraction of a muscle will increase its CSA, whilst passive elongation will decrease its CSA. These factors make CSA measures of the spine particularly susceptible to postural changes, particularly as the spine has a large degree of flexibility and trunk flexion results in decreased CSA measurement of the extensor muscles (Jorgensen, Marras and Gupta, 2003). Measuring volume may minimise errors associated with passive elongation and active contraction as muscle tissue is generally considered to be incompressible (Ehret, Böl and Itskov, 2011). Therefore, changes in muscle length and CSA would not be expected to alter volume (Barber, Barrett and Lichtwark, 2009).

5.1.3 Aims, Objectives and Hypotheses

The aim of this study was to investigate age-related differences in LPM morphology. A secondary aim was to investigate the age-response on different muscles in the lumbar spine. An additional aim was

to investigate the association between exploratory variables and age-related LPM degeneration. To achieve these aims, specific objectives were to:

Table 5.2 *Objectives and hypotheses for chapter 5*

	Objective	Null Hypothesis
1	Determine an appropriate imaging modality and image processing technique to quantify muscle volume and fat infiltration in the lumbar spine	n/a
2	Accurately and reliably perform manual segmentation of the LPMs	n/a
3	Calculate volumetric and fat infiltration data for each LPM in the YG and OG	Age-related differences in LPM morphology will not be muscle specific
4	Analyse volumetric and fat infiltration data using appropriate statistical tests to allow inference of age-related differences	a) Normalised volume of the lumbar spine muscles will not be significantly less in the OG compared to the YG b) Fat infiltration in the lumbar spine muscles will not be significantly greater in the OG compared to the YG
5	Control for the moderating effect of VPA	VPA will not moderate the relationship between muscle degeneration and age
6	Use regression modelling to explore the association between age-related degeneration and exploratory factors (e.g. PA level, whole-body composition, handgrip strength and anatomical variations in lumbar curvature)	PA, BMI, whole-body fat and lean mass, handgrip strength and lumbar lordotic angle will not significantly associate with age-related loss of muscle volume or increase in fat infiltration

5.2 Methods

5.2.1 Imaging Acquisition

Scans of the lumbosacral spine were performed in a 3T MR imaging scanner (Discovery MR750w, GE Medical Systems, Milwaukee, Wisconsin, USA). Participants were positioned supine in the magnetic bore with a pillow placed under their legs resulting in slight flexion of the hips and knees. This position was assumed as lying supine with a neutral spine has been shown to provide the most accurate measure of paraspinal muscle anatomical CSA (Jorgensen, Marras and Gupta, 2003). A flexible 16-element body-matrix coil (GEM Anterior Array, GE Healthcare, Waukesha, Wisconsin, USA) was used in combination with an in-table GEM Posterior Array (GE Healthcare, Waukesha, Wisconsin, USA) consisting of a 5 x 8 array to improve signal reception. Axial T2-weighted fast recovery fast spin-echo (FRFSE) images were acquired from the L2 inferior endplate to the L5 inferior endplate, using a slice thickness of 4 mm, no interslice gap, repetition time (TR) 6643 ms, echo time (TE) 107 ms, acquisition matrix 240x240, flip angle 150°, field of view (FOV) 240 mm, voxel size 0.938 x 0.938 x 4 mm, 30 slices provided sufficient coverage. Images were stored as DICOM format for processing.

5.2.2 Image Analysis

Image analysis was performed using ITK-SNAP (ITK-SNAP, version 3.8.0, www.itk-snap.org) (Yushkevich et al., 2006), a general-purpose interactive tool for image visualisation and segmentation. Right and left sides of the PS, QL, ES and MF were manually segmented for each axial slice between the superior endplate of L3 to the superior endplate of L4 (**Figure 5.2**). The superior endplate of L4 was chosen as the inferior-most level to avoid obliquity at lower levels. Age-related differences in lumbar lordotic angle (LLA) (Arshad et al., 2019), and steep angulation of the L5 vertebra (Keller et al., 1999), would likely confound within and between-group comparisons. Therefore, the chosen levels provided identifiable anatomical planes that were approximately parallel to the axial slices in all participants, minimising inter-subject measurement error. Each muscle's boundary was identified following the instructions of Crawford et al. (2017). When a large fat-filled "tent" was observed between the longissimus thoracis and iliocostalis lumborum, this region was excluded from the ROI. Fatty regions lateral to the illiocostalis lumborum and beneath the fascial plane were also excluded.

5.2.2.1 Muscle Volume

The summation of axial ROIs provided volumetric measurements for each muscle. Volumetric measures are preferable to CSAs as they are more meaningful functionally (Akagi et al., 2009; Blazevich et al., 2009; Boom et al., 2008) and minimise errors associated with postural variations

during scanning (Meakin et al., 2013). Due to variations in participants' anatomy, the number of analysed slices ranged from 10 to 11. To account for differences in muscle volume as a result of stature, muscle volumes were normalised to the straight-line distance between the anterior superior border of the L1 vertebra and the anterior superior border of the S1 vertebra (Deng et al., 2015) (**Figure 5.3**) giving normalized muscle volume (NMV) in arbitrary units (a.u.). Normalisation to vertebral body CSA (Thakar et al., 2016) and stature (Hamaguchi et al., 2016) has also been performed to account for differences in body size. However, the current approach was adopted as vertebral column length is moderately to strongly associated with stature (Nagesh and Pradeep Kumar, 2006) and provides reliable normalisation while negating postural variations (Voss et al., 1990).

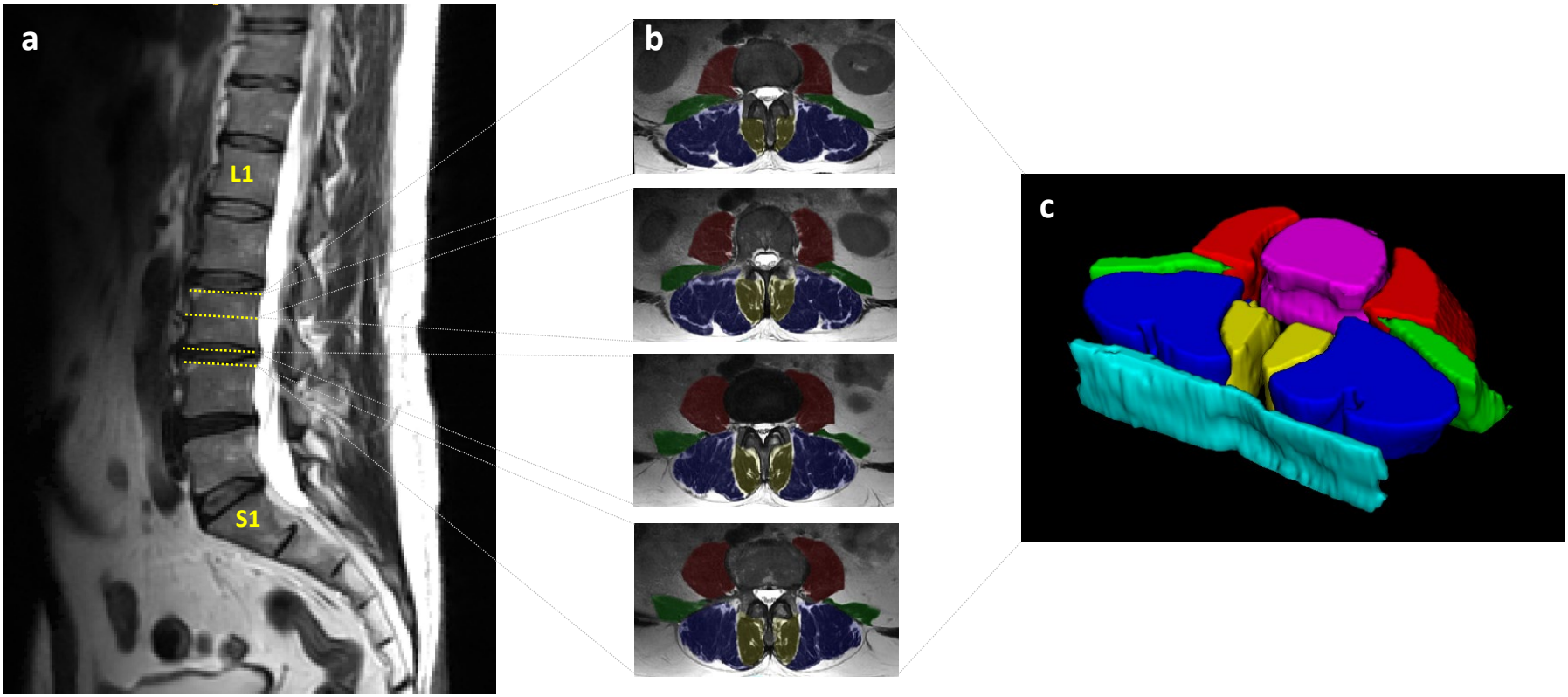


Figure 5.2 *Sagittal (a) and selective axial MRI images (b) of the spine showing the ROI for the psoas (red), quadratus lumborum (green), erector spinae (blue) and multifidus (yellow) muscles at the superior endplate L3, mid-vertebral slice L3, intervertebral disc L3/L4 and superior endplate L4. 3-D rendering of the axial image segmentations (c) is show with additional region of subcutaneous back fat (turquoise) and vertebral column (pink) for visualisation.*

5.2.2.2 Fat Infiltration

MFI was used to estimate and compare intramuscular fat infiltration between participants. For MFI, mean signal intensity (MSI) of each muscle across all included slices was reported as a percentage relative to MSI of a homogenous region of subcutaneous back fat across all included slices, given by the equation:

$$MFI\ (\%) = \frac{Muscle\ MSI}{Subcutaneous\ Fat\ MSI} \times 100$$

Similar approaches have been used previously (D'Hooge et al., 2012; Hyun et al., 2016; Valentin, Licka and Elliott, 2015; Gibbons et al., 1997) to account for inter-subject and temporal variations in measured signal intensity due to field strength variations and distance of voxels from the detector coils. Variations in field strength and background intensity between images introduces error when comparing the signal intensities of paravertebral muscle and fatty tissue. Adjusting muscle signal intensity to the signal intensity of subcutaneous back fat allowed comparison between participants. The fat ROI was selected from an area of subcutaneous back fat as defining a region of intermuscular fat may be difficult in every individual (D'Hooge et al., 2012). Other studies have used cerebrospinal fluid to adjust signal intensity (Battié et al., 1995; Videman et al., 1994; Gibbons et al., 1997). Deposition of fat and connective tissue show as high signal intensity on fast spin echo T2-weighted images (Kader, Wardlaw and Smith, 2000). Therefore, hyperintense regions within the paravertebral muscles observed on T2 axial images were considered fatty tissue (Teichtahl et al., 2015a; Kader, Wardlaw and Smith, 2000). It should be noted that a range of approaches to quantify LPM fat infiltration have been reported in the literature (Crawford et al., 2017). Whilst none of these have been validated, the current approach has shown high inter and intra-observer reliability (Hu et al., 2011), low variability in several studies (D'Hooge et al., 2012; Hyun et al., 2016; Valentin, Licka and Elliott, 2015; Gibbons et al., 1997) and minimises inter-subject variance seen with histogram techniques that do not correct for temporal variations in signal intensity.

5.2.2.3 Lumbar Lordotic Angle

Sagittal plane images were acquired to measure LLA. The images were acquired using a T2-weighted FRFSE sequence with a magnetic field strength of 3T, slice thickness 6 mm, interslice gap 8 mm, TR 2877 ms, TE 109 ms, flip angle 142°, voxel size 1.41 x 1.41 x 8 mm, images in acquisition 12. The slice representing the mid-vertebral line was identified by the presence of the conus medullaris and spinous processes and used for analysis. The Cobb L1-L5 method was used to measure LAA (Hong et al., 2010); the angle between the superior endplate of L1 and the inferior endplate of L5 ($\theta_{L1\text{-}L5}$) (**Figure 5.3**).

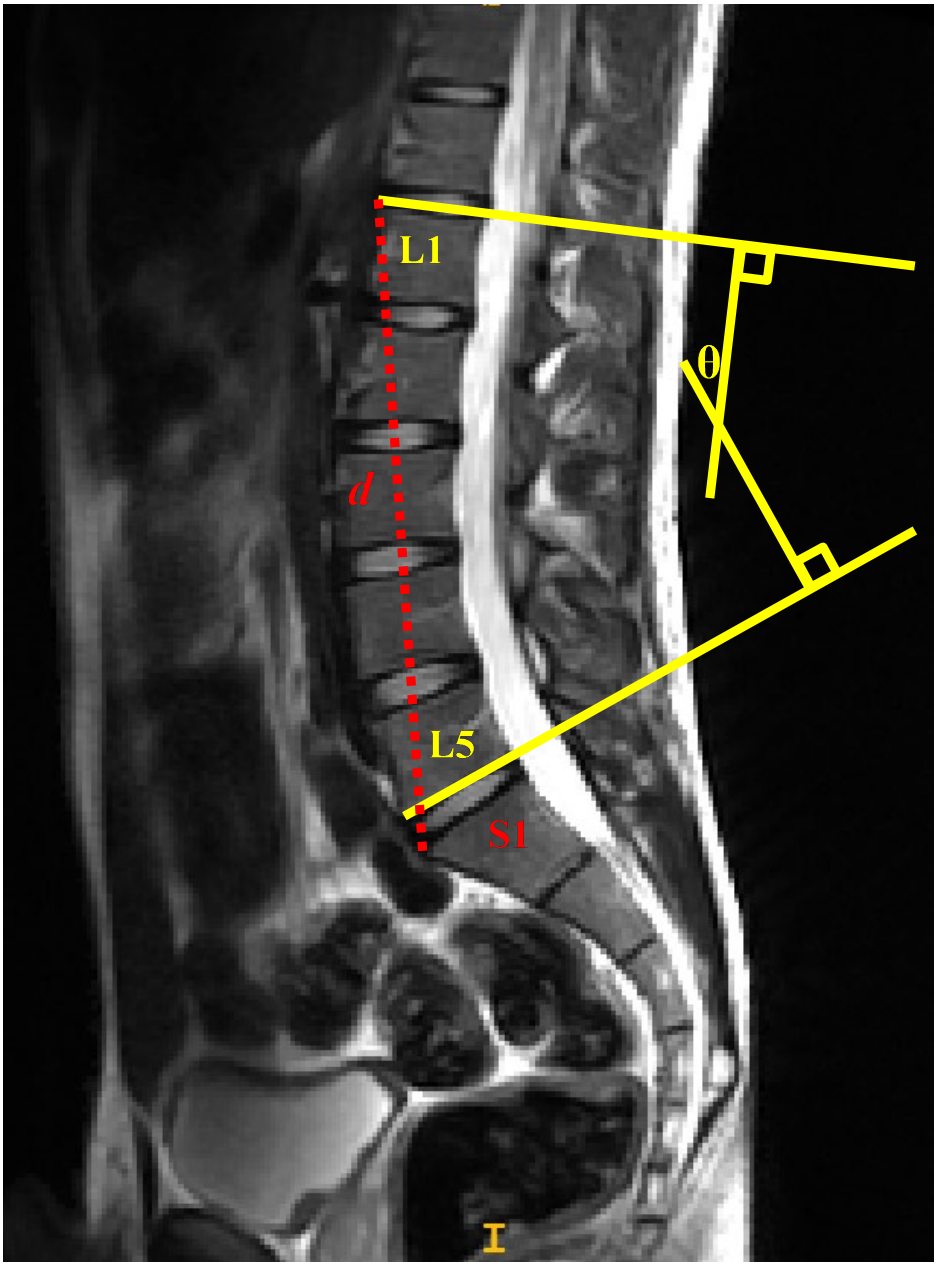


Figure 5.3 *LLA measured using Cobb's method. T2-weighted mid-sagittal MRI image with lines drawn along the superior endplate of L1 and inferior endplate of L5, extending past the vertebral body. Orthogonal lines were added, on the side of convergence of the two lines, and the angle of the intersection (ϑ) was measured. The straight-line distance (d) was measured between the anterior superior border of the L1 vertebra and the anterior superior border of the S1 vertebra.*

5.2.3 Statistical Analysis

Statistical analyses were performed using SPSS software (Version 24.0, IBM, Armonk, New York) and graphical presentation performed using GraphPad Prism (Version 8.3.1, San Diego, California). For each muscle group, right and left sides were combined and NMV and MFI were presented as mean ± standard deviation (SD) unless otherwise stated. All variables were normally distributed (Shapiro

Wilk's test: $p > .05$) and equal variances between groups were assumed (Levene's test: $p > .05$). Multivariate analysis of covariance (MANCOVA) were conducted for NMV and MFI; VPA was included as a potential covariate. Group differences were investigated based on the linear composite of outcome variables for each muscle. Significant between-subject results were followed up with univariate analysis of variance (ANOVA) to compare NMV and MFI for each muscle between age groups.

An independent t-test was used to compare LLA between groups. The strength of the relationships between LLA and muscle morphology outcomes were assessed from the Pearson correlation coefficient. Stepwise multiple linear regression was used to explore potential variables that may be related to total NMV (summation of each muscle's NMV) and mean MFI (mean MFI across all muscles). Input variables included: age (age group), MVPA (hrs/day), VPA (hrs/day), BMI, whole body fat composition (%), whole body lean mass (kg), dominant and non-dominant handgrip strength (kg), and LLA (°). A stepwise regression model was chosen due to its ability to reduce the number of predictor variables without substantially reducing the explanatory power of the data (Huang and Townshend, 2003). An alpha level of 0.05 was required for statistical significance in all tests. Standardized effect size (η_p^2) and observed power (1-β) were also determined where possible.

5.2.3.1 Reliability

Segmentation of the LPMs was performed independently and sequentially for every participant. Segmentation was repeated on a random sub-sample (n = 4) after six months to assess long-term intra-observer reliability and measurement error. To avoid bias, the observer was blinded to the first measurement before the second measurement was completed. The observer was also blinded to the participant's information. Intra-rater reliability of the NMV and MFI measurements were assessed by calculating the average measures intra-class correlation coefficient (ICC) using a two-way mixed absolute agreement model. Intra-rater reliability was excellent across measures of Volume, NMV and MFI. ICC [95% CI], root mean square difference (RMSD) and mean residual difference (%) values are presented in **Table 5.3**. Segmentation agreement maps for visualisation are presented in **Appendix k**, with reported values for the first and second measurements in **Appendix l**.

Table 5.3 *Intra-rater reliability for the MRI analysis outcome measures presented as Intraclass Correlation Coefficient (ICC) with 95% Confidence Intervals [95% CI], Root Mean Squared Difference (RMSD) and percentage mean difference in residuals.*

Outcome Measure	Muscle	ICC [95% CI]		RMSD	Mean Residual Difference (%)
Volume (cm^3)	PS	1.000	[.996, 1.000]	0.45	0.3
	QL	.997	[.966, 1.000]	1.49	2.6
	ES	.999	[.986, 1.000]	1.58	1.0
	MF	.995	[.943, 1.000]	0.91	1.5
NMV (a.u.)	PS	.998	[.976, 1.000]	0.07	1.2
	QL	.994	[.940, 1.000]	0.09	2.9
	ES	.997	[.970, 1.000]	0.11	1.2
	MF	.995	[.944, 1.000]	0.05	1.7
MFI (%)	PS	.918	[.075, .995]	0.58	3.4
	QL	.990	[.895, .999]	0.98	7.9
	ES	.998	[.973, 1.000]	0.66	2.0
	MF	.999	[.978, 1.000]	0.74	1.7
Vertebral Height (cm)		.982	[.710, .999]	0.2	1.0
LLA (°)		.989	[.844, .999]	1.9	3.3

PS = Psoas, QL = Quadratus lumborum, ES = Erector spinae, MF = Multifidus, NMV = Normalised Muscle Volume, MFI = Muscle-fat-infiltrate, LLA = Lumbar Lordotic Angle

5.3 Results

Descriptive statistics (mean ± SD) for each muscle outcome stratified by age group are presented in **Table 5.4**. MANOVA revealed statistically significant differences in NMV ($F(4,19) = 5.07$, $p = .006$; Wilks' $\Lambda = 0.48$, $\eta_p^2 = 0.52$) and MFI ($F(4,19) = 9.64$, $p < .001$; Wilks' $\Lambda = 0.33$, $\eta_p^2 = 0.67$) between age groups.

Table 5.4 *Means ± SD for muscle volume normalised to vertebral height (NMV) and mean intramuscular fat infiltration (MFI).*

	NMV (a.u.)			MFI (%)		
	Young	Old	Cohen's d	Young	Old	Cohen's d
Psoas	6.40 ± 0.85	6.09 ± 0.78	0.38	10.18 ± 1.77	13.04 ± 2.53**	1.31
Quadratus lumborum	3.03 ± 0.51	2.23 ± 0.50***	1.58	9.46 ± 1.56	14.87 ± 3.56***	1.97
Erector spinae	10.13 ± 1.00	8.94 ± 1.65*	0.87	13.48 ± 2.79	23.77 ± 5.56***	2.34
Multifidus	3.05 ± 0.50	3.38 ± 0.84	0.48	18.53 ± 4.74	33.48 ± 6.63***	2.59

Significant difference with young group * p < .05, ** p < .01, *** p < .001

5.3.1 Age-related Differences in Normalised Muscle Volume

Follow-up ANOVA revealed a significant effect of age on NMV for the QL ($F(1,22) = 15.98$, $p < .001$, $\eta_p^2 = 0.421$, $1\text{-}\beta = 0.968$) and ES ($F(1,22) = 4.77$, $p = .040$, $\eta_p^2 = 0.178$, $1\text{-}\beta = 0.551$) muscles (**Figure 5.4**). Compared to the YG, the OG had significantly lower NMV for the QL (2.2 ± 0.5 vs 3.0 ± 0.5 a.u.) and ES (8.9 ± 1.7 vs 10.1 ± 1.0 a.u.). Differences in NMV between groups were not significant for the PS and MF muscles. The greatest difference between groups was observed in the QL, where the YG exhibited a 36.47% greater NMV than the OG.

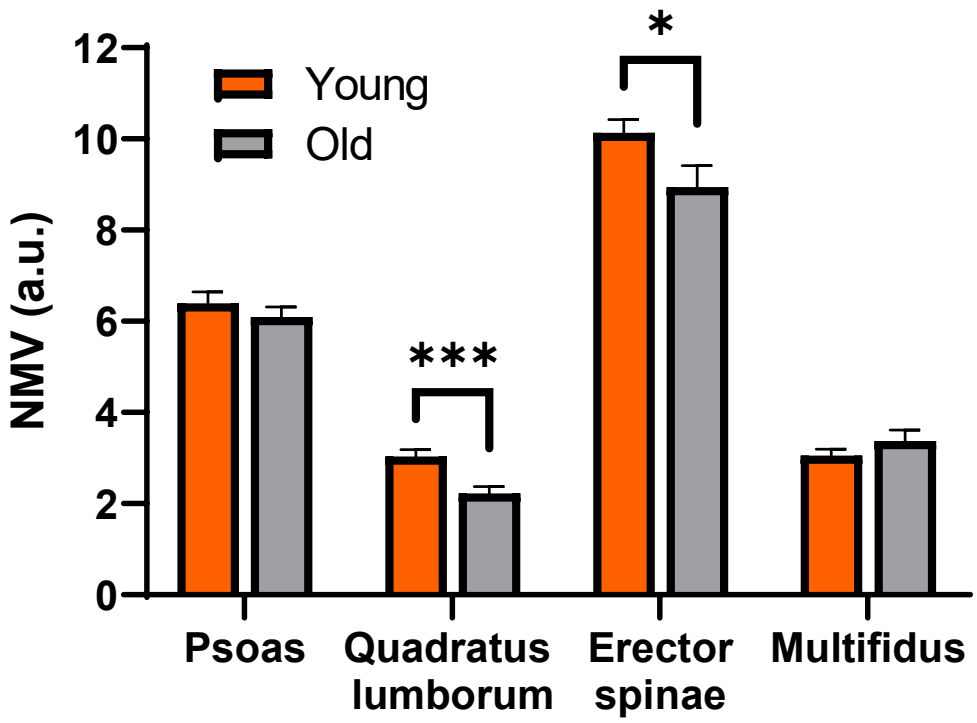


Figure 5.4 *Paravertebral muscle volume (L3 superior endplate - L4 superior endplate) normalised to L1-L5 vertebral height (mean ± SEM). * p < .05, ** p < .01, *** p < .001*

5.3.2 Age-related Differences in Intramuscular Fat Infiltration

There was a significant effect of age on fat infiltration for all muscle groups: PS ($F(1,22) = 10.30$, $p = .004$, $\eta_p^2 = 0.318$, $1-\beta = 0.864$); QL ($F(1,22) = 23.10$, $p < .001$, $\eta_p^2 = 0.512$, $1-\beta = 0.996$); ES ($F(1,22) = 32.73$, $p < .001$, $\eta_p^2 = 0.598$, $1-\beta = 1.0$); MF ($F(1,22) = 40.43$, $p < .001$, $\eta_p^2 = 0.648$, $1-\beta = 1.0$) (**Figure 5.5**). The greatest mean difference in MFI was observed in the MF, where MFI was significantly greater in the OG compared to the YG (33.48 ± 6.63 % vs 18. 53 ± 4.74 %). The OG also exhibited significantly greater fat infiltration in the PS (13.04 ± 2.53 % VS 10.18 ± 1.77 %), QL (14.87 ± 3.56 % vs 9.46 ± 1.56 %) and ES (23.77 ± 5.56 % vs 13.48 ± 2.79 %).

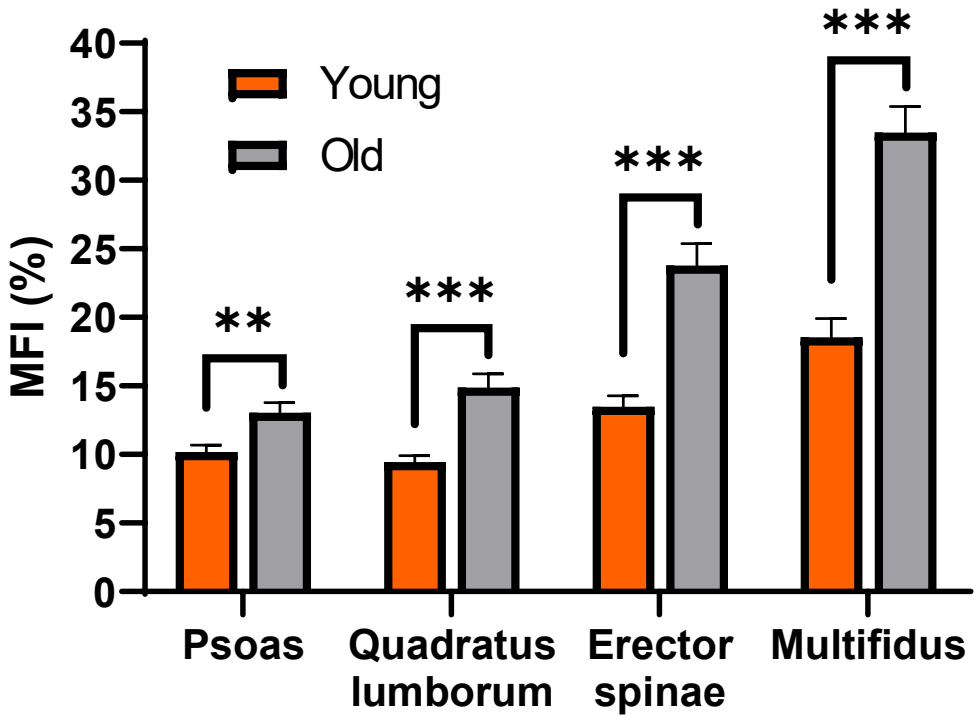


Figure 5.5 *Paravertebral muscle mean fat infiltration (mean ± SEM). ** p < .01, *** p < .001*

5.3.3 Moderating Effect of Physical Activity

VPA was not significantly related to NMV (p = .44) or MFI (p = .94), when included as a covariate in statistical models (MANCOVA). The effect of age on MFI for all muscles remained significant after controlling for VPA. However, whilst a significant main effect of age on NMV for the QL remained after controlling for VPA, age did not have a significant effect on NMV for the ES after controlling for VPA (p > .05). Adjusted values indicated that ES NMV remained lower in the OG (9.0 ± 1.5 a.u.) compared to the YG (10.0 ± 1.5 a.u.).

5.3.4 Lumbar Lordotic Angle

The OG (36.9 ± 10.6°) had a significantly greater LLA than the YG (26.6 ± 5.2°), $t(22) = 3.02$, p = .006; Cohen's d = 1.23. Of the morphological features measured across all muscles, only MF NMV was significantly correlated with LLA, $r(22) = 0.434$, p = .034.

5.3.5 Exploration of Influential Variables

Of the variables included in the stepwise linear regression, only non-dominant handgrip strength was found to be a significant predictor of total NMV, $F(1,22) = 10.93$, $p = .003$, $R^2 = 0.332$. Predicted total NMV was equal to 13.511 + 0.205*(non-dominant handgrip strength). Age was the only significant predictor of mean MFI, $F(1,22) = 38.27$, $p < .001$, $R^2 = 0.635$. Predicted mean MFI across all LPMs was equal to 12.922 + 8.370*(age group), where age group was coded as 0 = YG, 1 = OG. Mean MFI in the OG was 8.4 % greater than the YG.

5.4 Discussion

Ageing skeletal muscle is known to undergo adaptive reductive remodelling (Rogers and Evans, 1993; Roubenoff and Hughes, 2000; Kalichman, Carmeli and Been, 2017), characterised by muscle atrophy and fat infiltration (Marcus et al., 2010; Addison et al., 2014; Miljkovic et al., 2015; McGregor, Cameron-Smith and Poppitt, 2014; Frontera et al., 2000; Mitchell et al., 2012; Narici and Maffulli, 2010; Delmonico et al., 2009). The extent of these morphological alterations in the lumbar musculature is not fully understood, especially in healthy populations. Therefore, this study aimed to investigate age-related differences in LPM morphology in healthy younger versus older men. The main finding of this study was that the OG exhibited increased fat infiltration across the lumbar musculature whilst muscle atrophy was only found in the QL and ES muscles. These findings suggest that morphological changes in muscle fat content, rather than size, are a better indicator of age-related degeneration in the lumbar musculature.

Type I fibres are susceptible to fat infiltration with advancing age (Gueugneau et al., 2015; Choi et al., 2016; St-Jean-Pelletier et al., 2017), whilst type II fibres are more vulnerable to atrophic changes (Gueugneau et al., 2015; Novotny, Warren and Hamrick, 2015; Lexell, Taylor and Sjöström, 1988). Since the LPMs are predominantly composed of slow-twitch fibres (Ng et al., 1998; Mannion et al., 1997; Parkkola et al., 1993; Kimura, 2002; Sirca and Kostevc, 1985), it is unsurprising that changes in muscle fat content were more apparent than those in muscle volume. Other MRI studies have observed similar results, reporting greater fat infiltration with ageing than atrophy in the LPMs (Crawford et al., 2016a; Valentin, Licka and Elliott, 2015; Dahlqvist et al., 2017). A longitudinal study showed that muscle size decreased and fat infiltration increased with ageing over 15 years (Fortin et al., 2014), although the effect of age itself had a surprisingly small effect on the degree of change in fat infiltration. Whilst the current findings are supported by other studies assessing age-related degeneration of the muscles in the lumbar spine (Crawford et al., 2016a; Valentin, Licka and Elliott,

2015; Dahlqvist et al., 2017; Bayat et al., 2019), this study extends the concept of spinal sarcopenia (Kuo et al., 2020; Kim et al., 2019) by presenting volumetric and fat infiltration data for all of the main LPMs in relation to healthy ageing. This new knowledge will help in designing effective exercise training programmes that target the LPMs.

5.4.1 Fat Infiltration of the Lumbar Musculature

5.4.1.1 Psoas Muscle Fat Infiltration

The current results show that PS MFI is greater in healthy older than younger males, although support from the literature is equivocal. Whilst some studies show that fat infiltration increases in the PS with advancing age (Kita et al., 2013; Dahlqvist et al., 2017; Hedermann et al., 2018), others have shown the PS is resistant to age-related fat infiltration (Lee et al., 2017; Valentin, Licka and Elliott, 2015). Valentin and colleagues' (2015) sample was composed of healthy adults who were demographically similar to the current study's sample. A similar approach was also implemented to quantify fat infiltration. However, estimations in PS MFI of 37.0 ± 5.2 % in the older male group and 34.6 ± 6.4 % in the younger male group were considerably higher. The use of T1-weighted images may have caused disparity as pixel intensity represents different tissues compared with T2-weighted images (Bloem et al., 2018). Lee et al. (2017) also reported no ageing effects on PS muscle quality. Again, sample characteristics were comparable to the current study's except for ethnicity. It has been demonstrated that the rate of change to muscle fat content in the lumbar differs between Asian and Caucasian populations (Crawford, Elliott and Volken, 2017). Therefore, differences with Lee et al. (2017) may be due to ethnicity.

5.4.1.2 Quadratus Lumborum Fat Infiltration

This study showed that QL MFI increases in older age in healthy men. In support of this finding, Johannesdottir et al. (2018) and Lorbergs et al. (2019) observed moderate to strong effects of ageing on QL muscle density in large population-based cohorts. However, conflicting results have been reported (Kim, H. et al., 2013; Zhang et al., 2019; Gibbons et al., 1997; Sions et al., 2017b). Indeed, Anderson et al. (2013) suggested that the effect of ageing on muscle quality was significantly less in the QL compared to the average effect across muscles in the thoracic and lumbar spine, indicative of less fat accumulation with age. Whilst the studies supporting the current findings use CT, most of the aforementioned studies in opposition use MRI (Gibbons et al., 1997; Sions et al., 2017b; Kim, H. et al., 2013). Disparities may be due to the choice of MRI sequence. T2-weighted imaging is more likely to overestimate fatty muscle degeneration due to fat as well as water and high glycogen content

appearing hyperintense (Lollert et al., 2018; Bloem et al., 2018). However, Sions et al. (2017b) stated that intramuscular fat may comprise up to 54 % of QL CSA, compared to 14.9 ± 3.6 % in the current study. This discrepancy may be explained by the reference region of fat. A less intense extramuscular fat ROI was used by Sions et al. (2017b); the current study used a more intense subcutaneous fat ROI. Therefore, the relatively smaller difference in signal intensity between the fat reference and muscle ROI in Sions and colleagues' (2017b) study may have inflated estimations of fat infiltration in the QL muscle. Conflicting results are therefore likely due to technical differences; a factor that increases between-study heterogeneity as shown in **Chapter 4** and precludes conclusions from being drawn.

5.4.1.3 Paraspinal Fat Infiltration

The OG exhibited an increase in ES and MF MFI compared to the YG. This is consistent with the literature (Lorbergs et al., 2019; Johannesdottir et al., 2018; Crawford et al., 2016a; Shahidi et al., 2017b; Sasaki et al., 2017; Kalichman et al., 2010; Dahlqvist et al., 2017; Hedermann et al., 2018; Lee et al., 2017; Fortin et al., 2014), however, caution must be taken when comparing findings. Some studies have combined the MF and ES to define paraspinal muscle ROI (Dahlqvist et al., 2017; Hedermann et al., 2018), which may lead to overestimations in fat infiltration. An increase in non-contractile tissue between muscles is seen with advancing age (Addison et al., 2014). It is therefore likely that the inclusion of age-related increases in non-contractile tissue between the MF and ES results in greater estimates of paraspinal fat infiltration. Dahlqvist et al. (2017) and Hedermann et al. (2018) indicated that fat fraction within the paraspinals is approximately 30 to 35 % at the same age as the mean of the OG in the current study. This is comparable to the results for MF MFI (33.5 ± 6.6 %) but greater than MFI for the ES (23.8 ± 5.6 %), substantiating the assertion that overestimations are likely to occur when evaluating the ES and MF as a whole. Regardless, age explained a similar amount of variance in paraspinal muscle fat fraction (R^2 = 46 – 56 %) (Dahlqvist et al., 2017; Hedermann et al., 2018) as it did with MF and ES MFI in the current study (η_p^2 = 60 and 65 %, respectively); support that these muscles undergo similar morphological changes over time (Fortin et al., 2014).

5.4.1.4 Muscle-Specific Fat Infiltration

Whilst the effect of age on MFI was different for each muscle, significantly greater fat infiltration was observed across the lumbar musculature with age. Indeed, age accounted for a large proportion (63.5 %) of variation in mean MFI in the LPMs. This finding suggests that myosteatosis has a global effect on skeletal muscle in the lumbar spine. This reflects the effects of sarcopenia which are thought to be

systemic, making it difficult to explain localised changes in fat infiltration. However, efforts were made to explore muscle-specific differences in MFI in the lumbar spine and this should be considered the novel contribution of this chapter to the literature.

The greatest change in MFI was observed in the MF followed by the ES, whilst the PS exhibited the smallest decline. Muscle fibre-type distribution could be responsible for muscle-specific differences. Whilst the PS and MF/ES muscles are typically slow twitch phenotypes, there is a greater proportion of type I fibres in the MF/ES than the PS (Ng et al., 1998; Mannion et al., 1997; Parkkola et al., 1993; Kimura, 2002; Sirca and Kostevc, 1985; Rantanen, Rissanen and Kalimo, 1994), with the PS comprising over 25% of fast type II isoforms (Regev et al., 2010). Therefore, the deeper intrinsic muscles such as the MF and ES are likely to be more prone to fatty degeneration (Gueugneau et al., 2015; Choi et al., 2016). Muscle-specific denervation may also explain observed differences. The MF and ES are innervated by the dorsal rami of the spinal nerve (Kramer et al., 2001), while the ventral ramus innervates the PS (Mahan et al., 2017). Ageing is associated with degeneration of structures in the spine, such as nerves becoming compressed or impinged due to degenerative discs and vertebrae shifting (Benoist, 2003). Similar to wasting of the MF caused by Lumbar Dorsal Ramus Syndrome, this could lead to degeneration of structures innervated by the dorsal ramus nerve (e.g. MF and ES) (Kader, Wardlaw and Smith, 2000). However, there is insufficient evidence to suggest that dorsal rami are more damaged with ageing than ventral rami. Further research is needed to confirm the age-related effects on spinal nerve branches that innervate the different muscles in the lumbar spine.

5.4.2 Age-Related Differences in Muscle Volume

5.4.2.1 Quadratus Lumborum Atrophy

The QL exhibited the greatest age-related atrophy in the current study. Few studies have directly investigated the effect of ageing on QL muscle size (Sions et al., 2017b; Aboufazeli et al., 2018; Johannesdottir et al., 2018), however, these studies all reported atrophy of the QL muscle with ageing. The rate of atrophy in the QL may be as great as 9 % per decade in males; greater than any other muscle in the lumbar spine (Johannesdottir et al., 2018). Whilst this supports the current results, comparison with Johannesdottir et al. (2018) may be confounded by methodological differences such as imaging modality and muscle size measures. Despite disparities, conversion amongst effect sizes revealed that ageing had an almost identical effect on the change in QL muscle size in Johannesdottir et al's (2018) and the current study. Whilst more studies are needed to substantiate the magnitude of the ageing effect on QL muscle size, it appears that this muscle is vulnerable to age-related atrophy.

However, the mechanism for its degeneration is ambiguous. This is reflected in the lack of consensus regarding its action on the lumbar spine.

A plausible mechanism concerns hypertrophy of the QL in response to frontal plane segmental instability during dynamic movements (Ranson et al., 2008; de Visser et al., 2007). Whilst high impact movements, such as fast bowling in de Visser et al's (2007) and Ranson et al's (2008) studies, are not applicable to older individuals the same principles can be applied to typical activities performed in older adult populations, such as walking (Walsh et al., 2001). Older adults typically adopt a more conservative gait pattern to reduce trunk accelerations and maintain balance (Woollacott and Tang, 1997; Kavanagh, Barrett and Morrison, 2004). Therefore, it is likely a reduced demand on lateral stabilisation of the trunk in older age may lead to a detraining effect and subsequently disuse atrophy (Ikezoe et al., 2012). The effect of age on trunk biomechanics during walking gait, and the moderating effect of LPM morphology, is explored in **Chapter 7**.

5.4.2.2 Loss of Erector Spine Muscle Volume

The OG exhibited a significantly lower NMV for the ES. Compared to 41% for the QL, age accounted for only 17% of the variance in ES atrophy, suggesting that factors other than age play an important role in mediating the loss of ES muscle size. The low coefficient of determination for age is reflected in the literature as a number of researchers have observed negligible age-related decrements in ES muscle size (Valentin, Licka and Elliott, 2015; Crawford et al., 2016a; Hiepe et al., 2015). Age-related declines may be due to a combination of disuse and denervation atrophy in the ES. Skeletal muscle has been shown to undergo adaptive reductive remodelling in response to both physical inactivity (Paddon-Jones et al., 2006; Fortney, Schneider and Greenleaf, 2011) and ageing (Rogers and Evans, 1993; Roubenoff and Hughes, 2000; Kalichman, Carmeli and Been, 2017). The effect of muscle disuse atrophy in older adults is likely exacerbated by their inter-relationship (Wall, Dirks and Van Loon, 2013). According to Ikezoe et al. (2015), mechanical unloading preferentially affects the antigravity muscles. Given that the ES muscles are the primary antigravity muscles of the spine and are predominantly composed of type I muscle fibres (Ng et al., 1998; Mannion et al., 1997; Parkkola et al., 1993; Kimura, 2002; Sirca and Kostevc, 1985), which are susceptible to inactivity atrophy (Wang and Pessin, 2013), less engagement with VPA is a likely mechanism for ES muscle atrophy (Kalimo et al., 1989).

5.4.2.3 Preservation of Psoas and Multifidus Muscle Volume

Age-related atrophy of the PS and MF was not significant. Interestingly, MF NMV was larger in the OG. Change in LLA with age may explain why the MF is spared from atrophic decline. The NMV of the MF was significantly and moderately associated with LLA. Other studies have shown that lumbar MF muscle volume is moderately to strongly correlated with sagittal curvature of the lumbar spine (Meakin et al., 2013; Menezes-Reis et al., 2018), whilst the composition and volume of other muscles in the lumbar spine are not (Menezes-Reis et al., 2018). According to previous mathematical models, forces applied by the LPMs should be greater in spines with increased lumbar lordosis (Meakin and Aspden, 2012). Larger muscle forces are required to provide biomechanical stability in spines with greater lumbar curvature (Meakin and Aspden, 2012). Therefore, the significantly greater LLA in the OG may provide a training effect for the MF, whereby the MF plays a role in generating follower loads (Patwardhan et al., 1999) (i.e. resultant forces that travel tangentially to the spine's sagittal curvature to provide lumbar spine stability) (Aspden, 1989). It should be noted that in contrast to this study, others have observed reductions in LLA with ageing (Takeda et al., 2009; Hammerberg and Wood, 2003) although the mechanism for this may concern age-related degeneration of spinal features (Aylott et al., 2012; Takeda et al., 2009) and not the musculature.

Methodological decisions may also explain why MF NMV was greater in the OG. It is likely that inclusion of intramuscular fat within the ROI masked MF atrophy (Elliott et al., 2008). Furthermore, exclusion of adipose tissue extraneous to the muscle border may explain why atrophy was observed in the ES but not the MF. Fatty regions were more visible under the posterior thoracolumbar fascia lateral to the MF than directly under the MF (**Figure 5.6**). The propensity for fat to accumulate in this region means that the ES is likely to exhibit a loss in muscle size with ageing compared to the MF, when this region is excluded from the measurement.

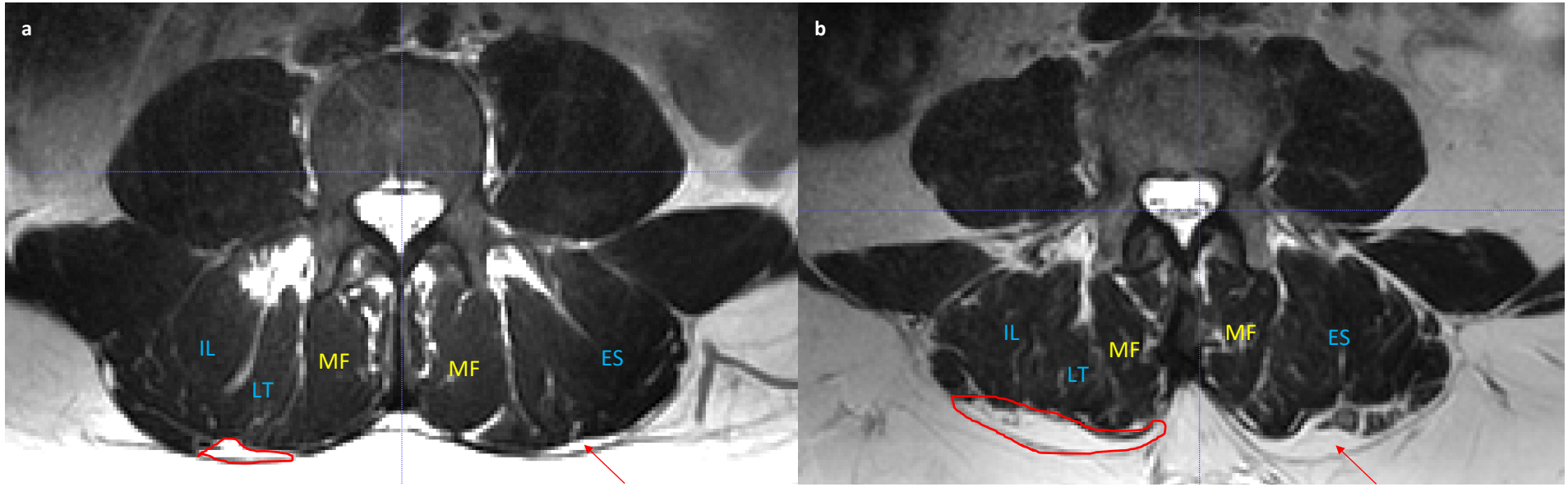


Figure 5.6 *Axial MRI images at the superior endplate L4 of a younger (a) and older (b) participant. The arrows and outlined areas identify the region of fatty tissue under the fascial plane and between the longissimus thoracis (LT) and iliocostalis lumborum (IL).*

Based on the current results, the PS does not appear to atrophy with ageing in healthy men. Previous studies are in agreement (Crawford et al., 2016a; Ikezoe et al., 2015; Hedermann et al., 2018) although not unequivocally so (Mäki et al., 2019). The relative preservation of PS NMV may be due to changes in motor control strategies in older age. Older adults exhibit a distal-to-proximal redistribution of joint power during gait (Cofré et al., 2011; DeVita and Hortobagyi, 2000), demonstrated by greater reliance on hip flexor activity to propel the leg into swing (Cofré et al., 2011). As the PS functions as a primary flexor of the hip joint (Penning, 2000; Juker, McGill and Kropf, 1998; Bogduk, Pearcy and Hadfield, 1992; Santaguida and McGill, 1995; Yoshio et al., 2002), increased reliance on the hip flexor muscles in older age may provide sufficient stimulus to attenuate atrophy of the PS.

5.4.2.4 Morphological Changes in the Paravertebral Muscles

Of the degenerative features investigated, fat infiltration appears to have a global effect on the LPMs whereas atrophy appears to be muscle-specific. Mechanisms for this remain undetermined, although the suggestions above provide plausible explanations. Briefly, increase in intramuscular fat tissue is likely due to slow-twitch fibre distribution in the postural muscles and propensity for these fibres to accumulate fatty deposits with ageing. Muscle-specific atrophy likely concerns the specific functions of the lumbar muscles and their exposure to reduced mechanical loading resulting from a shift in the locus of function in motor performance with ageing (DeVita and Hortobagyi, 2000). Furthermore, accretion of intramuscular fat may be an early change in muscle as it ages, which may explain why fat infiltration was the more apparent degenerative feature in the lumbar musculature.

5.4.3 Influence of Physical Activity on Muscle Degeneration

Controlling for VPA did not influence age-related differences in MFI. VPA also had no effect on age-related atrophy for the MF, PS and QL. However, VPA moderated age-related differences in ES muscle atrophy. This suggests that VPA may have a positive effect on attenuating ES muscle atrophy in older age, although this finding must be interpreted with caution as VPA was not significant as a covariate. Skeletal muscle tissue, like osseous tissue, is mechanoresponsive (Trumbull, Subramanian and Yildirim-Ayan, 2016). Yet PA seemingly has little or no effect on the size or fat content of the LPMs. In support of the current findings, it has been consistently demonstrated that PA does not relate to changes in LPM morphology with ageing (Dahlqvist et al., 2017; Lee et al., 2017; Anderson et al., 2013). Indeed, increases in PA levels over time has been shown to be ineffective at attenuating MF and ES muscle atrophy (Lee et al., 2017). This supports the idea that functional changes to the muscle in older age may be responsible for muscle degeneration rather than the mechanical loads exerted on them.

5.4.4 Handgrip Strength as a Predictor of Muscle Atrophy

Handgrip strength is well-established as an indicator of muscle status, particularly in older adult populations (Bohannon, 2015). However, its ability as a predictor of LPM atrophy has not been previously investigated to the author's knowledge. In this study, variance in non-dominant handgrip strength explained 33.2 % of the variance in the loss of total paravertebral NMV, which suggests that handgrip strength is a good indicator of muscle atrophy in the lumbar spine. However, mechanisms directly linking atrophy of the LPMs and forearm muscle strength are unlikely to exist. A more plausible explanation is that atrophy of the LPMs is simply representative of the systemic decline in muscle status throughout the body.

5.4.5 Clinical and Practical Applications

Being able to identify early signs of age-related muscle degeneration could assist clinical decision making with regards to the timely implementation of targeted intervention strategies. The current results indicate that T2-weighted MRI analysis is able to identify age-related differences in fat infiltration and atrophy of the LPMs. Whilst the Dixon sequence has been suggested as a superior method of measuring fat infiltration (Ma et al., 2004), this study showed that a routinely used sequence in clinical examinations (i.e. T2-weighted) is able to distinguish the age-effect in the LPMs. However, it should be noted that the approach used in this study (i.e. volumetric measures) may not be applicable in clinical settings due to the time cost. Manual segmentation is time consuming and requires a high level of expertise. Until automatic processes can be used to accurately and precisely segment the individual muscles that make up the lumbar musculature, measurements will likely remain as CSAs taken at single representative slices and aggregated values which masks the muscle-specific nature of age-related degeneration as shown in this chapter. These findings should support the use of volumetric measures in current research whilst future research should look to establish reliable auto-segmentation procedures to facilitate clinical use of MRI analysis in the lumbar spine musculature.

5.4.6 Limitations

The current study was limited by its cross-sectional design, although the wide age range and close matching of the groups mitigated this somewhat. Longitudinal studies are needed to infer causality between ageing and degeneration of the lumbar musculature. Another limitation concerned the

measure of fat infiltration. Fat infiltration derived from T2-weighted images, whilst a conventional and useful measure (Kim et al., 2019; Ploumis et al., 2011; Heo et al., 2019; Gibbons et al., 1997; Fortin, Yuan and Battié, 2013), has known limitations. T2-weighted imaging risks overestimating fatty muscle degeneration because water and other tissues in addition to fat appear hyperintense. The Dixon MRI technique has demonstrated its superiority over T1- and T2- weighted imaging techniques in terms of fat fraction quantification (Ma et al., 2004). In addition to providing objective measurement of fat infiltration on a continuous and observer-independent scale, it is able to detect small changes in fat infiltration not achievable with T1- and T2-weighted imaging (Ma, 2008). In the current study, fat fraction quantification using a Dixon MRI sequence was not possible due to the poor quality of the resulting images (**Figure 5.7**). A systematic error was evident within each individual fat fraction image (increase in noise posteriorly-to-anteriorly), whilst the error between participant scans was random making it impossible to reliably remove noise. The results from the Dixon fat fraction images suggested that the PS had the highest intramuscular fat content, which is in direct contrast to numerous other studies. Confidence in the current approach is high, given that the findings from the T2-weighted images are in agreement with the literature (Lee et al., 2017; Lorbergs et al., 2019; Johannesdottir et al., 2018; Anderson et al., 2013; Hedermann et al., 2018; Sions et al., 2017b) and similar in value to comparable populations (Dahlqvist et al., 2017; Crawford et al., 2016a). However, it should be noted that fat composition of the LPMs in the literature has a wide range of 2 % to 45 % in younger adults (Pezolato et al., 2012; Valentin, Licka and Elliott, 2015). Discrepant findings and variance between studies is the result of methodological differences (see **Chapter 4**). Therefore, consensus is needed for a standardised measurement of MRI-derived muscle morphology in the lumbar spine.

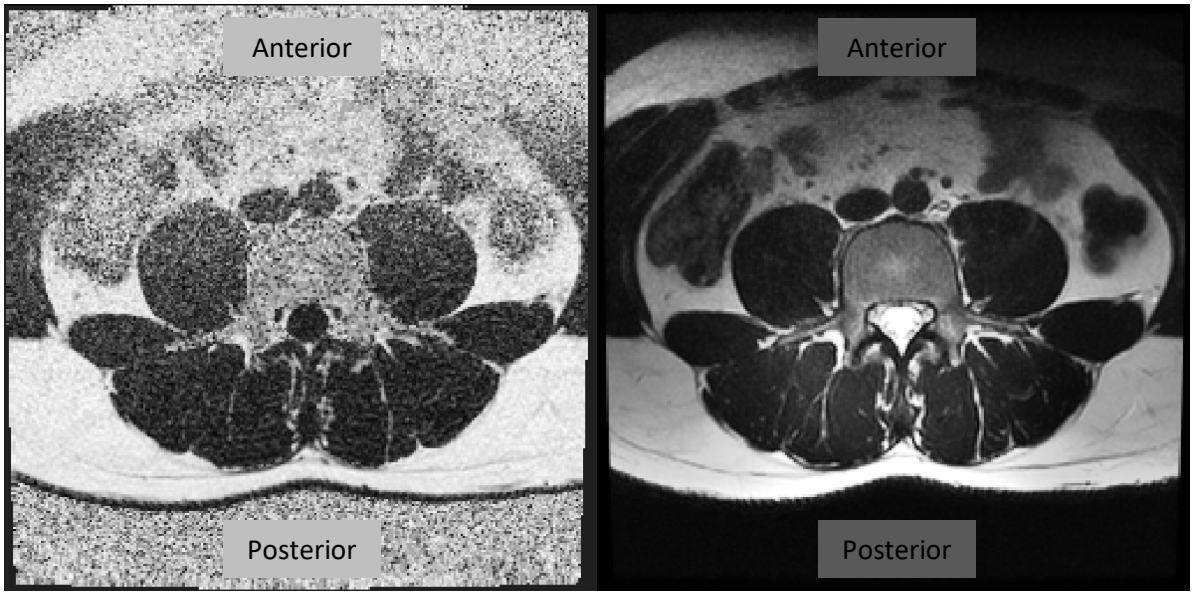


Figure 5.7 *Comparison between a) Dixon fat fraction image and b) T2-weighted image. The fat fraction image shows noise increasing posteriorly-to-anteriorly*

5.5 Conclusion

For the first time, this study provides age-related differences in volumetric and fat infiltration data for all the main LPMs. These findings will further understanding of age-related degeneration in the lumbar musculature and extend the concept of spinal sarcopenia, which is important in designing effective interventions that target the lumbar musculature as well as establishing normative features of ageing muscle morphology in this region. Furthermore, important relationships were revealed with exploratory variables (e.g. handgrip strength) that may enable early diagnosis of spinal sarcopenia. Such information would be valuable in clinical settings where handgrip strength tests can be easily performed and potentially allow interventions to be administered earlier. In this study older age had a detrimental effect on LPM morphology, although engaging in PA did not appear to attenuate muscle degeneration. The current findings indicate that age-related fat infiltration has a global effect across the lumbar musculature, whereas atrophic changes appear to be muscle-specific. The MF was most susceptible to compositional changes with age, whilst the QL exhibited the greatest reductions in muscle volume. Therefore, exercise interventions designed to attenuate the loss of muscle size and accumulation of fat in the lumbar spine may need to target specific muscles to maximise effectiveness. The design of effective muscle-specific exercise programs could potentially preserve mobility and reduce falls and injury risk in older adult populations. Whilst it was implied that muscle function is impaired by the accumulation of fatty deposits, changes in LPM morphology should be investigated with respect to muscle function in older age, specifically dynamic function under controlled conditions.

Table 5.5 *Thesis Map*

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance
		Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
		Aims	• To investigate age-related differences in LPM morphology

Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men	• Studies investigating muscle degeneration with ageing have typically focused on the appendicular muscles • There is increasing recognition for the importance of the lumbar paravertebral muscles in maintaining health and mobility in older age • Few studies have characterised features of age-related degeneration in the lumbar musculature • Few studies have provided volumetric information on all of the paravertebral muscles using high-resolution imaging modalities		• A secondary aim was to investigate the age-response on fat infiltration and volume of the different lumbar muscles (i.e. MF, ES, QL and PS) • An additional aim was to explore other predictors of lumbar paravertebral muscle degeneration
		Key findings	• Older age negatively affected all paravertebral muscles, although some showed greater degenerative changes than others • Age-related fat infiltration has a global effect across the lumbar musculature, whereas atrophic changes appear to be muscle-specific • Only the QL and ES showed significant age-related declines in muscle volume • All muscles showed age-related declines in muscle quality (i.e. increase in intramuscular adipose tissue) • The MF was most susceptible to compositional changes with age, whilst the QL was most vulnerable to reductions in muscle volume • Physical activity did not influence age-related differences in muscle degeneration in the lumbar spine • Non-dominant handgrip strength was a predictor of muscle atrophy in the lumbar musculature

	Implications	• The QL and ES appear to be most affected in older age since they exhibited declines in size and quality • When investigating the effects of ageing on lumbar muscle function, macroscopic changes in the paravertebral muscles should be considered • Structural changes, resulting in a loss of contractile tissue, may reduce muscle function in the lumbar spine • Convenient and easily administered measures such as handgrip strength may be able to predict muscle atrophy in the lumbar spine
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men	Aim	
	Key findings	
	Implications	
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men	Aim	
	Key findings	
	Implications	

Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men

The work from this chapter has been accepted for publication in a peer-reviewed journal.

Dallaway, A., Hattersley, J., Tallis, J., Renshaw, D., Griffen, C., and Duncan, M. (in press) 'Age-related changes in concentric and eccentric isokinetic peak torque of the trunk muscles in healthy older versus younger men'. *Journal of Aging and Physical Activity*

Chapter Abstract

Background The maximal capacity of the trunk muscles to generate torque is important for postural support and aiding in the performance of daily activities. It is well-known that appendicular muscle strength decreases with ageing, which has detrimental consequences on physical function. However, strength loss in the trunk musculature as a function of normal ageing is not fully understood. Therefore, this study investigated age-related differences in dynamic trunk muscle function in healthy men and the moderating effect of PA.

Methods Twelve healthy older (67.3 ± 6.0 years) and 12 healthy younger men (24.7 ± 3.1 years) performed isokinetic trunk flexion and extension tests across a range of angular velocities (15°·s^{-1} - 180°·s^{-1}) and contractile modes (concentric and eccentric). Peak isokinetic torque normalised to body mass was obtained for each condition.

Results For concentric trunk extension, mixed-effects ANCOVA revealed a significant interaction between angular velocity x age group ($p = .026$) controlling for VPA. Follow-up univariate ANCOVA revealed that the YG produced significantly greater peak torque for each concentric extension condition. Both groups exhibited a general decline in peak torque with increasing angular velocity, although strength loss was greater in the OG with increasing angular velocity. No significant interactions or main effects were observed for any other condition.

Conclusions The normal loss of trunk muscle strength in older age is muscle and contractile mode specific. Concentric strength of the trunk extensor muscles decreases in older age and with increasing angular velocity. Loss of concentric flexor strength is somewhat attenuated, whilst eccentric strength of the trunk muscles is preserved in older age. These findings should contribute to the early identification of trunk strength deficits, which will assist public health and clinical decision making with regards to timely implementation of targeted intervention strategies.

Key words: muscle strength, ageing, sarcopenia, abdominal muscles, paravertebral muscles

Table 6.1 Thesis Map

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance
		Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men	• Studies investigating muscle degeneration with ageing have typically focused on the appendicular muscles • There is increasing recognition for the importance of the lumbar paravertebral muscles in maintaining health and mobility in older age	Aims	• To investigate age-related differences in LPM morphology • A secondary aim was to investigate the age-response on fat infiltration and volume of the different lumbar muscles (i.e. MF, ES, QL and PS) • An additional aim was to explore other predictors of lumbar paravertebral muscle degeneration

	• Few studies have characterised features of age-related degeneration in the lumbar musculature • Few studies have provided volumetric information on all of the paravertebral muscles using high-resolution imaging modalities	Key findings	• Older age negatively affected all paravertebral muscles, although some showed greater degenerative changes than others • Age-related fat infiltration has a global effect across the lumbar musculature, whereas atrophic changes appear to be muscle-specific • Only the QL and ES showed significant age-related declines in muscle volume • All muscles showed age-related declines in muscle quality (i.e. increase in intramuscular adipose tissue) • The MF was most susceptible to compositional changes with age, whilst the QL was most vulnerable to reductions in muscle volume • Physical activity did not influence age-related differences in muscle degeneration in the lumbar spine • Non-dominant handgrip strength was a predictor of muscle atrophy in the lumbar musculature
		Implications	• The QL and ES appear to be most affected in older age since they exhibited declines in size and quality • When investigating the effects of ageing on lumbar muscle function, macroscopic changes in the paravertebral muscles should be considered • Structural changes, resulting in a loss of contractile tissue, may reduce muscle function in the lumbar spine • Convenient and easily administered measures such as handgrip strength may be able to predict muscle atrophy in the lumbar spine
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men	• Dynamic trunk strength in older adults has not been fully explored • Studies have typically investigated age-related strength loss using handgrip dynamometry or lower limb isokinetic dynamometry • Majority of studies have used clinical assessments which may not be appropriate to assess maximal trunk strength • No study has assessed eccentric trunk strength in older adults and contractile modes are typically limited • The findings from chapter 5 have also influenced the need for this study. Research investigating how muscle	Aims	• To investigate age-related differences in dynamic trunk strength • The secondary aim was to explore the moderating effect of muscle morphology degeneration on extensor muscle strength
		Key findings	
		Implications	

	morphology degeneration in the lumbar spine impacts on trunk extensor strength is warranted	
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men		Aim
		Key findings
		Implications

denotes links to previous chapters. Links to chapter 3 – 1) VPA was included as a potential covariate; 2) The moderating effects of body composition measures and handgrip strength were explored with respect to trunk strength loss. Links to chapter 5 – Whilst it was implied that muscle function was impaired by the accumulation of fatty deposits in chapter 5, changes in LPM morphology should be investigated with respect to dynamic muscle function in older age. Therefore, muscle morphology measures were included as potential moderators to assess whether trunk strength changes were associated with atrophy or fat infiltration of specific paravertebral muscles.

CHAPTER 6

6.1 Introduction

Skeletal muscle atrophy is associated with muscle weakness, however, studies have shown that the rate of strength loss is disproportionately greater than muscle atrophy (von Haehling, Morley and Anker, 2010; Frontera et al., 1991; Goodpaster et al., 2006; Metter et al., 1999; Overend et al., 1992; Delmonico et al., 2009; Narici and Maffulli, 2010; Mitchell et al., 2012) and better in predicting adverse health outcomes (Schaap, Koster and Visser, 2013; Menant et al., 2017; Schaap et al., 2018). Indeed, low muscle strength is now the principal determinant for identifying sarcopenia (Cruz-Jentoft et al., 2019). In the previous chapter it was shown that the LPMs undergo age-related atrophy and fat infiltration. To understand the implications of this, assessment of dynamic muscle function is needed. Studies investigating age-related declines in muscle function typically focus on the appendicular musculature (Young, Stokes and Crowe, 1985; Rogers and Evans, 1993; Faulkner, Brooks and Zerba, 1991; Heath et al., 1981; Mitchell et al., 2012), despite growing evidence for the importance of trunk muscles in performing ADLs (Hicks et al., 2005a; Granacher et al., 2013; Shahtahmassebi et al., 2017; Higuchi et al., 2018; Hernandez, Goldberg and Alexander, 2010) and constituting an important factor for overall health (Zouita et al., 2018; Ebenbichler et al., 2001; Valentin, Licka and Elliott, 2015; Crawford et al., 2016c; Cho et al., 2014).

The abdominal and LPMs are inextricably linked, controlling trunk movement and promoting mechanical stability in the lumbopelvic region (Barr, Griggs and Cadby, 2005; Gardner-Morse and Stokes, 1998; Cholewicki, Juluru and McGill, 1999). The importance of maintaining strength in the lumbar extensor muscles is highlighted by the large forces they generate. Due to a relatively small moment arm, the lumbar extensor muscles must produce a substantially larger force than the weight of the upper torso and ventral loads to counterbalance the external moment. In older adults, decreased neuromuscular control of the trunk muscles compromises their ability to stabilise the spine in response to perturbations in the environment, which increases susceptibility to injury (Mannion, Adams and Dolan, 2000; Hwang et al., 2008). A strength reserve is therefore needed to react to unpredictable occurrences such as falls, sudden loading of the spine and quick movements (Barr, Griggs and Cadby, 2005). A sudden need to regain spinal stability may also result in excessive muscle activity; a mechanism implicated in the genesis of LBP and injury (Cholewicki and McGill, 1996; Mannion, Adams and Dolan, 2000). Since older adults exhibit slower trunk movements during ADLs (McGill, Yingling and Peach, 1999), it is imperative that trunk strength is maintained for balance (Granacher et al., 2013; Suri et al., 2009) and to mitigate excessive muscle activity in response to instability (Anderson and Behm, 2005). Therefore, maximum strength of the trunk muscles is an important factor in older adults when dynamic stabilisation is required (Rantanen, Era and Heikkinen, 1994).

The effect of ageing on isokinetic trunk strength has been seldom studied with only a few studies reporting on adults over 50 years of age (Lee et al., 2012; Hasue, Fujiwara and Kikuchi, 1980; Gomez et al., 1991; Langrana and Lee, 1984; Hulens et al., 2002; Danneskiold-Samsøe et al., 2009). Of these studies, the effect of age on trunk strength varies from insignificant to large and it is unclear whether the age-response is equivalent between the abdominal and paravertebral muscles. Confounding factors which are often overlooked, such as PA level, may also moderate the age-response (Rantanen, Era and Heikkinen, 1997). Furthermore, the lack of consensus regarding isokinetic parameters, such as ROM limits, angular velocity and contractile mode, precludes conclusions from being drawn on age-related loss of trunk strength. Most importantly, research on eccentric trunk strength with respect to ageing does not exist to the author's knowledge, leaving a considerable gap in our understanding of dynamic muscle function in older age.

6.1.1 Strength Measurement in the Trunk Muscles

Trunk extensor performance has typically been assessed with clinical tests, which usually measure endurance rather than absolute strength (Demoulin et al., 2012). In a recent review, Prieske, Muehlbauer and Granacher (2016) suggested that applied trunk muscle strength tests lack external validity as they do not evaluate the maximal force producing capacity of the trunk muscles appropriately for dynamic activities. However, clinical assessments such as the Sorensen test (Biering-Sorensen, 1984; Demoulin et al., 2006) are quick and easy to perform, do not require specific equipment and are inexpensive. This often makes clinical tests more feasible compared to dynamometric tests. Dynamometric testing machines are expensive and have a high operational complexity (Barbado et al., 2016; Shahtahmassebi et al., 2017) although they provide a more precise, accurate and specific assessment of trunk muscle function (Demoulin et al., 2012). Studies using dynamometric approaches to evaluate trunk strength have typically opted for isometric conditions (Granacher et al., 2014; Shahtahmassebi et al., 2017; Sasaki et al., 2018; Hernandez, Goldberg and Alexander, 2010; Kassebaum et al., 2016; Sinaki et al., 2001; Porto et al., 2020). Although isometric measures provide valid and reliable outcomes for peak torque of the trunk musculature (Roth et al., 2017; De Blaiser et al., 2018), torque measured at a single standardised joint angle may not reflect muscle function across the functional ROM effectively (Rousanoglou and Boudolos, 2008). Given that torque production is joint angle dependent (Samuel and Rowe, 2009), measurement at one or a few discrete joint angles provides limited information about the force generating capacity of the muscles. Continuous measurement of joint torque elicits more detailed evaluation of a muscle group's function under controlled movement conditions. Furthermore, assessing force generation whilst the muscle is shortening and lengthening is more indicative of dynamic muscle activity during ADLs. Indeed,

isokinetic dynamometry permits continuous measurement through a predetermined ROM. It is a valid (Mueller et al., 2012) and widely accepted tool for measuring trunk muscle strength (Newton et al., 1993) and considered the gold-standard approach for dynamic muscle performance testing (Felicio et al., 2014; Dvir and Müller, 2019; Stark et al., 2011; Nugent, Snodgrass and Callister, 2015; Dvir, 2004).

6.1.2 Age-related Strength Loss in the Trunk

While several studies have assessed trunk muscle strength for sports performance (Barbado et al., 2016; Williams and Singh, 1997; Iwai et al., 2008), injury risk identification (Lee et al., 1999; Yahia et al., 2010; Cho et al., 2014) and progress monitoring of rehabilitation programs (Bayramoğlu et al., 2001; Ganzit et al., 1998), few studies have investigated strength loss in the trunk muscles with healthy ageing. Of these studies (Lee et al., 2012; Hasue, Fujiwara and Kikuchi, 1980; Langrana and Lee, 1984; Hulens et al., 2002; Danneskiold-Samsøe et al., 2009; Bidwell, Thauvette and Townshend, 1993; Sasaki et al., 2018), sampling variance and disparate methodologies makes it difficult to compare findings with any degree of confidence. An interesting finding that was consistent amongst studies was that reductions in strength loss appear to be biphasic. Numerous researchers (Danneskiold-Samsøe et al., 2009; Smith et al., 1985; Hause, Fujiwara and Kikuchi, 1980) have reported that trunk strength is largely maintained up till the 5th decade of life in men, which then declines specifically in the extensor muscles (Smith et al., 1985). Women also demonstrate a biphasic reduction in trunk muscle strength, with substantial declines in trunk extensor and flexor torque after the age of 60 years (Skrzek and Bolanowski, 2006).

Although the relationship between ageing and trunk strength appears unequivocal, these findings provide a limited understanding and caution must be taken when inferring relationships due to between-study variance. One source of variance between studies concerns the different isokinetic machines that were used. Greenberger, Wilkowski and Belyea (1994) suggest that errors are likely when comparing results from different dynamometers, possibly due to moderate inter-machine reliability (Bandy and McLaughlin, 1993; Lund et al., 2005). Positioning of participants also differed between prone/supine, standing and seated, which may affect trunk strength outcomes due to differing contributions from pelvic girdle and lower limb muscles (Langrana and Lee, 1984). This is exacerbated by the lack of consensus regarding ROM. Studies used maximum ROM from full extension to flexion, while others assessed trunk strength from 20° extension to between 30°-60° flexion. Amongst all the parameters, the angular velocity used to measure trunk muscle strength varied considerably (6°s^{-1} to 180°s^{-1}), although 30°s^{-1}, 60°s^{-1} and 120°s^{-1} were most common. A wide range of angular velocities at a greater number of increments would yield a more comprehensive

understanding of trunk muscle function. Another potential source of error is gravity correction; a procedure that corrects for gravitational torque produced by the limb-lever system (Baltzopoulos, 2008). Failing to correct for gravity has shown to produce up to 52% greater trunk extension peak torque (Hulens et al., 2002). Despite this procedure being particularly important when testing the trunk as its larger mass accounts for a substantial gravitational moment, few studies have implemented it (Danneskiold-Samsøe et al., 2009; Hulens et al., 2002). Comparisons may therefore be confounded by overestimated flexor torques and underestimated extensor torques of the trunk muscles. Finally, none of these studies (Lee et al., 2012; Hasue, Fujiwara and Kikuchi, 1980; Langrana and Lee, 1984; Hulens et al., 2002; Danneskiold-Samsøe et al., 2009; Bidwell, Thauvette and Townshend, 1993) indicated that peak torque was obtained during the stable isokinetic phase of the movement. Indeed, one study suggested that their results may be invalid due to inclusion of torque overshoot (Bidwell, Thauvette and Townshend, 1993). Perhaps the most obvious oversight in the literature does not concern methodological differences, but the fact that no study to date has explored the effect of age on eccentric muscle function in the trunk. This represents a considerable gap in the literature and severely limits our understanding of dynamic trunk muscle function in older age. Characteristics of the studies mentioned in this literature review are presented in **Table 6.2**.

Table 6.2 Characteristics of studies investigating age-related changes in isokinetic trunk muscle strength

Study	Study sample characteristics	Isokinetic dynamometer	Isokinetic test conditions	Gravity correction	ROM	Findings
Hasue, Fujiwara and Kikuchi (1980)	Healthy males (n = 50) and females (n = 50) age stratified into decades ranging from 2nd to 6th decade of life	Cybex	Concentric flexion in a supine position and concentric extension in a prone position. Tests were performed at 12°.s^{-1} for flexion and at 6°.s^{-1} for extension	✗	?	Ageing was associated with declines in trunk strength. r = -0.58 for flexion and r = -0.53 for extension trunk strength in males. r = -0.57 for flexion and r = -0.55 for extension trunk strength in females. Decline in strength was greater after the age of 40 years.
Langrana and Lee (1984)	Males employees (n = 140), aged 20-65 years, working at a heavy manufacturing plant. n = 19 with a previous history of back pain problems were excluded from the analyses	Cybex II	Concentric flexion and extension in a seated position. Tests were performed at 30°.s^{-1}	✗	~ -20° (extension) to ~ +40° (flexion)	Ageing was associated with loss of trunk strength. Flexion torque decreased from approximately 170 Nm to 70 Nm, whilst extension torque decreased from approximately 230 Nm to 110 Nm.
Smith et al. (1985)	Sedentary and recreationally active male (n = 62) and female (n = 63) volunteers without neuromuscular disorders or back pain were stratified into age categories of 18-29 years, 30-44 years and, > 45 years	Cybex II	Concentric flexion and extension in a standing position. Tests were performed at 30°.s^{-1}, 60°.s^{-1}, 90°.s^{-1}, 120°.s^{-1} and 150°.s^{-1} (data not reported for 150°.s^{-1})	✗	~ 0° (neutral) to ~ +50° (flexion)	The extensor, but not flexor, muscles appeared slightly weaker with advancing age in men. There were no significant differences in flexor or extensor strength between the young and middle age groups. However, strength reduced after 45 years, specifically in the extensors of males. The same decrements were not seen in women.
Gomez et al. (1991)	Males (n = 85) and females (n = 83) without LBP or history of back surgery were stratified into age categories of < 30 years, 30-39 years, 40-49 years and > 50 years	Isostation B-200 lumbar dynamometer	Concentric extension and flexion in a standing position. Test velocity determined by 55% of peak isometric extension torque	✗	Maximal flexion and extension	Strength was assessed through peak velocity against 55% of peak isometric extension torque. Peak flexion and extension velocity declined past the age of 50 years for women and 40 years for men.

Bidwell, Thauvette and Townshend (1993)	Healthy male (n = 8) and female (n = 12) volunteers aged stratified into 6th and 7th decades of life	Biodex with back station attachment	Concentric flexion and extension in a seated position. Tests were performed at 60°.s^{-1}, 120°.s^{-1} and 180°.s^{-1}	✗	Individually adjusted according to participant's maximal flexion and extension	Strength decreased with age for women but increased with age for men. Inferential statistics not used.
Hulens et al. (2002)	Obese Caucasian women (n = 241) aged 39 ± 12 years	Cybex TEF unit	Concentric flexion and extension in a standing position. Tests were performed at 60°.s^{-1} and 120°.s^{-1}	Proportional mass of the trunk was used to calculate the gravity effect torque (GET)	?	Significantly less trunk flexion and extension torque at both angular velocities in the older group (41-59 years) compared to the younger group (18-40 years). At 60°.s^{-1} low to moderate significant correlations were observed between ageing and trunk strength (r = -0.37 for extension and r = -0.16 for flexion).
Skrzek and Bolanowski (2006)	Women (n = 288) stratified into four age groups by decade ranging from 5th to 8th decade of life	Biodex System 3 Multi Joint	Concentric flexion and extension in a seated position. Tests were performed at 90°.s^{-1} and 120°.s^{-1}	✗	-20° (extension) to +50° (flexion)	Isokinetic extensor and flexor torque (normalised to body mass) significantly decreased with age. A substantial decline in trunk strength was observed after the age of 60 years at both isokinetic velocities.
Danneskiold-Samsøe et al. (2009)	Randomly selected sample of healthy community-dwelling adults (n = 53 m : 121 f) stratified into age categories by decade ranging from 3rd to 8th decade of life	Lido Active	Concentric flexion and extension in a seated position. Tests were performed at 10°.s^{-1}, 20°.s^{-1} and 30°.s^{-1}	Standard gravity correction procedure for LIDO	-20° (extension) to ~ +30° (flexion)	Isokinetic muscle strength decreased with age. For males, the decline in strength happened in two steps. A smaller decline was seen from 40–49 and an increased decline was seen from 50–59. Trunk strength in women declined from 40–49. Trunk extension torque decreased approximately 80-90 Nm, whilst flexion torque decreased about 40-60 Nm across all test conditions.

Lee et al. (2012)	Patients (n = 7 m : 21 f) aged 48 ± 12 years with chronic low back pain	Biodex	Concentric flexion and extension in a seated position. Tests were performed at 60°.s^{-1}	✗	Individually adjusted according to the patient's maximal flexion and extension	Age was significantly associated with trunk flexion strength loss in females only (β = -1.32, p = .01).

Note: ✗ *denotes not reported*

6.1.3 Aims, Objectives and Hypotheses

The loss of strength in older age is detrimental to physical function and is associated with adverse health outcomes, however, normal age-related decline in trunk strength is not fully understood. This study aimed to investigate age-related differences in trunk muscle strength in healthy men. The secondary aim was to explore the moderating effect of muscle degeneration of the LPMs on trunk extensor strength. In order to accomplish the aims, specific objectives for this study were to:

Table 6.3 *Objectives and hypotheses for chapter 6*

	Objective	Null Hypothesis
1	Identify appropriate parameters and develop a comprehensive protocol to assess dynamic trunk strength	n/a
2	Measure concentric and eccentric torque of the trunk extensor and flexor muscles in healthy older and younger males using isokinetic dynamometry	n/a
3	Analyse data to allow identification of peak torque values within the isokinetic phase of the movement	n/a
4	Compare peak torque values between the YG and OG using appropriate statistical tests	a) Peak concentric trunk strength will not be significantly lower in the OG compared to the YG b) Peak eccentric trunk strength will not be significantly lower in the OG compared to the YG
5	Investigate the effect of angular velocity on isokinetic trunk strength	Angular velocity will not have a significant effect on isokinetic trunk strength
6	Explore the moderating effect of muscle morphology measures by including them as potential covariates in statistical models	a) Degeneration of the lumbar musculature will not be associated with changes in concentric extensor peak torque with age b) Degeneration of the lumbar musculature will not be associated with changes in eccentric extensor peak torque with age

6.2 Methods

6.2.1 Equipment

A Trunk Modular Component (TMC) docked to a HUMAC® NORM™ isokinetic dynamometer (HUMAC® NORM™ Testing and Rehabilitation System, CSMI, MA, US) with proprietary software (HUMAC® 2009, v10.000.0082) was used for data acquisition. Calibration was performed according to the manufacturer's guidelines before each testing session. The TMC (**Figure 6.1**) was docked on the base of the Humac® NORM™ and the adapter placed into the input arm to enable the system to assess trunk flexion/extension performance.

6.2.2 Participant Positioning

Incorrect positioning and stabilisation of participants, and differences in posture, can affect the fidelity of isokinetic dynamometry data (Shirado et al., 1995). Before each test session, the TMC was adjusted for each participant for maximal comfort while ensuring alignment of the dynamometer axis to the rotation axis of the trunk (**Appendix m**). The setup data were recorded and reproduced in following sessions. Participants stood on the footplate of the TMC with their heels placed against the footplate heel cups. Adhering to manufacturer's recommendations, the footplate height was adjusted to achieve alignment between the participant's vertical anatomical axis and the machine's axis. The rubber alignment pointer was positioned approximately 3.5 cm below the top of the participant's iliac crest, in direct line with the first segment below the iliac crest (approximately L5/S1). The pelvic belt was secured across the top of the anterior superior iliac spines (ASIS). Popliteal pads were positioned directly behind the patellae at the popliteal space. After the popliteal pad height was adjusted, a thigh pad was secured in place directly superior to the patellae and a tibial pad was secured directly inferior to the patellae. Participants assumed a comfortable position (standing with ≈ 15° knee flexion), whilst being stabilised by the lower body pads. Whilst supported against the sacral seat pad, the fore/aft position was adjusted until the rubber alignment pointer was approximately centred at the intersection of the participant's mid-axillary line and lumbosacral junction. After the scapular pad was positioned across the centre of the scapulae and inferior to the spine of the scapulae, the chest pad was attached parallel to the scapular pad and tightened to prevent excessive upper body movement. Participants held the downward-facing handle in front of the chest to prevent motion of the upper limbs. Restricting upper-body motion and stabilising the lower body was performed to avoid extraneous movements and minimise the unwanted contribution of muscles not being tested (Smith et al., 1985; Pollock et al., 1989; Shirado et al., 1995). Whilst standing the participant's anatomical zero position was determined as an angle of 0° between the trunk and thighs. Mechanical stoppers were

applied as a safety precaution to limit the participant's movement to within their maximum ROM. Alignment of the participant to the dynamometer was recorded to allow reproduction during subsequent sessions. Gravity correction was performed according to the manufacturer's instructions (i.e. trunk segment perpendicular to ground) to minimise the effect of gravity on reciprocal muscle groups (Baltzopoulos, 2008; Westing and Seger, 1989; Fillyaw, Bevins and Fernandez, 1986; Edouard, Calmels and Degache, 2009; Sugimoto et al., 2014; Hulens et al., 2002). Measured torque data were then adjusted based on the maximum gravity effected torque of the trunk as a product of the cosine of its angle.

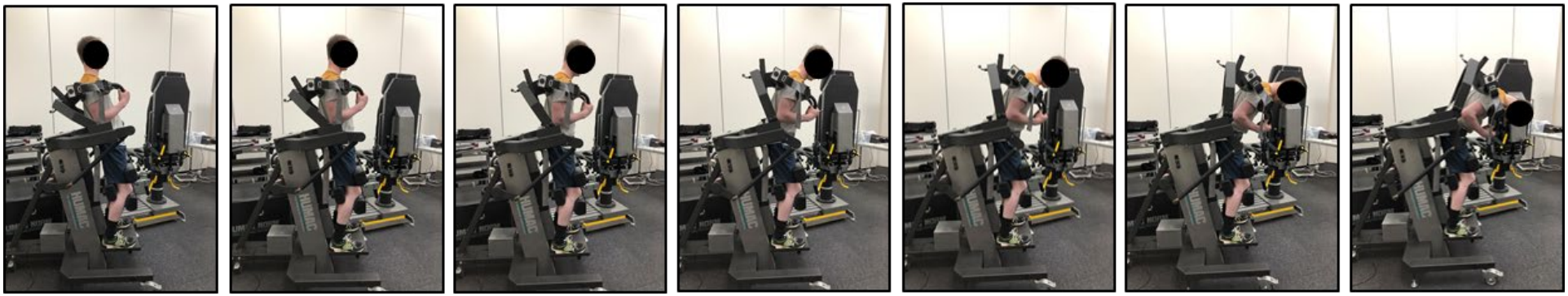


Figure 6.1 *Participant performing a concentric flexion trial on the TMC HUMAC® NORM™ system*

6.2.3 Familiarisation

Practice-based improvement is a known source of error that affects the reliability of repeated isokinetic dynamometer tests due to a learning effect. To attenuate these detrimental effects, familiarisation sessions were undertaken prior to testing. Impellizzeri et al. (2008) and Nugent, Snodgrass and Callister (2015) suggest that one familiarisation session is sufficient to ensure consistent peak torque values for isokinetic strength testing in healthy individuals. García-Vaquero and colleagues (2016) support this, recommending that reliable isokinetic trunk strength data can be obtained from one session. Therefore, participants performed a familiarisation session at least 10 days before testing to ensure adequate recovery. Participants were instructed to perform the familiarisation session at sub-maximal effort, approximately 50 % maximal voluntary contraction (MVC), to prevent excessive muscle damage (Deschenes et al., 2000). In accordance with previous studies (Ly and Handelsman, 2002), familiarisation was considered complete when participants were confident in performing the trials consistently for each condition. Confirmation was sought by visually inspecting torque-time graphs, where participants were able to successfully perform three sub-maximal consecutive contractions.

6.2.4 Test Protocol

Participants abstained from caffeine ingestion on the day of testing and from undertaking strenuous PA within seven days of testing. A five-minute warm-up on a cycle ergometer (Wattbike Ltd, Nottingham, UK) against low resistance (target power = 50 W; cadence = 60-80 rpm) was completed before participants performed a series of sub-maximal concentric flexion/extension contractions on the TMC through a full ROM to specifically target the trunk musculature. During these sub-maximalefforts, the testing ROM was determined by reducing the participant's maximum ROM by 10° from maximum extension and flexion (YG ROM = 99.6 ± 5.9°; OG ROM = 91.1 ± 8.5°; $t(22) = 2.85$, $p = .009$) to minimise the injury risk (Page, 2012) and allow sufficient force production to initiate the movement during eccentric contraction trials.

Prior to each test condition, participants performed five sub-maximal efforts (≈ 50 %) that replicated the test. This approach ensured sufficient preparation and correct performance whilst serving as another familiarisation to minimise learning effects (Johnson and Siegel, 1978; Nugent, Snodgrass and Callister, 2015). Following the warm-up trials participants rested for as long as required until they felt fully recovered and prepared for the three reciprocal flexion and extension MVCs. Similar protocols have been adopted previously for isokinetic strength testing of the trunk (Melo Filho, Eduardo and Moser, 2014; Karataş, Göğüş and Meray, 2002; Cramer et al., 2017; Holt et al., 2016) and lower limbs

(Segal et al., 2010). During the measurement verbal encouragement was given to facilitate maximal voluntary efforts (Matheson et al., 1992). The test conditions were performed in the order shown in **Table 6.4**. Contractions at slower angular velocities were tested first to increase the reproducibility of results between conditions (Wilhite, Cohen and Wilhite, 1992; Karataş, Göğüş and Meray, 2002). Previous studies have used a similar range of angular velocities to assess trunk muscle strength (Matheson et al., 1992; Bayramoğlu et al., 2001; Wang et al., 2017; Gabr and Eweda, 2019; Cramer et al., 2017).

Table 6.4 *Isokinetic dynamometry protocol*

Test Order	Condition	Protocol		
1	15°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
2	15°·s^{-1} Ecc/Ecc	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
3	30°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
4	30°·s^{-1} Ecc/Ecc	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
5	45°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
6	45°·s^{-1} Ecc/Ecc	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
7	60°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
8	60°·s^{-1} Ecc/Ecc	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
9	90°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
10	120°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials
	REST (min 60 secs)			
11	180°·s^{-1} Con/Con	Warm-up trials	Rest	Test trials

Con = Concentric contraction; Ecc = Eccentric contraction; warm-up trials consisted of 5 reciprocal sub-maximal (~50% MVC) repetitions; the rest period after warm-up trials was not limited; test trials consisted of 3 reciprocal MVC repetitions with 5 second pauses between consecutive movements; the rest period following test trials had no maximum limit.

To avoid inflated concentric torques augmented by preceding eccentric contractions (Finni et al., 2003; Herzog et al., 2016), reciprocal muscle groups were paired (i.e. extensor contraction followed by flexor contraction) with inter-contraction pauses of at least five seconds. Other studies have used short pauses between consecutive contractions (Ripamonti et al., 2009; Dvir and Keating, 2001), although the rest times were between one and three seconds. Due to the strenuous nature of the slower angular velocity contractions, longer pause times were given between consecutive contractions to ensure the participants' safety and recovery. To prevent the cumulative effects of fatigue influencing maximum muscle force generation (Sparto and Parnianpour, 1998; Nocella et al., 2011; Thomas and Raymond, 2000), adequate rest was given between trials. 30 seconds rest has been shown to provide sufficient recovery between isokinetic test conditions in older adult populations (Bottaro, Russo and Jacó De Oliveira, 2005), whilst others recommend a rest period of at least 60 seconds (Parcell et al., 2002). As a precaution, a minimum of 60 seconds rest time was given between test conditions. If the participant required more rest time, this period was extended until the sensation of fatigue abated. A maximum rest time was not prescribed due to the individualised recovery response to fatigue, especially between older and younger adults (Wang-Price et al., 2017; Solianik et al., 2017; Hautala et al., 2006).

6.2.5 Data processing

Torque, angular velocity and trunk angle data were acquired at a sampling rate of 100 Hz. The analogue torque signal from the dynamometer was filtered and digitised by the system's Digital Signal Processor (CYBEX, 1995). For each test condition, the contraction with the greatest peak torque was used for analysis. Peak torque values were identified during the isokinetic phase of the movement (**Figure 6.2**). Sagittal plane trunk strength outcomes are potentially affected by an initial spike in torque output followed by oscillations which appear in the initial part of the movement (Baltzopoulos and Brodie, 1989; Baltzopoulos, 2008; Ayers and Pollock, 1999; Perrin, 1993; Bemben, Grump and Massey, 1988). This limitation, known as torque overshoot, occurs due to the resistive force exerted by the dynamometer to decelerate the body segment to the pre-set angular velocity. As the limb accelerates beyond the pre-set velocity, the torque overshoot represents the torque required by the dynamometer to decelerate the limb-lever system (Baltzopoulos and Brodie, 1989; Baltzopoulos, 2008; Sapega et al., 1982), not originating from prime movers alone (Guilhem et al., 2014). Therefore, data which were not within 5% of the target velocity were discarded (Baltzopoulos, 2008). In addition, the first 20 consecutive data points that fell within the target velocity limits signified the start of the isokinetic phase. These constraints were designed to remove artefacts associated with torque

overshoot and impacts at the start and end of the movement. Torque values were normalised to body mass.

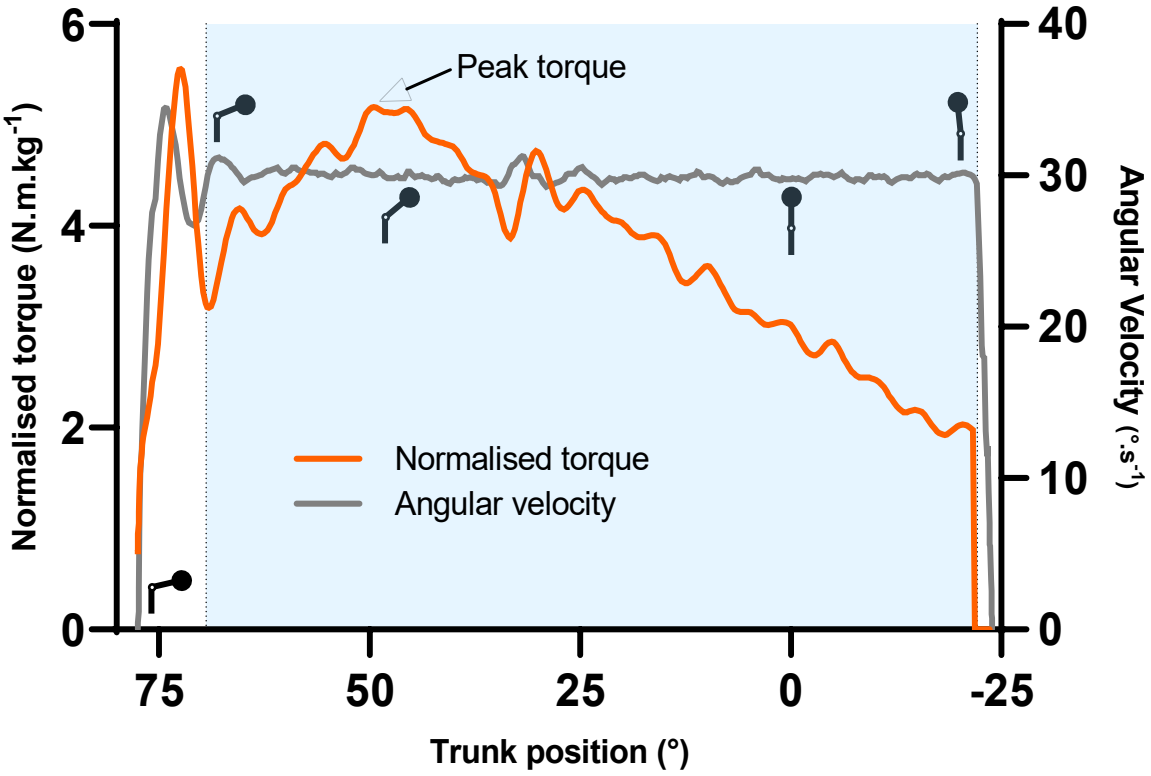


Figure 6.2 *Analysis of a concentric extensor trial performed at 30°·s^{-1}. The highlighted area represents the isokinetic phase of the movement*

6.2.6 Statistical Analysis

Statistical analyses were performed using SPSS (SPSS® for Windows Version 24.0, IBM Corp, Armonk, New York) and graphical presentation performed using GraphPad Prism (Version 8.3.1, San Diego, California). Data are presented as means with standard deviations (mean ± SD) unless otherwise stated. Independent samples t-tests were performed to compare statistical differences in trunk range of movement between the OG and YG. For the isokinetic data, two-way mixed-effects ANCOVA (angular velocity x age group) controlling for VPA were performed to compare mean differences in peak torque between the OG and YG. Concentric and eccentric conditions for the extensors and flexors were analysed separately. Following a significant interaction or main effect, multiple univariate ANCOVA with Bonferroni adjustments were performed to assess significant differences between age groups for each test condition. In accordance with Huberty and Morris (1989) and Huberty and Petoskey (2000), multiple ANCOVA were conducted rather than initially conducting a multivariate analysis of covariance (MANCOVA). MANCOVA do not necessarily control for Familywise Type I error

probability and are inappropriate as a preliminary step to multiple ANCOVA (Huberty and Morris, 1989; Huberty and Petoskey, 2000). Multiple linear regression analysis with stepwise elimination was also performed to explore the influence of LPM morphology and age on trunk extensor strength. Peak concentric and eccentric extensor torques at each angular velocity were the dependent variables. Nine independent variables were tested in the regression model: NMV and MFI of the PS, QL, ES and MF; and age group where the YG was coded as 0 and the OG coded as 1. An alpha level of 0.05 was required for statistical significance. Standardised effect size (η_p^2) and observed power (1-β) were also determined for each comparison where appropriate. Data for all conditions were normally distributed (Shapiro-Wilk test, $p > .05$) and homogeneous variances were assumed (Levene's test, $p > .05$). The assumption of sphericity was violated (Mauchly's test < 0.05). Therefore, Greenhouse-Geisser corrections were adopted.

6.2.6.1 Reliability

A sub-sample (n = 10) composed of participants from the YG (n = 5) and OG (n = 5) repeated the test protocol after 16 weeks to assess long-term intra-operator reliability. For each test condition, intra-class correlation coefficients (ICC) for peak torque were calculated using single-measurement, absolute-agreement, two-way mixed-effects models. ICC values less than 0.5 were considered indicative of poor reliability, values between 0.5 – 0.75 indicated moderate reliability, values between 0.75 – 0.9 indicated good reliability and values greater than 0.9 were considered indicative of excellent reliability (Koo and Li, 2016). Linear regression (difference vs mean) was also used to determine the existence of proportional bias for each test condition ($p \leq .05$).

Test-retest reliability was good to excellent across the range of test conditions for the concentric extensor trials. Test-retest reliability was moderate to excellent for the concentric flexor trials (ICC = 0.60 – 0.92), good to excellent for the eccentric extensor trials (ICC = 0.78 – 0.92) and moderate to good for the eccentric flexor trials (ICC = 0.72 – 0.86) (**Table 6.5**). For every test condition, regression coefficients were not significant ($p > .05$), indicating that proportional bias was not present.

Table 6.5 *Intraclass correlation coefficients (ICC) with 95% confidence intervals (CI) for test-retest reliability of each isokinetic test condition. Data are presented as ICC [95% CI]*

Angular velocity (°s^{-1})	Concentric Extension	Concentric Flexion	Eccentric Flexion	Eccentric Extension
15	.881 [.599, .969]	.897 [.642, .973]	.910 [.682, .977]	.720 [.225, .922]
30	.922 [.730, .980]	.635 [.048, .895]	.889 [.562, .972]	.718 [.243, .920]
45	.899 [.665, .973]	.813 [.397, .950]	.917 [.365, .983]	.861 [.532, .964]
60	.966 [.871, .992]	.693 [.180, .913]	.780 [.356, .940]	.813 [.410, .950]
90	.859 [.527, .963]	.599 [.034, .884]		
120	.953 [.829, .988]	.916 [.706, .978]		
180	.976 [.912, .994]	.645 [.107, .896]		

6.3 Results

Mixed two-way ANCOVA revealed a significant interaction between angular velocity x age group (F(3.3,69.8) = 3.2, p = .026) for concentric contractions of the trunk extensor muscles after controlling for VPA (F(1,21) = 0.32, p = .581). Significant main effects for age group (F(1,21) = 19.9, p < .001) and angular velocity (F(3.3,69.8) = 3.6, p = .015) were also revealed, showing that the YG produced greater peak concentric extension torque (4.64 N·m·kg^{-1}) than the OG (3.04 N·m·kg^{-1}) and that both groups showed a general decline in peak concentric extension torque with increasing angular velocity. No significant interactions or main effects were observed for any other condition. ALM was not a significant covariate of trunk extensor concentric strength between groups (F(1,21) = 0.03, p = .866). Peak torque data are presented in **Table 6.6**.

Table 6.6 *Peak torque normalised to body mass for each isokinetic test condition*

Angular velocity (°·s^{-1})	Concentric extension N·m·kg^{-1}		Concentric flexion N·m·kg^{-1}		Eccentric flexion N·m·kg^{-1}		Eccentric extension N·m·kg^{-1}	
	Young	Old	Young	Old	Young	Old	Young	Old
15	***5.11 ± 0.89***	***3.71 ± 0.71***	2.83 ± 0.33	2.68 ± 0.48	5.80 ± 1.06	5.26 ± 0.73	3.35 ± 0.36	3.28 ± 0.62
30	***4.92 ± 0.94***	***3.58 ± 0.73***	2.89 ± 0.38	2.60 ± 0.45	6.12 ± 0.99	5.29 ± 1.05	3.42 ± 0.35	3.34 ± 0.80
45	***4.67 ± 0.80***	***3.47 ± 0.80***	2.86 ± 0.40	2.70 ± 0.55	6.07 ± 1.16	5.30 ± 0.97	3.36 ± 0.39	3.39 ± 0.64
60	***4.83 ± 0.93***	***3.39 ± 0.90***	2.97 ± 0.48	2.70 ± 0.63	5.75 ± 0.96	4.98 ± 1.02	3.35 ± 0.42	3.29 ± 0.62
90	***4.86 ± 1.15***	***2.99 ± 0.77***	3.02 ± 0.37	2.78 ± 0.71				
120	***4.49 ± 1.03***	***2.60 ± 0.95***	2.77 ± 0.58	2.57 ± 0.83				
180	***3.62 ± 1.19***	***1.55 ± 0.81***	2.14 ± 0.70	1.81 ± 0.86				

Note: Bold italics denote significant difference between groups

6.3.1 Age-related Differences in Peak Concentric Extensor Torque

One-way univariate ANCOVA (Bonferroni adjustment) revealed significant differences between the OG and YG at all angular velocities for concentric contractions of the trunk extensors. At

- 15°.s^{-1} (F(1,21) = 14.0, p = .001, η_p^2 = 0.399, 1-β = 0.945); Cohen's d = 1.74
- 30°.s^{-1} (F(1,21) = 13.0, p = .002, η_p^2 = 0.382, 1-β = 0.930); Cohen's d = 1.59
- 45°.s^{-1} (F(1,21) = 12.0, p = .002, η_p^2 = 0.364, 1-β = 0.911); Cohen's d = 1.50
- 60°.s^{-1} (F(1,21) = 14.0, p = .001, η_p^2 = 0.399, 1-β = 0.945); Cohen's d = 1.57
- 90°.s^{-1} (F(1,21) = 20.2, p < .001, η_p^2 = 0.491, 1-β = 0.990); Cohen's d = 1.91
- 120°.s^{-1} (F(1,21) = 20.9, p < .001, η_p^2 = 0.499, 1-β = 0.992); Cohen's d = 1.91
- 180°.s^{-1} (F(1,21) = 19.0, p < .001, η_p^2 = 0.475, 1-β = 0.986); Cohen's d = 2.03

the YG produced significantly greater peak concentric extensor torque than the OG. Peak concentric extensor torque was generally consistent from 15°·s^{-1} to 60°·s^{-1} for both groups. The OG exhibited decrements thereafter, whilst the YG showed declines from 120°·s^{-1}. This resulted in a trend for increasing difference between OG and YG as angular velocity increased past 45°·s^{-1} (**Figure 6.3**). Significant pairwise differences (p < .001) in concentric extension peak torque were found between 120°·s^{-1} and 180°·s^{-1} and between each of these conditions with all other angular velocities.

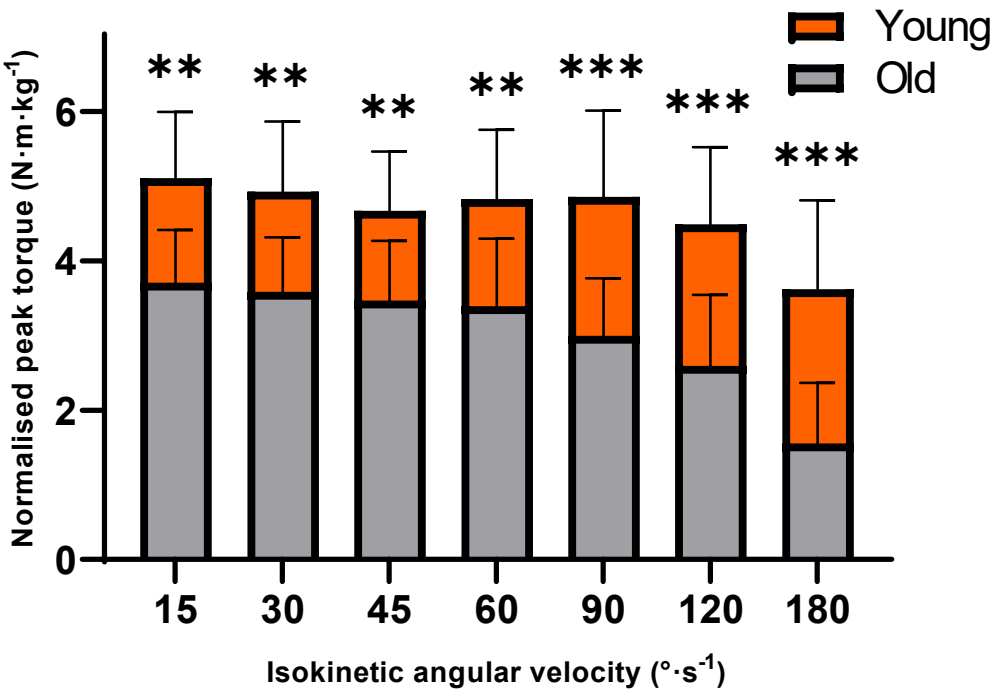


Figure 6.3 *Peak torque normalised to body mass produced for concentric contractions of the trunk extensor muscles. Error bars represent standard error of the mean. ** < .01, *** < .001*

6.3.2 Moderating Effect of Muscle Morphology

Across all concentric extensor conditions, only age group was a significant moderator of trunk strength. At the faster angular velocities (i.e. 90, 120 and 180°·s^{-1}), the moderating effect of age was greater. Age group explained the largest amount of variance (R^2 = 52.8%) in peak concentric extension torque at 180°·s^{-1}, $F(1,22) = 24.65$, $p < .001$. Predicted peak concentric trunk extensor torque at 180°·s^{-1} was equal to 3.620 – 2.066*(age group). Of the potential variables included in the multiple linear regressions for eccentric strength, muscle morphology variables were the only significant moderators. No variables were significant predictors of eccentric strength at 15°·s^{-1}. At 30°·s^{-1}, ES NMV and QL MFI were significantly associated with peak eccentric extensor torque, $F(2,21) = 5.52$, $p = .012$, R^2 = 34.5%. Predicted peak eccentric trunk extensor torque at 30°·s^{-1} was equal to 4.332 + 0.322*(ES NMV) – 0.083*(QL MFI). QL MFI remained the only significant moderator of eccentric extensor strength at 45°·s^{-1} and 60°·s^{-1} (**Table 6.7**).

Table 6.7 *Multiple Linear Regression Analysis with Stepwise Elimination*

Moderators	Peak Concentric Torque (Nm·kg^{-1})							Peak Eccentric Torque (Nm·kg^{-1})			
	15°·s^{-1}	30°·s^{-1}	45°·s^{-1}	60°·s^{-1}	90°·s^{-1}	120°·s^{-1}	180°·s^{-1}	15°·s^{-1}	30°·s^{-1}	45°·s^{-1}	60°·s^{-1}
Intercept	5.105 (< .001)	4.924 (< .001)	4.667 (< .001)	4.829 (< .001)	4.857 (< .001)	4.491 (< .001)	3.620 (< .001)	5.525 (< .001)	4.332 (.007)	7.546 (< .001)	7.462 (< .001)
Age Group	-1.400 (< .001)	-1.341 (< .001)	-1.198 (.001)	-1.438 (.001)	-1.863 (< .001)	-1.895 (< .001)	-2.066 (< .001)	X	X	X	X
PS MFI	X	X	X	X	X	X	X	X	X	X	X
QL MFI	X	X	X	X	X	X	X	X	-0.083 (.035)	-0.090 (.039)	-0.102 (.011)
ES MFI	X	X	X	X	X	X	X	X	X	X	X
MF MFI	X	X	X	X	X	X	X	X	X	X	X
PS NMV	X	X	X	X	X	X	X	X	X	X	X
QL NMV	X	X	X	X	X	X	X	X	X	X	X
ES NMV	X	X	X	X	X	X	X	X	0.322 (.022)	X	X
MF NMV	X	X	X	X	X	X	X	X	X	X	X

Values are presented as unstandardised regression coefficients with significance level, B value (p value). X denotes removed non-significant variables. PS = psoas, QL = quadratus lumborum, ES = erector spinae, MF = multifidus, MFI = muscle fat infiltrate, NMV = normalised muscle volume

6.4 Discussion

The current study demonstrates for the first time that the normal loss of dynamic trunk muscle strength in older age is muscle and contractile mode specific. The main findings were that trunk extensor muscles experience an age-related decrement in concentric strength and the age-effect increases with increasing angular velocity, eccentric strength is somewhat preserved in the trunk flexors and extensors, and the loss of concentric torque in the trunk extensor muscles is not moderated by muscle specific morphology measures. A progressive decline in muscle strength typically accompanies the ageing process, however, normal age-related decrements in the lumbar musculature are not fully understood. Given the inconsistent methods and equivocal nature of findings on this topic, this work provides an in-depth investigation into age-related differences in trunk strength and constitutes an important contribution to the literature base to date.

6.4.1 Loss of Extensor Concentric Torque

The results show that healthy men experience a loss of concentric extensor torque in older age in the trunk. The effect of age was large for all concentric extension conditions and is supported by other researchers reporting large effect sizes for the loss of concentric extensor strength with age in healthy men (Danneskiold-Samsøe et al., 2009; Hasue, Fujiwara and Kikuchi, 1980). As the proportion of contractile tissue in the LPMs decreases due to age-related atrophy and fat infiltration (see Chapters 4 and 5), the muscles' capacity to generate force and perform work is reduced (Ropponen, Videman and Battié, 2008). However, after the effect of age group on concentric extensor torque had been partialled out, the results of this study indicated that the loss of LPM volume and increased fat infiltration were unable to explain reductions in concentric extension strength in the trunk. This raises questions about how useful morphological measures in the LPMs are as they appear to have limited bearing on dynamic muscle function in healthy older men. Neurological changes in older age are more likely to have contributed to declines in trunk strength.

Neuropathic processes in older age bring about a decline in neural drive and cause muscle to express a slower phenotype (Campbell, McComas and Petito, 1973; Evans and Lexell, 1995; Häkkinen et al., 1996; Roos et al., 1997; Mitchell et al., 2012; Unhjem et al., 2015). The increasing disparity between groups in concentric extension torque with increasing angular velocity (**Figure 6.3**) suggests a shift towards a slower fibre-type composition in the OG. Indeed, muscles in the lumbar spine have a propensity towards slower isoforms (Regev et al., 2010). Given that the intrinsic strength of type I fibres is less than type II fibres (Bottinelli et al., 1996; Young, 1984), an increasing proportion of slow-twitch fibres is likely to reduce muscular force production resulting in a loss of concentric extension

strength (Ivy et al., 1981; Robles et al., 2015). This age-related remodelling of muscle phenotype may explain why loss of concentric extensor torque is greater with increasing angular velocity and more pronounced in the OG.

Motor unit remodelling is likely to contribute to this age-related shift towards slower muscle phenotypes. As the number of functioning motor units decreases with ageing, with a preferential loss of fast motor units, older adults experience a loss of muscle strength. Some of the denerved fibres are reinnervated by remaining motor units, resulting in a net conversion of faster glycotic muscle fibres into slow oxidative muscle fibres (Lexell, Downham and Sjöström, 1986). This shift towards a slower muscle phenotype compromises the force generating capacity of the muscle with a greater impact on faster movements, demonstrated by the disproportionately greater loss of muscle power than strength in healthy older adults (Skelton et al., 1994). This is reflected in current results, as concentric extensor torque loss was greater at higher movement speeds in the OG compared to the YG. In addition, increased co-activation of antagonist muscles has been implicated as another neuromuscular mechanism that contributes to lower force output in older adults (Macaluso et al., 2002; Bautmans et al., 2011).

6.4.2 Attenuation of Flexor Concentric Torque

The OG exhibited lower concentric flexion torque, although differences with the YG did not reach significance for any of the angular velocities. Similar findings have been previously reported (Smith et al., 1985), although not undisputed (Hasue, Fujiwara and Kikuchi, 1980; Skrzek and Bolanowski, 2006). Hasue, Fujiwara and Kikuchi (1980) suggested that the discrepancy in abdominal and paravertebral muscle strength may be due to the constant use of antigravity muscles in daily life whereas intra-abdominal pressure aids the function of the abdominal muscles. It is also likely that the apparent attenuation in trunk flexion strength is the result of abdominal morphometry preservation. Whilst the relative degeneration of the abdominal muscles compared to the paravertebral muscles cannot be determined, it has been shown that the abdominals are relatively spared from the effects of age-related degeneration compared to the paravertebral muscles (Valentin, Licka and Elliott, 2015; Meakin et al., 2013). This may preserve the contractile unit of the abdominal muscles relative to the paravertebral muscles, which may explain why concentric flexion torque was not significantly different between the OG and YG whilst concentric extension torque was.

6.4.3 Preservation of Eccentric Strength

This study found that both extensor and flexor muscle groups in the trunk experience a relative preservation of eccentric strength in older age. To the author's knowledge, this is the first study to investigate age-related differences in eccentric trunk strength, which precludes comparisons with other studies. Age-related preservation of eccentric strength has been observed in other muscle groups (Klass, Baudry and Duchateau, 2005; Poulin et al., 1992) although no mechanisms have been fully accepted (Hortobágyi et al., 1995; Roig et al., 2010). Compared to concentric contractions, muscle exhibits significantly lower neural activation at a given force output during eccentric contractions (Kellis and Baltzopoulos, 1998). Concentric contractions are also affected by increased antagonist coactivation in older age (Larsen et al., 2008; Macaluso et al., 2002) whilst the effect is diminished in eccentric contractions (Kellis and Baltzopoulos, 1999). Therefore, age-related deficits in neural drive (Unhjem et al., 2015; Häkkinen et al., 1996; Roos et al., 1997) are likely to have greater impact on concentric contractions than eccentric. In the current study, lean trunk mass was unable to explain the preservation of eccentric strength. This supports Hortobágyi et al. (1995) who state that eccentric strength is maintained independent of age-related morphometric muscle changes. However, pathways may differ between the abdominal and paravertebral musculature. The current findings demonstrate that preservation of eccentric extensor strength in older age is dependent upon changes in muscle morphology, particularly fat infiltration within the QL muscle.

Alterations in the passive structural elements and intrinsic factors associated with cross-bridge cycling may also mediate the relative preservation of eccentric strength (Hill et al., 2019; Power, Rice and Vandervoort, 2012; Herzog, 2014). According to Lombardi and Piazzesi (1990), the stretch produced during eccentric contractions may shift myosin heads into a strongly bound state. This would subsequently reduce the age-related deficits in force output during eccentric contractions that are commonly observed during concentric contractions (Phillips, Bruce and Woledge, 1991). However, it is still unknown why this phenomenon does not induce additional tension in the muscle fibres of younger adults. One explanation concerns the reduced speed of cross-bridge cycling. In older adults, the slower detachment rate of active cross bridges in muscle fibres (Larsson, Li and Frontera, 1997) may contribute to the preservation of eccentric strength (Ochala et al., 2006). As a result of strong cross-bridge binding, a configurational change of troponin and tropomyosin occurs that makes attachment sites available for titin on actin (Herzog, 2019). Once bound to actin, titin becomes stiffer due to a shorter free spring length and thus increases force output when a muscle is stretched (Herzog, 2014). Upon deactivation, titin remains bound to actin but can be dissociated immediately when the muscle shortens quickly (Herzog, 2019). As older muscle fibres exhibit slower detachment rates and increased instantaneous stiffness following stretch (Ochala et al., 2006), there may be elevated levels

of residual force enhancement during eccentric contractions (Power, Rice and Vandervoort, 2012). Furthermore, as eccentric movements are performed faster, there may not be sufficient time for the cross-bridges to dissociate in older adults, thus preserving the residual force enhancement. This is reflected in the current findings as torque did not appear to decline with increasing angular velocity.

Mechanisms regarding preservation of eccentric strength in older age focus on neurological, cellular and mechanical pathways (Roig et al., 2010; Hortobágyi et al., 1995), however, none consider biomechanical function, especially relating to the trunk. Thoracolumbar bending moment increases with ageing due to postural changes (Le Huec et al., 2018). The extensor muscles are subsequently activated to prevent forward flexion of the trunk (Waters and Morris, 1972; Cresswell, Oddsson and Thorstensson, 1994), which increases mechanical energy expenditure required for eccentric control of the lower trunk musculature (McGibbon and Krebs, 2001). Indeed, eccentric muscle contractions are inherently common during ADLs (Dickinson et al., 2000) to decelerate movements and store elastic recoil energy (LaStayo et al., 2003). This type of muscle contraction is highly important to most trunk movements during ADLs; evidenced during sit to stand preparation when the ES eccentrically contract to provide postural stability and movement control (Millington, Myklebust and Shambes, 1992; Dubost et al., 2005; Silva et al., 2015). Despite the low-level activity of trunk muscles during ADLs (McGill and Cholewicki, 2001), sustained low-intensity eccentric activation may provide enough stimulus for the muscles to maintain their strength. Even at low levels of exertion, eccentric resistance exercises can produce relatively large muscle workload (Lim, 2016) and improve strength in older adults (Chen et al., 2017). Given that kinematic changes in older age increase mechanical energy expenditure required for eccentric control of the LPMs during gait (McGibbon and Krebs, 2001), postural changes may inadvertently offset reductions in eccentric trunk strength. Whilst speculative, these suggestions are plausible and attempt to understand this phenomenon in a holistic manner. More importantly, the current results in the trunk reflect the eccentric strength age-response observed in the appendicular muscles (Klass, Baudry and Duchateau, 2005; Poulin et al., 1992). This suggests that eccentric strength preservation is systemic rather than a muscle- or site-specific phenomenon in the body.

6.4.4 Moderating Effect of Physical Activity

PA is generally believed to have a positive effect on muscular strength in older adults (Rantanen et al., 2016). Whilst VPA is more beneficial, low-intensity PA can still lead to better functional ability amongst older adults (Avlund et al., 1994). However, the results of this study suggest that habitual VPA does not moderate age-related differences in trunk strength amongst healthy men. Although this may seem

counterintuitive, previous research in a large community-dwelling population supports this finding (Viljanen, Viitasalo and Kujala, 1991). It was suggested that the small proportion of adults engaging regularly in resistance training (< 1 %) may have been insufficient to observe a training effect on maximal isometric trunk strength in their sample (Viljanen, Viitasalo and Kujala, 1991). In the current study, PA was measured using accelerometery and a recognised limitation of using accelerometers is their inability to detect non-ambulatory activities such as resistance exercise (Lee and Shiroma, 2014; Viljanen, Viitasalo and Kujala, 1991). Since resistance exercise is known to increase muscular strength (Yarasheski et al., 1995; Taaffe et al., 1999; Peterson et al., 2010), its potential omission from accelerometer data acquisition may have confounded the current findings.

6.4.5 Clinical and Practical Applications

In clinical settings, understanding the age-related loss of trunk strength could be crucial due to its association with physical function (Shahtahmassebi et al., 2017), lower back pain (Cho et al., 2014) and falls risk (Granacher et al., 2013). Whilst the rehabilitation of upper and lower limb muscles is often based on the relative strength of the unaffected limb, bilateral comparisons cannot be made in the trunk. Therefore, age-specific normative trunk strength values across a range of contraction types and angular velocities are needed to allow healthcare professionals to evaluate a patient's trunk strength and determine an effective rehabilitation intervention. Based on the current results, slower angular velocities than $60°\cdot s^{-1}$ may not provide additional information about the maximal force generating capacity of trunk muscles in healthy men. The substantial decline in concentric trunk extension torque from $90°\cdot s^{-1}$ to $180°\cdot s^{-1}$ indicates that investigation at greater angular velocities may be valuable. However, the range of conditions used in this study accounted for trunk activity typically observed during ADLs (Lindemann et al., 2014; Pigeon et al., 2003; Goutier et al., 2010). Faster conditions would represent more dynamic movements that may offer additional insight into injury mechanisms of the lumbar spine. Furthermore, these results should support the use of isokinetic testing in the trunk and establishment of population specific norms that could provide useful clinical guidelines for trunk assessment and rehabilitation.

6.4.6 Limitations

There were limitations in this study that should be acknowledged. Firstly, the samples comprised of healthy physically active men. Caution should be taken when generalising these findings as the participants, particularly in the OG, are unlikely to be representative of a general population. Strength values in the OG may be considerably greater than in a general population, which would suggest that

age-related loss of concentric strength in the LPMs is even more pronounced in general populations. Generalising the findings to female populations should also be done with caution, as women tend to show greater declines in trunk muscle strength with age (Lee et al., 2012; Keller et al., 1999; Danneskiold-Samsøe et al., 2009). Comparison with diseased populations may however provide useful information regarding pathological deviations in trunk strength. The results suggested that the eccentric tests and concentric flexion test were underpowered. However, the sample size was large enough to observe sufficient power for the concentric extension test. These results should be used to determine sample sizes in future studies. Furthermore, these findings are specific to the testing methodology. For example, this study was limited to sagittal plane movements. Movements in coronal and transverse planes may reveal different age effects. Testing conditions should be considered carefully when generalising these findings. Finally, whilst the current study highlights an important feature of age-related musculoskeletal decline, longitudinal studies are needed to infer causation.

6.5 Conclusion

This study indicates that ageing elicits a muscle and contractile mode specific response in isokinetic torque of the trunk muscles. Concentric extensor muscle strength declines in older age whilst eccentric trunk strength appears to be relatively preserved. Peak torque of the extensor muscles decreased with increasing angular velocity for concentric contractions and was more pronounced in the older group. As muscle morphology measures were not significant moderators of concentric trunk extension strength, the increasing disparity at greater angular velocities was likely due to age-related neuropathic processes affecting the intrinsic contractile function of the LPMs. VPA level did not moderate age-related differences in trunk strength, although this may be due to the way in which PA was measured. These findings are a useful step in establishing effective clinical and public health intervention strategies that could be used to offset adverse health outcomes related to trunk strength loss in older adult populations. Future research should look to assess trunk strength in a range of populations using a longitudinal design, which may enable identification of pathological deviations. Furthermore, there is a need to understand the consequence of these changes in relation to ADLs such as walking gait.

Table 6.8 Thesis Map

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance
		Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men	• Studies investigating muscle degeneration with ageing have typically focused on the appendicular muscles • There is increasing recognition for the importance of the lumbar paravertebral muscles in maintaining health and mobility in older age	Aims	• To investigate age-related differences in LPM morphology • A secondary aim was to investigate the age-response on fat infiltration and volume of the different lumbar muscles (i.e. MF, ES, QL and PS) • An additional aim was to explore other predictors of lumbar paravertebral muscle degeneration

	• Few studies have characterised features of age-related degeneration in the lumbar musculature • Few studies have provided volumetric information on all of the paravertebral muscles using high-resolution imaging modalities	Key findings	• Older age negatively affected all paravertebral muscles, although some showed greater degenerative changes than others • Age-related fat infiltration has a global effect across the lumbar musculature, whereas atrophic changes appear to be muscle-specific • Only the QL and ES showed significant age-related declines in muscle volume • All muscles showed age-related declines in muscle quality (i.e. increase in intramuscular adipose tissue) • The MF was most susceptible to compositional changes with age, whilst the QL was most vulnerable to reductions in muscle volume • Physical activity did not influence age-related differences in muscle degeneration in the lumbar spine • Non-dominant handgrip strength was a predictor of muscle atrophy in the lumbar musculature
		Implications	• The QL and ES appear to be most affected in older agesince they exhibited declines in size and quality • When investigating the effects of ageing on lumbar muscle function, macroscopic changes in the paravertebral muscles should be considered • Structural changes, resulting in a loss of contractile tissue, may reduce muscle function in the lumbar spine • Convenient and easily administered measures such as handgrip strength may be able to predict muscle atrophy in the lumbar spine
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men	• Dynamic trunk strength in older adults has not been fully explored • Studies have typically investigated age-related strength loss using handgrip dynamometry or lower limb isokinetic dynamometry • Majority of studies have used clinical assessments which may not be appropriate to assess maximal trunk strength • No study has assessed eccentric trunk strength in older adults and contractile modes are typically limited • The findings from chapter 5 have also influenced the need for this study. Research investigating how muscle	Aims	• To investigate age-related differences in dynamic trunk strength • The secondary aim was to explore the moderating effect of muscle morphology degeneration on extensor muscle strength
		Key findings	• Age had a significant and negative effect on peak concentric trunk extensor torque across all angular velocities • The difference in concentric extensor torque between the older and younger group increased with increasing angular velocity indicating that the lumbar extensor muscles express a slower phenotype with ageing • Peak concentric torque of the trunk flexor muscles decreases in older age but not significantly • Peak eccentric torque of the extensors and flexors in the trunk is preserved in older age

	morphology degeneration in the lumbar spine impacts on trunk extensor strength is warranted		• Concentric strength of the trunk extensor muscles is negatively associated with age, but not paravertebral muscle morphology • Eccentric strength of the trunk is primarily related to quadratus lumborum muscle quality, but not age
		Implications	• Loss of trunk strength in older age is contractile mode- and muscle- specific • Training interventions should target the extensor trunk muscles using concentric exercises to improve strength in older adults • Improving paravertebral muscle quality may further preserve eccentric strength of the trunk extensors • Internal trunk moments produced during daily tasks should be combined with the peak values measured in this study to determine how functionally demanding these tasks are on the trunk musculature
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men		Aim	
		Key findings	
		Implications	

Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men

Chapter Abstract

Background The trunk plays an important role in mobility and providing postural support during everyday activities, such as walking. Gait analysis has tended to focus on the lower limbs. Therefore, kinematic and kinetic age-related changes in the trunk are not fully understood. The aim of this study was to investigate the effect of age on biomechanical function of the trunk during normal gait in healthy young and older men.

Methods Three-dimensional motion analysis was used to determine spatiotemporal parameters as well as kinematic and kinetic variables of the trunk and pelvis in 12 healthy older (67.3 ± 6.0 years) and 12 healthy younger men (24.7 ± 3.1 years). All participants performed three successful gait trials. Kinematic and kinetic data were analysed in the sagittal, coronal and transverse planes. Functional demand (FD) of the trunk during the GC was calculated in the sagittal plane. Independent t-tests were performed to compare the effect of age on outcome parameters. To determine if age-related differences within the GC were phase-specific, statistical parametric mapping (SPM) was used to compare trunk kinematic and kinetic waveforms between the OG and YG. Zero-order and partial correlations, controlling for age group, were conducted to determine interplanar and intersegment relationships in ranges of trunk and pelvis motion.

Results Trunk and pelvic rotations in all planes of motion were reduced with age. Trunk kinematics with respect to the global reference frame were most affected by age in the transverse plane. In the pelvic reference frame, trunk flexion/extension ROM was reduced in the OG as well as peak movement amplitudes in the coronal plane. Walking speed was not significantly different between the OG and YG ($p > .05$), therefore age-related differences in trunk kinematics were not due to walking speed. Phase-specific differences were observed in the coronal and transverse planes with midstance and swing phases highlighted as the instances when trunk and pelvic kinematics differed between age groups. Controlling for age, fewer correlations were revealed between trunk and pelvic ROMs and between planes of motion, indicating that older age causes an uncoupling of interplanar upper body movements during gait. The YG performed significantly more negative work during the GC than the OG ($p = .023$) and exhibited a significantly greater power absorption peak in the coronal plane during swing phase ($p = .010$). Trunk moment and power waveforms were similar between the YG and OG.

Walking was approximately 20 % more functionally demanding on the trunk in the OG than the YG across the GC.

Conclusions Age-related differences in trunk kinematics were apparent in the OG, particularly in the coronal and transverse planes. However, changes in the pelvis were highly responsible for observed changes in the trunk with age. Trunk moments and powers in the sagittal and coronal planes may also be modified in older age, although these changes may not be phase specific. Age-related differences in biomechanical function of the trunk during walking may be indicative of a conservative gait strategy to reduce falls and injury risk at the cost of increasing energetic and functional demands. These findings may provide important information for rehabilitation programmes in older adults designed to improve trunk motion as well as enabling identification of higher risk movement patterns.

Key words: 3-D motion analysis, walking gait, trunk, kinematics, kinetics, functional demand

Table 7.1 *Thesis Map*

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		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance
		Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men	• Studies investigating muscle degeneration with ageing have typically focused on the appendicular muscles • There is increasing recognition for the importance of the lumbar paravertebral muscles in maintaining health and mobility in older age	Aims	• To investigate age-related differences in LPM morphology • A secondary aim was to investigate the age-response on fat infiltration and volume of the different lumbar muscles (i.e. MF, ES, QL and PS) • An additional aim was to explore other predictors of lumbar paravertebral muscle degeneration

	• Few studies have characterised features of age-related degeneration in the lumbar musculature • Few studies have provided volumetric information on all of the paravertebral muscles using high-resolution imaging modalities	Key findings	• Older age negatively affected all paravertebral muscles, although some showed greater degenerative changes than others • Age-related fat infiltration has a global effect across the lumbar musculature, whereas atrophic changes appear to be muscle-specific • Only the QL and ES showed significant age-related declines in muscle volume • All muscles showed age-related declines in muscle quality (i.e. increase in intramuscular adipose tissue) • The MF was most susceptible to compositional changes with age, whilst the QL was most vulnerable to reductions in muscle volume • Physical activity did not influence age-related differences in muscle degeneration in the lumbar spine • Non-dominant handgrip strength was a predictor of muscle atrophy in the lumbar musculature
		Implications	• The QL and ES appear to be most affected in older age since they exhibited declines in size and quality • When investigating the effects of ageing on lumbar muscle function, macroscopic changes in the paravertebral muscles should be considered • Structural changes, resulting in a loss of contractile tissue, may reduce muscle function in the lumbar spine • Convenient and easily administered measures such as handgrip strength may be able to predict muscle atrophy in the lumbar spine
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men	• Dynamic trunk strength in older adults has not been fully explored • Studies have typically investigated age-related strength loss using handgrip dynamometry or lower limb isokinetic dynamometry • Majority of studies have used clinical assessments which may not be appropriate to assess maximal trunk strength • No study has assessed eccentric trunk strength in older adults and contractile modes are typically limited • The findings from chapter 5 have also influenced the need for this study. Research investigating how muscle morphology degeneration in the lumbar spine impacts on trunk extensor strength is warranted	Aims	• To investigate age-related differences in dynamic trunk strength • The secondary aim was to explore the moderating effect of muscle morphology degeneration on extensor muscle strength
		Key findings	• Age had a significant and negative effect on peak concentric trunk extensor torque across all angular velocities • The difference in concentric extensor torque between the older and younger group increased with increasing angular velocity indicating that the lumbar extensor muscles express a slower phenotype with ageing • Peak concentric torque of the trunk flexor muscles decreases in older age but not significantly • Peak eccentric torque of the extensors and flexors in the trunk is preserved in older age • Concentric strength of the trunk extensor muscles is negatively associated with age, but not paravertebral muscle morphology • Eccentric strength of the trunk is primarily related to quadratus lumborum muscle quality, but not age
		Implications	• Loss of trunk strength in older age is contractile mode- and muscle- specific • Training interventions should target the extensor trunk muscles using concentric exercises to improve strength in older adults • Improving paravertebral muscle quality may further preserve eccentric strength of the trunk extensors

			• Internal trunk moments produced during daily tasks should be combined with the peak values measured in this study to determine how functionally demanding these tasks are on the trunk musculature
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men	• Studies investigating the relationship between ageing of the lumbar spine and loss of physical function have typically used clinical assessments • Clinical assessments and performance batteries are not specific to the lumbar spine • Few studies have investigated age-related changes in trunk kinematics during gait, and fewer still have investigated kinetic changes in older age. Therefore, the effects of age on trunk movements and kinetics during gait are not well known • Functional demand is a measure that has been applied to the lower limbs to investigate how biomechanically demanding everyday activities are. However, functional demand has never been applied to the trunk • There is a need to understand how biomechanical function of the lumbar spine is related to muscle morphology degeneration and strength loss in older age	Aims	• To investigate age-related differences in trunk biomechanics during normal walking gait • A secondary aim was to determine the functional demand of the trunk during normal walking and investigate how it is affected in older age • A further aim was to investigate the relationship between morphological degeneration of the lumbar musculature and biomechanical outcomes
		Key findings	
		Implications	

denotes links to previous chapters. Links to chapter 3 – VPA was included as a potential covariate. Links to chapter 5 – Muscle morphology measures were included as potential covariates to assess whether age-related differences in biomechanical trunk function covaried with atrophy or fat infiltration of the paravertebral musculature. Links to chapter 6 – Isokinetic strength measures were combined with functional moments during gait to calculate how functionally demanding walking was in the trunk

CHAPTER 7

7.1 Introduction

Walking, the most common form of exercise amongst older adults (Sallis et al., 1986; Lawlor et al., 2002; Walsh et al., 2001; Shephard, 2003), is an important daily task that requires synchronised actions of the musculoskeletal system to function independently. Changes in gait, such as reduced walking speed, are a useful indicator of overall health status and are associated with adverse health outcomes such as increased falls risk and mortality in older adults (Maki, 1997; Ferrucci et al., 2000; Hausdorff, Rios and Edelberg, 2001; Studenski et al., 2003; Verghese et al., 2009, 2006, 2007; Studenski et al., 2011; Cesari et al., 2005). Indeed, low gait speed (≤ 0.8 $m \cdot s^{-1}$) is one criterion used to confirm a sarcopenia diagnosis (Cruz-Jentoft et al., 2019, 2010a). Spaciotemporal measures such as walking speed may not always be sufficient to explore the effects of ageing on gait (Stephan, Sutin and Terracciano, 2015). Biomechanical analysis however may be more useful in detecting subtle changes. For example, reduced plantarflexion power causes older adults to redistribute joint powers proximally (Judge et al., 1996; Cofré et al., 2011), which increases the metabolic cost of walking (Das Gupta, Bobbert and Kistemaker, 2019) due to involvement of greater muscle mass.

Whilst biomechanical function of the lower limbs is well understood with the ageing process, age-related changes in trunk biomechanics have not been established. Due to an ageing population (World Health Organization, 2011; Storey, 2018), understanding age-related changes in gait has become increasingly important and is the first step in devising effective strategies that preserve independence and quality of life in older adults. A greater understanding of the ageing process on trunk biomechanics during walking is therefore needed. Furthermore, in the previous chapters it was implied that morphological degeneration of the lumbar musculature and strength loss in the trunk affects physical function in older age. The need to relate these findings to biomechanical function of the trunk during a typical activity such as walking is great. Indeed, only one study (Shahtahmassebi et al., 2017) has attempted to understand these interdependencies but focused on physical performance rather than specific trunk measures. In the systematic review and meta-analysis this study was also judged as poor quality (**Table 4.5**). The findings from previous chapters has highlighted that high-quality research investigating the effect of ageing on trunk biomechanics during walking, as well as relationships with strength loss and morphological changes in the muscle, is warranted.

7.1.1 The Gait Cycle

Human gait is characterised by a complex bipedal locomotion pattern. Centre of gravity (COG) is maintained over a continually changing base of support (BOS) that alternates in cycles between single- and double-limb support (Harris et al., 2008). The GC (**Figure 7.1**) is defined by two consecutive

occurrences of initial contact (IC) by the same foot (ipsilateral limb) and divided into two phases (Neumann, 2017). The stance phase accounts for approximately 60% of the GC and defines the period of time when the foot is in contact with the ground. Stance phase is from IC to toe-off (TO) and functions to provide weight acceptance, vertical support and propulsion. Swing phase refers to when the foot is airborne, causing limb advancement. Accounting for approximately 40% of the GC, swing phase is initiated at TO and terminates at ipsilateral heel-strike (IC). Stance and swing phases are sub-divided to identify specific periods during the GC.

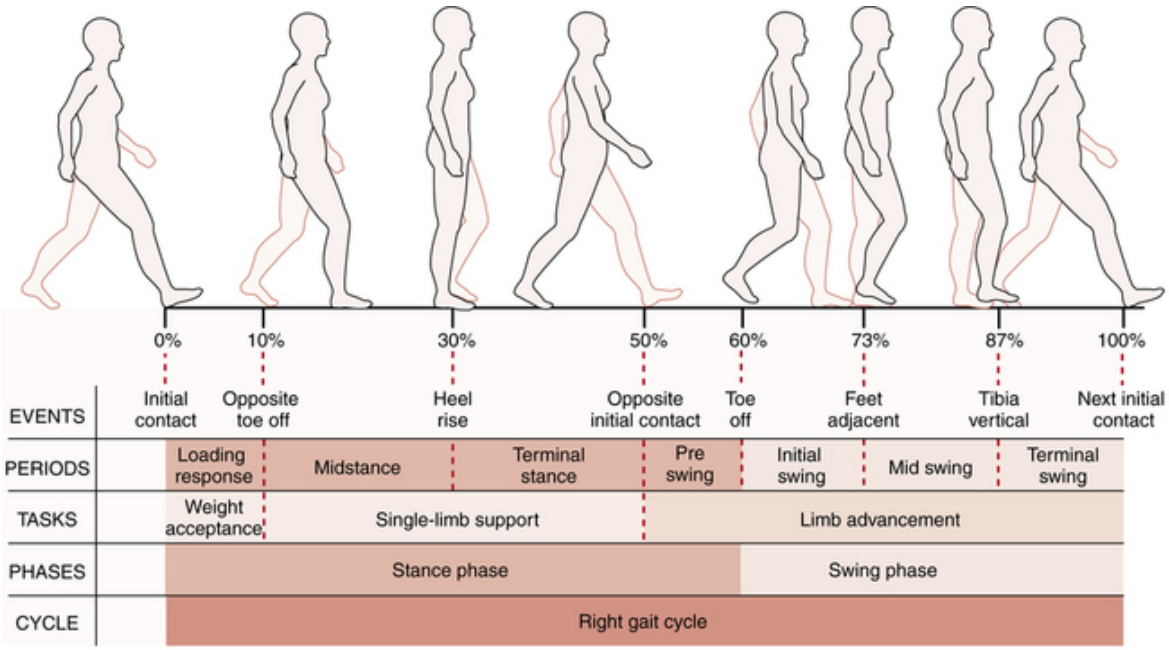

Figure 7.1 *Illustration of a typical gait cycle* (Neumann, 2017)

7.1.2 Age-related Changes in Gait

It is well-known that gait characteristics are modified with ageing (Cruz-Jimenez, 2017; Osoba et al., 2019; Boyer et al., 2017). Some of these features are considered to not just change but decline in older adults. Preferred walking speed is the most consistently observed age-related change in usual gait (Winter et al., 1990; Judge et al., 1996; Cruz-Jimenez, 2017; Byrne et al., 2002; Monaco et al., 2009; Riley, Della Croce and Kerrigan, 2001; Kerrigan et al., 1998, 2001; Anderson and Madigan, 2014). Slower walking speed is related to fear of falls (Chamberlin et al., 2005), muscle weakness (Busse, Wiles and Van Deursen, 2006) and impairment of motor control (Kaya, Krebs and Riley, 1998). However, it is still ambiguous whether speed decline with ageing is a compensatory effort to reduce injury risk (Winter et al., 1990; Chamberlin et al., 2005) or merely the manifestation of deteriorated muscle activity (Ko, Hausdorff and Ferrucci, 2010), or both. Adoption of a more cautious gait to reduce falls risk in older age is indicative of changes in other spatiotemporal parameters such as a shorter

step length, greater step timing variability, increased double-support time and reduced step frequency (Hollman, McDade and Petersen, 2011; Kang and Dingwell, 2008; Menz, Lord and Fitzpatrick, 2003; Toebes et al., 2012; Osoba et al., 2019).

Joint kinematics have also been shown to change with advancing age. Compared to younger adults, older adults exhibit reduced hip extension, reduced ankle plantarflexion and power generation, and increased anterior pelvic tilt during gait (Winter et al., 1990; Kerrigan et al., 2001; Byrne et al., 2002), independent of gait speed (Kerrigan et al., 1998; Cofré et al., 2011). Joint kinetics are also affected by ageing; representing an alteration in the motor pattern used to perform walking gait (DeVita and Hortobagyi, 2000). Walking ability in older adults is maintained through increased activity of the proximal muscles (DeVita and Hortobagyi, 2000; McGibbon and Krebs, 2001; Monaco et al., 2009). This results in changes in joint power generation and absorption throughout the GC, which act as key markers to identify differences in aged gait (DeVita and Hortobagyi, 2000). Older adults typically generate less propulsive power at the ankle during pre-swing but generate greater relative hip power to maintain trunk stability during stance and assist in leg swing (DeVita and Hortobagyi, 2000; Watelain et al., 2000; McGibbon, 2003; Silder, Heiderscheit and Thelen, 2008; Monaco et al., 2009; Winter et al., 1990; Judge et al., 1996; Kerrigan et al., 1998, 2001; Riley, Della Croce and Kerrigan, 2001). Studies suggest that older adults walk with greater hip flexion and reduced plantarflexion torque as a strategy to compensate for age-related decrements in plantarflexion strength and lower-limb ROM (DeVita and Hortobagyi, 2000; Silder, Heiderscheit and Thelen, 2008; Monaco et al., 2009; Judge et al., 1996; Goldberg and Neptune, 2007; Cofré et al., 2011). Whilst the exact mechanism is equivocal, it is established that hip power during pre-swing (PS) and initial swing (IS) increases in older adults to overcome the loss in ankle function and increase forward propulsion. As the majority of lower-limb joint power and mechanical work is generated or absorbed in the sagittal plane (Neptune, Sasaki and Kautz, 2008; Eng and Winter, 1995), most studies have focused on flexion/extension kinematics and kinetics with less attention on the coronal and transverse planes. However, functional gait is determined by biomechanical contributions in all three cardinal planes.

7.1.3 Age-related Changes in the Trunk during Gait

Previous research and clinical assessment have primarily focused on the lower limb joints in gait analysis. The role of the trunk during gait is often overlooked as an integral component that contributes to mobility and stability. The trunk muscles, particularly the LPMs, are critical to walking gait and actively contribute to dynamic balance during functional activities (Karthikbabu et al., 2011; Cromwell et al., 2001; Ceccato et al., 2009; Carmo et al., 2012; Hicks et al., 2005b; Cholewicki, Panjabi

and Khachatryan, 1997). They are responsible for maintaining an upright or neutral posture and controlling dynamic balance by maintaining COG over the BOS (Karthikbabu et al., 2011).

Compared to the lower limbs, the trunk undergoes negligible excursions in the sagittal, coronal and transverse planes during gait in healthy adults (Krebs et al., 1992; Leardini et al., 2013; Titus et al., 2018). However, trunk kinematics have been associated with age-related changes (Takahashi et al., 2005) and with maintenance of dynamic stability in older adults (Hurt et al., 2010). Therefore, understanding trunk motion during gait is of clinical importance not only in the presence of pathology (Engsberg et al., 2001), but also in uncovering the effects of ageing. To separate comorbidities from the effects of ageing, establishing normal age-related changes in trunk biomechanics during gait is needed. Whilst the effect of age on lower limb kinematics and kinetics during gait is well documented, the trunk has received less attention. This is likely due to the complexity of the human spine and the fact that lower limb joints are more easily and accurately modelled. Equally, gait is an activity that is driven by the lower limb musculature and may therefore be seen as more important to analyse. Regardless, increasing our knowledge of normal age-related changes in trunk kinematics and kinetics is essential for targeted interventions in healthy older adult populations as well as identifying and treating pathological gait patterns. The paucity of research and a range of disparate methods used to analyse the trunk has confounded our understanding of age-related changes in trunk biomechanics during gait.

7.1.3.1 Analysis of Trunk Kinematics during Gait

Trunk ROM is typically variable amongst individuals, although evidence suggests that spine and trunk motion is affected by ageing (Intolo et al., 2009; Arshad et al., 2019). Despite this, few studies have investigated the effect of ageing on trunk kinematics during gait (Schmid et al., 2017; Crawford et al., 2018; McGibbon and Krebs, 2001; Van Emmerik et al., 2005). Furthermore, Schmid et al. (2017) and Crawford et al. (2018) did not analyse the continuous time-series waveform of trunk kinematics, rather analysing discrete data points. This may have led to missed opportunities to uncovered subtle differences in trunk movements between older and younger adults during the GC. Schmid and colleagues (2017) analysed trunk kinematics in all three cardinal planes, reporting age-related ROM increases in all three (0.6° sagittal, 0.3° coronal and 1.0° transverse). These increases, whilst non-significant, were attributed to the ageing process and not an artefact of spaciotemporal parameters as walking speeds were similar between age groups. McGibbon and Krebs (2001) evaluated the entire kinematic waveform to investigate trunk and pelvis leading strategies in gait. Older adults were found to adopt a trunk leading strategy in contrast to younger adults. Furthermore, trunk ROM relative to

the pelvis in the sagittal plane was significantly lower in the OG (3.95 ± 1.68°) compared to the YG (4.78 ± 1.80°). Modelling the thorax relative to the pelvis similar to McGibbon and Krebs (2001), Van Emmerik et al. (2005) also found that trunk flexion/extension in older individuals was reduced yet trunk axial rotation was increased compared to younger adults. Increased trunk rotations are an indicator of immature and pathological gait and generally seen as a destabilising feature of walking (Winter, 1995; Ledebt and Bril, 2000). To account for this, pelvic rotations in the coronal and transverse planes are able to create a more energy efficient gait by decreasing COM vertical oscillations (Saunders, Inman and Eberhart, 1953). Interestingly, Van Emmerik et al. (2005) also reported that pelvic rotations in all three planes of motion were systematically reduced with increasing age. This indicates that the pelvis may be particularly influential in determining age-related changes in trunk kinematics. However, it is important to note that these differences were specific to slower walking speeds (Van Emmerik et al., 2005), which were typically slower than preferred walking speeds in both younger and older adults (**Table 7.2**). This may cause inconsistent findings with other studies and ecological validity may be diminished by the prescription of unnatural walking speeds.

Disparities in kinematics may have also occurred due to different modelling methods. Each of these studies used a different approach to define the trunk segment. This is reflected in the discrepancy between Schmid et al. (2017) and McGibbon and Krebs (2001). Schmid et al. (2017) defined the lumbar segment using individual markers representing each lumbar spinous process; McGibbon and Krebs (2001) used the mid-sections of the trunk and pelvis to approximate the L4/5 junction. The latter approach is more susceptible to skin movement artefact and with fewer degrees of freedom it may overestimate articulations in the lumbar spine (Raabe and Chaudhari, 2016). Whilst validation of these models is required to ascertain which is superior, the different modelling assumptions offers an explanation for the disparity in age-related changes in trunk kinematics during gait. Another model, Plug-in Gait (PIG), has been widely used in the literature to estimate lumbar spine kinematics as the intersection between modelled trunk and pelvic segments (Titus et al., 2018; Romkes et al., 2007; Sanz-Mengibar et al., 2017; Chung et al., 2010). Whilst more specific models for the assessment of spinal motion exist (Taylor, Goldie and Evans, 1999; Schmid et al., 2017; Konz et al., 2006), the increased complexity, data collection and computational time make such detailed approaches less clinically applicable (Gutierrez et al., 2003). Furthermore, more complex models may not necessarily produce more accurate data as each model will intrinsically contain modelling assumptions. Indeed, complex spinal models may even increase measurement error as the size of functional spinal units is small and limits our ability to position three non-colinear markers on the skin overlying multiple functional spinal units (Konz et al., 2006). Furthermore, as movement between vertebral bodies is smaller than between the trunk and pelvis segments, measurement error may be greater than

measured differences when modelling individual spinal units. This would prevent meaningful differences from being observed with any degree of confidence.

Gait studies analysing trunk kinematics in all three cardinal planes (Schmid et al., 2017; Krebs et al., 1992; Vogt, Pfeifer and Banzer, 2002; Fernandes et al., 2016; Chung et al., 2010; Hendershot and Wolf, 2014; Stokes, Andersson and Forssberg, 1989; Opila-Correia, 1990; Thurston, 1985; Leardini et al., 2013; Taylor, Goldie and Evans, 1999; Whittle and Levine, 1999; Sartor et al., 1999; Van Emmerik et al., 2005; Crosbie, Vachalathiti and Smith, 1997) have generally recruited healthy younger adults and not focused on the ageing process, although interesting findings have been revealed regarding the antiphase nature of the trunk and pelvis during gait (Titus et al., 2018; Krebs et al., 1992; Sartor et al., 1999; Van Emmerik et al., 2005). Krebs et al. (1992) found that trunk axial rotation was 180° out of phase with the pelvis during loading response. This timing difference was attributed to the pelvis continuing to rotate in the opposite direction to the trunk after the trunk had finished rotating shortly after heel strike (Krebs et al., 1992). Similar kinematic patterns have been reported for the trunk relative to the pelvic and global reference frames (Thurston, 1985; Titus et al., 2018; Chung et al., 2010; Vogt, Pfeifer and Banzer, 2002). However, other studies indicate that antiphase rotation between the trunk and pelvis does not reach 180° (Whittle and Levine, 1999; Taylor, Goldie and Evans, 1999; Leardini et al., 2013). Interestingly, trunk motion in the sagittal plane is considered variable amongst individuals whilst trunk obliquity and rotation exhibit more consistent between-subject waveforms (Whittle and Levine, 1999; Thorstensson et al., 1982). This can be seen in **Table 7.2**, as standard deviations are generally greater relative to ROM means in the sagittal plane, meaning that coefficients of variation are larger than in the coronal or transverse planes.

Studies analysing trunk kinematics have traditionally reported local maxima and minima values of the time series data (**Table 7.2**). To explore the age-response in trunk kinematics during gait, analysing phase-specific effects may be more valuable than simply comparing gait peaks and troughs. Indeed, comparing discrete measures may lead to inconsistent findings between studies. Needham, Stebbins and Chockalingam (2016) showed that trunk ROM in all three cardinal planes varied between clusters of studies due to specific configurations within different laboratories. Indeed, lumbar spine ROM could vary as much as 10° between studies depending on the plane of motion (Needham, Stebbins and Chockalingam, 2016). Despite differences in methodologies and equipment, the literature consistently shows less trunk ROM in the sagittal plane compared to the coronal and transverse planes (**Table 7.2**). However, it is unknown how trunk kinematics change with advancing age. In healthy ageing, gait speed decline is not always observed (Schmid et al., 2017), and possibly has little effect on trunk kinematics (Taylor, Goldie and Evans, 1999) although this is equivocal (McGibbon and Krebs, 2001). Therefore, typical changes associated with ageing gait may not alter trunk movement. Due to inconsistent

methods, disparate findings, a lack of research focusing on age-related changes in trunk kinematics and the importance of the trunk in performing ADLs, further investigation into age-related alterations in trunk kinematics during gait is warranted.

Table 7.2 *Characteristics of studies analysing trunk kinematics during gait in healthy populations*

Study	Sample characteristics	Segment / model	Gait conditions	Model outcomes	Planes	Findings
Thorstensson et al. (1982)	Healthy men (n = 7) aged 18 to 34 years	The angle between the line connecting C7 and L3 and the vertical in the frontal plane and the corresponding angle in the sagittal plane.	Set range of walking speeds from 1.0-2.5 m/s	Absolute trunk angle	Sagittal Coronal	Trunk movements in the frontal plane (2-9°) were less variable than in the sagittal plane (2-12°)
Thurston (1985)	Healthy male controls (n = 10) aged 63.4 ± 8.1 years	Clusters attached to the upper lumbar spine and the sacrum.	Self-selected walk speed	Relative lumbar spine angle	Sagittal Coronal Transverse	Relative spine flexion/extension ROM was 5.2 ± 1.1°, lateral flexion ROM was 6.8 ± 1.8° and axial rotation was 8.8 ± 2.5°.
Stokes, Andersson and Forssberg (1989)	Normal females (n = 3) and males (n = 5)	LEDs attached to triangular plates fitted to the posterior of the pelvis and thorax via a waist belt and shoulder harness.	Self-selected walk speed	Absolute trunk angle	Sagittal Coronal Transverse	The thorax had a biphasic rotational pattern. Subjects inflexion points occurred near FC. The right shoulder was elevated (with respect to the left) in synchrony with the forward swing of the left leg. ROM was 3.2 ± 0.9° in the sagittal plane, 4.7 ± 2.0° in the coronal plane and 4.6 ± 1.4° in the transverse plane.
Opila-Correia (1990)	Female subjects (n = 14)	Abstract only, model not described.	High vs low-heeled gait	Absolute trunk angle	Sagittal Coronal Transverse	Trunk ROM was 11.1° in the sagittal, 12.6° in the coronal and 17.5° in the transverse planes.
Krebs et al. (1992)	Healthy women (n = 6) and healthy men (n = 5) aged 58.9 ± 17.9 years	11 segment whole body model: head, trunk, pelvis, thighs, shanks, feet, and upper arms. Each segment was modelled as a rigid body having 6 degrees of freedom.	Self-selected walk speed (1.1 ± 0.1 m/s)	Absolute and relative trunk angles	Sagittal Coronal Transverse	Flexion peak near each heel-strike, with maximum extension occurring during single-limb support, but the amplitude of these motions was small. Frontal-plane trunk motions relative to the pelvis tended to occur toward the stance limb, reaching their maximum at the time of opposite side toe- off. Transverse trunk rotation relative to room coordinates was also 180 degrees out of phase with the pelvis and achieved maxima about 10% of a cycle after each heel-strike, rotating so that the ipsilateral shoulder was posterior to the heel-strike limb, nearly directly over the foot at

						mid-stance, and maximally anterior to the stance limb near toe-off.
Crosbie, Vachalathiti and Smith (1997)	Healthy volunteers (n = 108) from the local community. Men (n = 50) were aged 46.3 ± 18.3 years and women (n = 58) were aged 45.2 ± 18.6 years	13 body markers were attached to the dorsal surface of the trunk and three each to the right and left thighs. Segments modelled as rigid bodies and movement defined within spinal regions in terms of the relative motion between the rigid body above and the rigid body below the region of interest.	Self-selected walk speed (1.1 ± 0.1 m/s)	Relative and absolute upper and lower trunk, and lumbar angles	Sagittal Coronal Transverse	Consistent patterns were observed within and between segments and movements, with apparent consequential trunk motion following pelvic displacements. This suggests that the spinal movements associated with walking are linked to the primary motions of the pelvis and the lower limbs. Pelvis variance was substantially greater than that of the trunk segments. Lumbar ROM was 3.5 ± 2.0° in the sagittal plane, 9.0 ± 3.5° in the coronal plane and 4.5 ± 2.0° for axial rotation.
Callaghan, Patla and McGill (1999)	Healthy male university students (n = 5) aged 25.0 ± 2.8 years	Fifteen infrared diodes were attached to define a five-segment rigid link model: right foot, right leg, right thigh, pelvis, and trunk. A rigid plate with three markers was attached to the posterior aspect of the sacrum with a second similar plate attached at the T12/L1.	Fast, slow and normal walking based on cadence (normal = 103.2 ± 4.4 steps/min)	Relative lumbar angles	Sagittal Coronal Transverse	The motion of the trunk, at least at the straddle position, seems to offset the rotation of the pelvis maintaining a smaller net range of motion for the lumbar spine. Flexion extension of the trunk was found to be the most variable measure in lumbar kinematics. Mean ROMs were 6.2° (flexion/extension), 6.7° (lateral bend), 7.1° (axial twist).
Sartor et al. (1999)	Healthy men (n = 6) and women (n = 11) with a mean age of 28 and range of 21 – 47 years	Markers were placed on the spinous processes of T4 and T9, sternal notch, bilateral ASIS, spinous process of S2. For the trunk, the sternal notch and T4 markers were used to create the anatomical axis, while the T9 marker made up the plane.	Self-selected walk speed (1.36 m/s)	Absolute and relative trunk angles	Sagittal Coronal Transverse	Relative to the pelvis, the trunk was extended an average of 5° at initial contact. The trunk, relative to the pelvis, was extended throughout the gait cycle but exhibited two small peak oscillations in extension at the end of mid-stance and during mid-swing. Relative to the pelvis, the trunk was laterally extended an average of 1° toward the stance limb at initial contact. As the body moved through the gait cycle, the trunk continued to laterally extended until midstance, when it reached a maximum value of 6° of lateral extension over the stance limb.

Taylor, Goldie and Evans (1999)	Normal healthy young adults (n = 27) aged 23.5 ± 5.1 years and 20.6 ± 2.8 years dependent on walking speed group (slow and self-selected, respectively).	Retroreflective markers placed on orthogonal rigs on subject's sacrum and superior lumbar spine. Rigid segments created from markers to determine the angular movements of lumbar flexion/extension, axial rotation and lumbar lateral flexion.	Self-selected treadmill walk speed	Absolute and relative lumbar angles	Sagittal Coronal Transverse	Amplitude of lumbar lateral flexion decreased with slower walking. In contrast, absolute lumbar spine movements did not differ due to decreased walking speed. The lumbar spine should be interpreted with respect to a frame of reference. In the local reference frame, lumbar ROM was 3.8 ± 1.6° (sagittal), 12.0 ± 1.9° (coronal) and 6.4 ± 1.9° (transverse). In the global reference frame, lumbar ROM was 3.2 ± 0.7° (sagittal), 3.5 ± 1.3 (coronal) and 9.0 ± 3.0° (transverse).
Whittle and Levine (1999)	Healthy young adult males (n = 20)	Motion of the lumbar spine was derived from the differences between the motion of the pelvis and the motion of the thoraco-lumbar junction, as determined by the relative motion between the two sets of marker cluster rigs.	Slow and self-selected treadmill walk speed	Relative lumbar spine angle	Sagittal Coronal Transverse	Change in lumber lordosis across the gait cycle was consistent within subjects but varied considerably between subjects. The phase relationships between pelvic tilt and lumbar lordosis also varied considerably between subjects. Movement patterns were more consistent in the coronal plane. Lateral bend generally followed the pattern of pelvic obliquity. The transverse plane showed similar waveforms between axial rotation of the pelvis and axial rotation of the lumbar spine, except that motion of the pelvis was of greater magnitude and occurred later in the gait cycle than the motion of the lumbar spine. Lumbar spine ROM was 4.0 ± 1.2° in the sagittal, 7.6 ± 1.7° in the coronal and 8.3 ± 2.2° in the transverse planes.
Cromwell et al. (2001)	Healthy female (n = 2) and male (n = 6) volunteers aged 25.6 ± 3.3 years	Trunk segment defined by markers placed at the interspace between the fifth lumbar and the first sacral vertebrae and the interspace between the sixth and the seventh cervical vertebrae. Segmental angles were calculated with respect to an external horizontal reference.	Self-selected walk speed	Absolute trunk angle	Sagittal	During initial double limb support, the trunk moved from a position of flexion toward extension. At the beginning of single limb support, the trunk was at maximum extension and then began to flex approaching terminal double limb support. Trunk maintained flexion throughout the gait cycle. Mean average excursion was 3.3° and mean forward position was 7.9°.

McGibbon and Krebs (2001)	Healthy adults (n = 93) stratified into an old group (70.7 ± 8.7 years) and young group (29.8 ± 6.8 years).	Trunk defined as a rigid segment between the C1–C2 and L4–L5, and the pelvis as a rigid segment between the low-back joint and the hips. Lower trunk referred to the trunk–low-back joint–pelvis system.	Self-selected walk speed (Old 1.1 ± 0.2 m/s; Young 1.3 ± 0.2 m/s)	Absolute and relative trunk angles	Sagittal	Low-back ROM was significantly greater in young subjects compared with old subjects 4.8 ± 1.8° vs 3.9 ± 1.7°. However, trunk ROM not significant between age groups 4.4 ± 1.2° vs 4.0 ± 1.6°. Peak low back and trunk angles were also not significantly different (trunk: 5.5 ± 2.1° vs 6.1 ± 4.0°; low back: 7.3 ± 3.5° vs 7.4 ± 4.8°).
Vogt, Pfeifer and Banzer (2002)	Healthy men (n = 9) aged 28.7 ± 4.4 years	Plate mounted ultrasound markers were attached on the S1 and the T12. The projection method of angle definition was used for calculation of net angular displacements in the sagittal, transverse and frontal plane.	Self-selected walk speed (1.1 m/s)	Relative thorax angle	Sagittal Coronal Transverse	Differences exist for some angular lumbar spine movement parameters between walkway and treadmill locomotion, specifically in the coronal and transverse planes. Flexion/extension ROM was 4.4°, lateral flexion ROM was 3.9° and axial rotation ROM was 8.2°.
Van Emmerik et al. (2005)	Healthy men (n = 15) and women (n = 15) stratified into equally sized (n = 10) younger (23.3 ± 4.0 years), middle (49.3 ± 5.4 years) and older (72.6 ± 3.8 years) age groups.	The trunk was defined by three markers, one aligned with C7 and the other two near the bottom of the rib cage. Markers on the pelvis were attached to the left and right posterior aspect of the ilium and the sacrum. Lumbo-sacral joint angle was defined as the rotations of the trunk with respect to the pelvic reference frame.	Predetermined walking speeds (0.2 to 1.8 m/s)	Relative trunk angle	Sagittal Coronal Transverse	Pelvic rotations in sagittal, frontal and transverse planes of motion were systematically reduced with age. Older individuals showed reduced trunk flexion–extension in the sagittal plane and increased trunk axial rotation in the transverse plane.
Leteneur et al. (2009)	Healthy young men (n = 25) aged 26.2 ± 5.2 years	14 body segment model. The neck and trunk were considered as a single segment using shoulder and hip markers. A marker was placed on the L5.	Self-selected walk speed (1.4 ± 1.1 m/s)	Absolute trunk angle	Sagittal	Trunk inclination variation is associated with the type of walking patterns. Forward leaners exhibit a flexed trunk position which is reversed in backward leaners. Regardless of trunk inclination, trunk ROM is approximately 2-3°.
Chung et al. (2010)	Healthy men (n = 11) and women (n = 9) aged 32 ± 6	PiG full body marker model	Self-selected walk speed (1.2 ± 0.1 m/s)	Absolute and relative trunk angles	Sagittal Coronal Transverse	Trunk motions to the ground showed narrow ranges in all three planes, whereas trunk motions relative to the pelvis tended to be larger. Trunk tilt relative to the pelvis and global reference planes of women were

	years and 29 ± 6 years, respectively					about 5° less than those of men, meaning a more extended trunk posture in women. Trunk ROM relative to the global was 4° in an anterior position, 3° lateral flexion and 7° rotation. Whilst ROM relative to the pelvis was 5° in a generally extended position, 13° in the coronal plane and 14° in the transverse plane.
Leardini et al. (2013)	Healthy male (n = 15) and female (n = 15) participants aged 26.5 ± 3.5 years	Marker-set included 14 markers, 4 on the pelvis and 10 on the trunk. At the thorax, a technical reference frame was first defined by applying the Single Value Decomposition procedure to the T2, MAI, PX and IJ markers for optimal pose, i.e. position and orientation, and estimation.	Self-selected walk speed (1.3 m/s)	Absolute and relative trunk angles	Sagittal Coronal Transverse	Sagittal thorax inclination attitude altered three-dimensional kinematic patterns of the upper trunk segments during natural gait. Trunk flexion/extension ROM (~3°) was generally less than in lateral flexion (global = ~3°, relative = ~13°) and in axial rotation (global = ~7°, relative = ~ 12°).
Hendershot and Wolf (2014)	Male able-bodied controls (n = 20) aged 28.1 ± 4.8 years	Markers were placed on the S1, T10, C7, sternal notch, xiphoid, acromion processes, ASIS, PSIS, and lower extremities (modified Cleveland Clinic marker set). The trunk was a single rigid segment, defined proximally by the acromia, C7, and sternal notch, and attached distally to the pelvis at the lumbosacral (L5/S1) joint.	Self-selected walk speed (1.4 ± 0.1 m/s)	Relative trunk angle	Sagittal Coronal Transverse	In the frontal plane, the trunk flexed laterally towards the support leg, reaching a peak during single-limb stance (ROM ~ 8°). In the sagittal plane, the trunk flexed forward following heel strike, extending prior to subsequent heel strike (ROM ~ 2°). In the transverse plane, the trunk rotated towards the support leg, with peak rotations occurring around heel strike (ROM ~ 13°).
Fernandes et al. (2016)	Convenience sample of healthy men (n = 11) and women (n = 12) aged 35 ± 7.3 years	9-segment model. Lumbar joint centre was defined through a virtual marker created along the distance connecting the L5–S1 marker and the midpoint between the two ASIS markers, projected from the thoracic joint centre.	Self-selected walk speed (1.2 m/s)	Relative lumbar angle	Sagittal Coronal Transverse	Varied reliability indices for multi-segment trunk joint angles and joint moments during gait and an acceptable level of error, particularly for sagittal plane parameters. Lumbar flexion/extension ranged from -6.5 to -9.0°, lateral flexion ranged from 1.9 to -2.0° and axial rotation ranged from 0.9 to -4.3°.

Aminiaghdam et al. (2017)	Healthy male (n = 6) and female (n = 6) volunteers aged 26.0 ± 3.4 years	A thirteen-body segment model. Trunk angle defined by the line connecting the L5 marker and the C7 marker with respect to the vertical axis of the lab coordinate system.	Self-selected walk speed (~ 1.5 m/s)	Absolute trunk angle	Sagittal	Able-bodied individuals recovered almost all assessed kinematic parameters comprising the vertical position of the CoM, effective leg length and angle as well as hip, knee and ankle joint angles at the end of the step-up, suggesting an adaptive capacity and hence a robustness of human walking with respect to imposed trunk orientations. Trunk inclination at foot contact was 6.2 ± 3.4° and 5.0 ± 3.4° at foot off.
Schmid et al. (2017)	Healthy volunteers: n = 14 adolescents (14 ± 2 years); n = 13 adults (27 ± 3 years); n = 15 older adults (70 ± 2 years)	IfB full body marker set combined with PiG full body marker set.	Self-selected walk speed (Adolescents 1.3 ± 0.1 m/s; Adults 1.5 ± 0.3 m/s; Older 1.6 ± 0.1 m/s)	Absolute and relative lumbar angles	Sagittal Coronal Transverse	Kinematic differences from early adulthood to older age are gait speed dependent. Lumbar ROM increased from 4.4 ± 1.2° to 5.0 ± 2.6° (sagittal), 6.5 ± 2.1° to 6.8 ± 2.7° (coronal) and 9.9 ± 3.8° to 10.9 ± 4.7° (transverse).
Crawford et al. (2018)	Asymptomatic women (n = 3) and men (n = 7) aged 26.3 ± 2.5 years in the younger group, and nine asymptomatic adults (n = 3 women) aged 67.1 ± 4.2 years in the older group	Lumbar lordosis (defined by SACR, LUM, and TLJ markers), and trunk inclination (C7-SACR line versus vertical).	Predetermined walk speeds of 2 and 4 km/h	Lumbar lordosis angle and trunk angle	Sagittal	Trunk kinematics change with ageing and lumbar lordosis angles are not gait speed dependent. Trunk inclination ROM decreased from 2.9 ± 0.8° to 2.3 ± 0.8° with age whilst lumbar lordosis decreased from 4.4 ± 3.8° to 2.7 ± 1.4°.

7.1.3.2 Analysis of Trunk Kinetics during Gait

Fewer studies have analysed trunk kinetics during gait compared to studies assessing kinematics. Whilst valuable to understand movement patterns, it is of prime importance to understand the internal forces that bring about these movements. Internal joint moments are crucial in maintaining dynamic stability (Yack and Berger, 1993) and actuating trunk movements (Hendershot and Wolf, 2014). In the spine, the LPMs provide a continuous extensor moment to counteract the external flexor moment produced by the anterior location of the trunk's COM (Cresswell, Oddsson and Thorstensson, 1994). With ageing the thoracolumbar bending moment increases due to postural changes causing an anterior shift in the COM of the trunk (Le Huec et al., 2018), which may be further influenced by the age-related increase in abdominal adiposity (Ponti et al., 2020). The LPMs are more solicited under these conditions, which increases the compressive and shear forces on the lumbar discs by as much as 20% (Le Huec et al., 2018). The increase in external flexion moment and subsequent counterbalancing muscular effort may be theoretically great enough to cause vertebral fractures in older adults(Le Huec et al., 2018).

To the author's knowledge, no studies have investigated the effect of ageing on spinal moments in all cardinal planes during gait and only one has investigated age-related changes in lumbar spine joint power (McGibbon and Krebs, 2001). Of the gait studies identified, lumbar spine moments have been analysed to explore normal patterns in healthy individuals (Callaghan, Patla and McGill, 1999), pathological deviations in populations with lower-extremity amputation (Hendershot and Wolf, 2014), the effects of different postures (Leteneur et al., 2009) and test-retest reliability (Fernandes et al., 2016). However, the reliability of lumbar spine moments may be questionable due to data processing methods (Leteneur et al., 2009; Hendershot and Wolf, 2014; Callaghan, Patla and McGill, 1999). Different cut-off frequencies were used to filter kinematic and force data (Leteneur et al., 2009; Hendershot and Wolf, 2014; Callaghan, Patla and McGill, 1999), which may have introduced joint moment artefacts (Bisseling and Hof, 2006; Kristianslund, Krosshaug and Van den Bogert, 2012; Bogert and Koning, 1996). Furthermore, there is low confidence in what normal trunk moment waveforms look like during gait due to studies reporting highly disparate results (Callaghan, Patla and McGill, 1999). These ambiguous findings were likely due to modelling assumptions as Callaghan, Patla and McGill (1999) excluded the left limb in their biomechanical model and subsequently observed conflicting joint moment and EMG results. These results, which show that the flexor muscles are solicited throughout most of the GC (Callaghan, Patla and McGill, 1999), are also in contrast to the literature; reporting extensor muscle activity is more dominant throughout the GC (Hendershot and Wolf, 2014; Fernandes et al., 2016; Leteneur et al., 2009; Raabe and Chaudhari, 2016).

McGibbon and Krebs (2001) found that peak eccentric low-back power increased with ageing, which may be a consequence of muscles absorbing more energy to compensate for increased tendon stiffness in older age (Tuite, Renström and O'Brien, 2007; Gajdosik et al., 2005). However, there are no other studies to substantiate these findings and increased tendon stiffness with ageing is not a universal observation. Other researchers have shown that tendon stiffness decreases with advancing age (Onambélé, Narici and Maganaris, 2006). Despite different methodological approaches, trunk kinetics are surprisingly consistent (Hendershot and Wolf, 2014; Fernandes et al., 2016; Leteneur et al., 2009). The LPMs appear to be active throughout most of the GC, evidenced through internal extensor moments (Hendershot and Wolf, 2014; Fernandes et al., 2016; Leteneur et al., 2009; Raabe and Chaudhari, 2016). A biphasic pattern is also apparent, where extensor peaks are produced at approximately 10 – 20 % (loading response to midstance) and 55 – 65 % (TO) of the GC (Hendershot and Wolf, 2014; Leteneur et al., 2009; Raabe and Chaudhari, 2016). EMG findings substantiate this phenomenon, revealing peaks of ES and MF electrical activity during the same GC phases (Lamoth et al., 2004; Callaghan, Patla and McGill, 1999). McGibbon and Krebs (2001) found that during double support and early single support, the LPMs contract eccentrically then concentrically during late single support in older adults. Interestingly, the activation pattern of the LPMs is reversed in older age (McGibbon and Krebs, 2001). It has also been suggested that older adults increase lower back joint power to advance the lower limbs into swing phase to compensate for weakened lower extremity muscles (McGibbon, Krebs and Puniello, 2001). However, due to the lack of available literature on this topic and variability in reported joint moments (**Table 7.3**), the effect of age on trunk kinetics during gait is not fully understood.

Table 7.3 Characteristics of studies analysing trunk kinetics during gait in healthy populations

Study	Sample characteristics	Segment / model	Gait conditions	Model outcomes	Planes	Findings
Callaghan, Patla and McGill (1999)	Healthy male university students (n = 5) aged 25.0 ± 2.8 years	15 infrared diodes were attached to define a five-segment rigid link model: right foot, right leg, right thigh, pelvis, and trunk. A rigid plate with three markers was attached to the posterior aspect of the sacrum with a second similar plate attached at the T12/L1.	Fast, slow and normal walking based on cadence (normal = 103.2 ± 4.4 steps/min)	Joint moments at L4/5	Sagittal Coronal Transverse	The flexion/extension moment curve exhibited two maximums, which occurred approximately at toe off. The two minimum values started just prior to heel strike. At heel contact there was a flexor peak moment present followed by an extensor peak around toe off. Peaks ranged from -1.6 to 1.8 Nm/(kg*m). The lateral bend moment produced a consistent pattern, oscillating about the moment zero axis. Corresponding to heel contact there was a lateral bend moment to the side of contact. Prior to toe off and swing phase there was a lateral bend moment to the swing leg side returning to the opposite side following the consequent heel strike. Peaks ranged from -1.4 to 2.6 Nm/(kg*m). Axial twist moments were small ranging from -0.7 to 0.7 Nm/(kg*m). From toe off to the following heel contact there was a twist moment to the contralateral side.
McGibbon and Krebs (2001)	Healthy adults (n = 93) stratified into an old group (70.7 ± 8.7 years) and young group (29.8 ± 6.8 years).	Trunk defined as a rigid segment between the C1–C2 and L4–L5, and the pelvis as a rigid segment between the low-back joint and the hips. Lower trunk referred to the trunk–low-back joint–pelvis system.	Self-selected walk speed (Old 1.1 ± 0.2 m/s; Young 1.3 ± 0.2 m/s)	Net mechanical power at the L4/5	Sagittal	Net mechanical power waveform is reversed in older age for the low-back due to trunk leading strategy. Peak eccentric power is greater in elderly compared to younger adults and occurs when entering single-support phases (approx. young = 4-5 Watts/kg vs old = 10-15 Watts/kg). Peak concentric power is similar between old and young (approximately 4-7 Watts/kg).
Leteneur et al. (2009)	Healthy young men (n = 25) aged 26.2 ± 5.2 years	14 body segment model. The neck and trunk were considered as a single segment using shoulder and	Self-selected walk speed (1.4 ± 1.1 m/s)	Lumbosacral L5 moments	Sagittal	Thoraco-lumbar extension moment peaks were 1.4 times higher for the forward leaners while flexion moment peaks were approximately 1.4 times higher

		hip markers. A marker was placed on the L5.				for the backward leaners. Moments ranged from 0.7 Nm/kg in flexion to 1.2 Nm/kg in extension.
Hendershot and Wolf (2014)	Male able-bodied controls (n = 20) aged 28.1 ± 4.8 years	Markers were placed on the S1, T10, C7, sternal notch, xiphoid, acromion processes, ASIS, PSIS, and lower extremities (modified Cleveland Clinic marker set). The trunk was a single rigid segment, defined proximally by the acromia, C7, and sternal notch, and attached distally to the pelvis at the lumbosacral (L5/S1) joint.	Self-selected walk speed (1.4 ± 0.1 m/s)	Lumbosacral L5/S1 moments	Sagittal Coronal Transverse	Increased and asymmetric peak moments at the low back among persons with unilateral lower-extremity amputation, particularly in the frontal plane, suggest potential mechanistic pathways through which repeated exposure to altered trunk motion and spinal loading may contribute to low-back injury risk. For the control group, mean low back moments were 0.2 ± 0.1 Nm/(kg*m) in flexion/extension, 0.1 ± 0.1 Nm/(kg*m) in lateral flexion and 0.1 ± 0.03 Nm/(kg*m) in axial rotation.
Fernandes et al. (2016)	Convenience sample of healthy men (n = 11) and women (n = 12) aged 35 ± 7.3 years	9-segment model. Lumbar joint centre was defined through a virtual marker created along the distance connecting the L5–S1 marker and the midpoint between the two ASIS markers, projected from the thoracic joint centre.	Self-selected walk speed (1.2 m/s)	Lumbar joint moment	Sagittal Coronal Transverse	Varied reliability indices for multi-segment trunk joint moments during gait and an acceptable level of error, particularly for sagittal plane parameters. Moments in the sagittal plane varied from 0.42 to -0.23 Nm/kg, from 0.19 to -0.33 Nm/kg in the coronal plane and 0.07 to -0.13 Nm/kg in the transverse plane.

7.1.3.3 Functional Demand

Physical function has typically been assessed by scoring physical tasks based on timed performance (Zeng et al., 2016; Sternfeld, 2002; Hicks et al., 2005a; Shahtahmassebi et al., 2017). This approach may oversimplify complex biomechanics within the performance of a physical task; exemplified by DeVita and Hortobagyi (2000) showing altered kinetics and kinematics with ageing gait despite performance being similar amongst younger and older adults. Other studies have provided more in-depth biomechanical analyses for assessing physical function in older adults (Reeves et al., 2008; Samuel et al., 2011; Samuel, Rowe and Nicol, 2013). These studies reported the FD of everyday tasks by normalising joint moments to their maximum capacity assessed through isokinetic dynamometry. For movements such as normal gait, muscles are constantly changing their role throughout the GC despite their relatively constant activation (Winter et al., 1990; Eng and Winter, 1995; Silder, Heiderscheit and Thelen, 2008). This is demonstrated by the function of muscles transitioning between propulsion and stabilisation causing contraction type and angular velocity to change accordingly. To accurately determine FD, the kinematics and contraction type corresponding to the joint during the movement must be replicated using isokinetic dynamometry. Therefore, FD as an expression of how biomechanically challenging a task is provides an intuitive metric. It is also easily translated to real-world settings as FD is expressed as a percentage of MVC, rather than more abstract quantities such as joint moments. Whilst FD has been applied to the lower limbs (Reeves et al., 2008; Samuel et al., 2011; Samuel, Rowe and Nicol, 2013), it has not been investigated in the trunk. Due to changes in trunk kinematics and muscular activation patterns in older age, mechanical energy demands of the lower-back muscles increases (McGibbon and Krebs, 2001). Together with trunk strength reductions (see **Chapter 6**), it is likely that older adults experience an increase in FD in the lower-back during gait. To fill this gap in the literature, investigation into the age-related change in FD of the trunk during gait is needed.

7.1.4 Aims, Objectives and Hypotheses

The main aim of this study was to investigate the effect of age on biomechanical function of the trunk during normal walking gait. A secondary aim was to determine FD of the trunk during normal walking and investigate how it is affected in older age. A further aim was to investigate the relationship between morphological degeneration of the lumbar musculature and biomechanical outcomes. To achieve the study aims, specific objectives were to:

Table 7.4 *Objectives and hypotheses for chapter 7*

	Objective	Null Hypothesis
1	Determine an appropriate experimental set-up and biomechanical model to allow kinematic and kinetic analysis of the trunk in older and younger adults	n/a
2	Analyse trunk kinematics relative to the global and pelvic reference frames in all three Cardinal planes during the GC	Peak amplitudes and ranges of trunk motions will not be significantly reduced in the OG compared to the YG
3	Analyse trunk moments and powers in all three Cardinal planes during the GC	Peak trunk moments and powers in the OG will not be significantly different to the YG
4	Identify peak trunk moments at key instances during the GC and normalise to individual's maximal capacity to derive a measure of functional demand	Functional demand will not be significantly greater in the OG compared to the YG
5	Calculate total positive and negative work performed and trunk powers produced during the GC	a) The OG will not perform significantly more negative work than the YG b) The OG will not perform significantly less positive work than the YG
6	Compare trunk kinematics, kinetics and FD between the OG and YG using traditional and novel statistical methods to explore discrete and phase-specific differences in the GC	b) Kinematic and kinetic differences between age groups will not be phase-specific
7	Analyse the moderating effects of LPM morphology and VPA by including them as covariates in appropriate statistical tests	a) Age-related differences in LPM morphology will not significantly influence changes in trunk biomechanics during gait b) VPA will not significantly influence changes in trunk biomechanics during gait

7.2 Methods

7.2.1 Laboratory Set-up

Data collection sessions were conducted in the Gait Laboratory at UHCW (**Figure 7.2**). Three-dimensional (3-D) motion analysis was achieved using a Vicon motion capture system consisting of 14 infrared cameras (Vero2.2, Vicon, Vicon Motion Systems, Oxford, UK) sampling at 100 Hz and three integrated floor-mounted force plates (AMTI-OPT400600-1K-STT, AMTI, Watertown, MA, US) sampling at 1000 Hz. The system also consisted of two synchronised Vue video cameras (Vicon Motion Systems, Oxford, UK) capturing at 50 Hz. Capture frequencies were chosen based on Nyquist theorem (Nyquist, 1928; Shannon, 1998), where sampling frequency must be greater than twice the highest frequency movement in the signal plus one. Typical kinematic frequency content of gait is less than 15 Hz, therefore a minimum of 31 Hz capture frequency is required to preserve 99% of the signal power (Antonsson and Mann, 1985). Indeed, other studies have suggested that 10 Hz capture frequency is sufficient as the frequency of most fundamental harmonics during gait is less than 2 Hz (Dujardin et al., 1997). During normal gait the lower limb experiences a high frequency impulsive load at heel strike, with frequency components between 10 – 75 Hz (Simon et al., 1981). Therefore, a sampling frequency of 1000 Hz for the force plates should prevent signal aliasing. Others have suggested that sampling frequency should be ten times greater than the highest anticipated frequency in the signal to avoid signal aliasing (Challis, 2008). However, this is more pertinent when movements are atypical (Challis, 2008). Gait is highly cyclical and due to a large body of literature frequency content of gait is well established. Two sets of Brower timing gates (TCi System, Brower, Utah, US) were positioned 4 m apart in the centre of the 10 m walkway. This allowed sufficient distance for participants to achieve a steady walking speed. The timing gates were used to calculate walking speed for the gait trials and provided real-time feedback during the practice trials to ensure consistency.

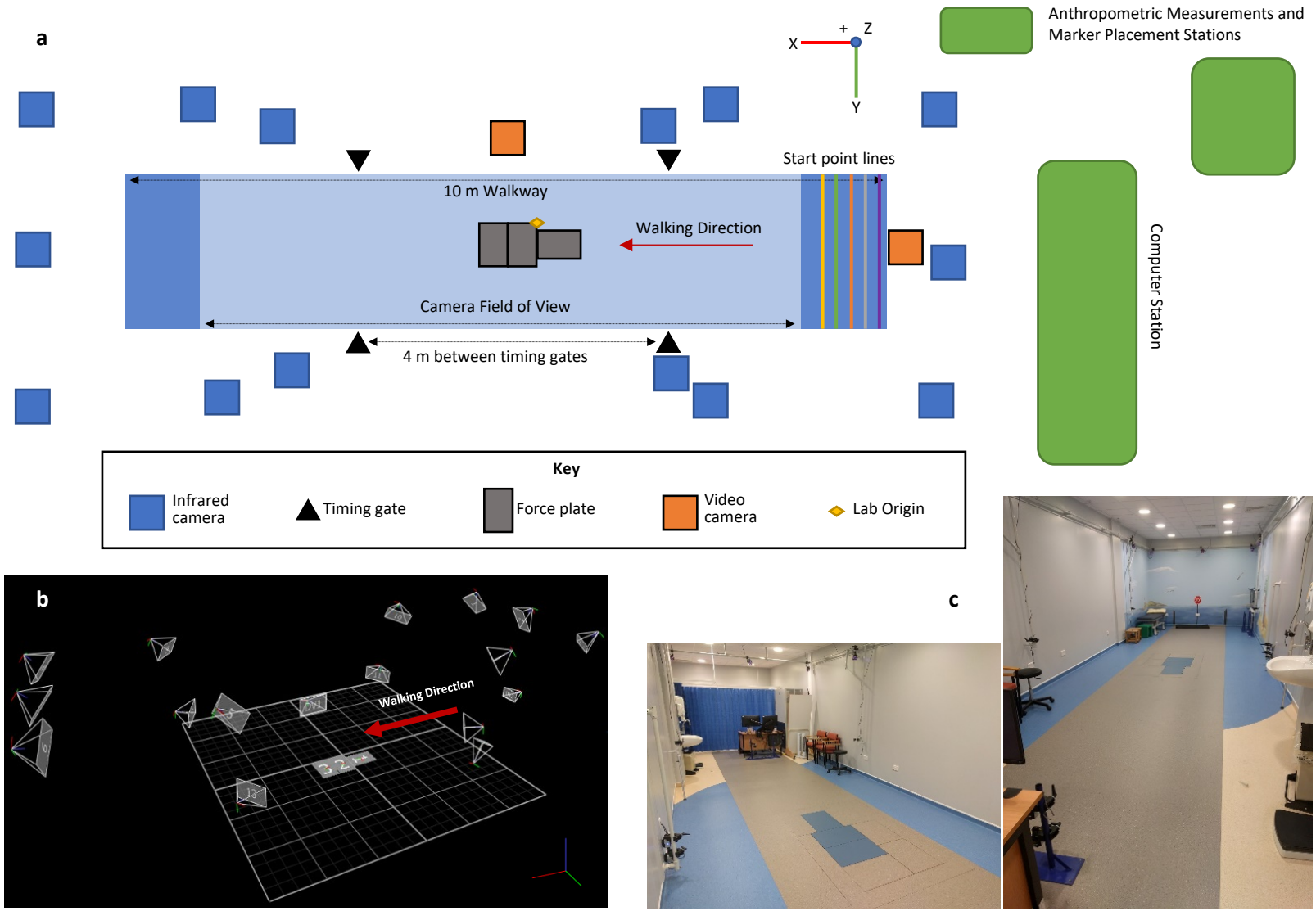

Figure 7.2 *Schematic diagram of experimental set-up (a), 3-D viewpoint (b) and images of the Gait Laboratory set-up for data collection (c)*

7.2.2 Anthropometric Measurements

Anthropometric measurements, performed by AD, were obtained for each participant to the nearest millimetre using an anatomical tape measure and digital callipers (RS PRO 150mm Digital Calliper, RS Components, UK). The following measurements were input into Vicon's proprietary software (Vicon Nexus 2.10.1, Vicon Motion Systems, Oxford, UK) to facilitate scaling and biomechanical modelling of segments with the PiG Model (Vicon, 2017): height (mm), body mass (kg), shoulder offset (mm), elbow width (mm), wrist width (mm), hand thickness (mm), leg length (mm), knee width (mm) and ankle width (mm). PiG automatically calculates other anthropometric parameters required by the biomechanical model (**Appendix n**).

7.2.3 Three-dimensional Motion Capture

7.2.3.1 Calibration Procedure

In accordance with manufacturer guidelines, the system was initialised and cameras were given at least one hour to warm-up prior to each data collection to ensure temperature variations did not affect accuracy. A dynamic calibration was then performed by waving a calibration wand (**Figure 7.3**) to determine the capture volume. A systematic approach was adopted for the calibration procedure; always starting at ground level by the force plates and covering the capture volume within the cameras' FOV (6 x 6 x 2 m). Once all cameras had processed the required number of samples in Full Calibration mode, a static calibration was performed to determine the laboratory global coordinate system. This was achieved by placing the calibration wand on the corner of force plate two (**Figure 7.2a**). The global Z axis defined the vertical, the global X axis defined the antero-posterior axis (direction of walking) and the global Y axis defined the medio-lateral axis. Residual errors of less than 2 mm were accepted for each camera.

Figure 7.3 *Calibration Wand (Vicon Active Wand v2)*

7.2.3.2 Plug-in Gait Model

Thirty-five passive retro-reflective markers (14 mm diameter with a 17 mm hard plastic base and 3 mm thread) were attached to anatomical landmarks (**Appendix o**) on the participant according to the PiG Full Body Marker Model (**Figure 7.4**). Participants wore tight underwear only, allowing markers to be directly attached to the skin, eliminating marker movement artefacts caused by clothing. To eliminate inter-rater error, marker placement and data collection for all participants were performed by the same researcher (AD), who was experienced in 3-D motion capture and marker placement (8 years of experience). Anatomical landmarks were located by palpating the joint area. Although accuracy is inevitably affected by marker movement artefacts caused by the interposition of soft tissues between the markers and bony landmarks (Gao and Zheng, 2008; Leardini et al., 2005; Della Croce et al., 2005), markers were placed precisely on the bony landmarks where the skin is thin and the markers move with the underlying bony structure (Dujardin et al., 1997). Precise marker placement was paramount as large errors can result from misplacement. For example, misplacement of the lateral epicondyle knee marker by 5 mm can cause an error of 2° in knee angles (Szczerbik and Kalinowska, 2011). Whilst the researcher applying the markers was experienced, ten familiarisation sessions prior to the first data collection session were undertaken to ensure anatomical landmarks could be reliably identified.

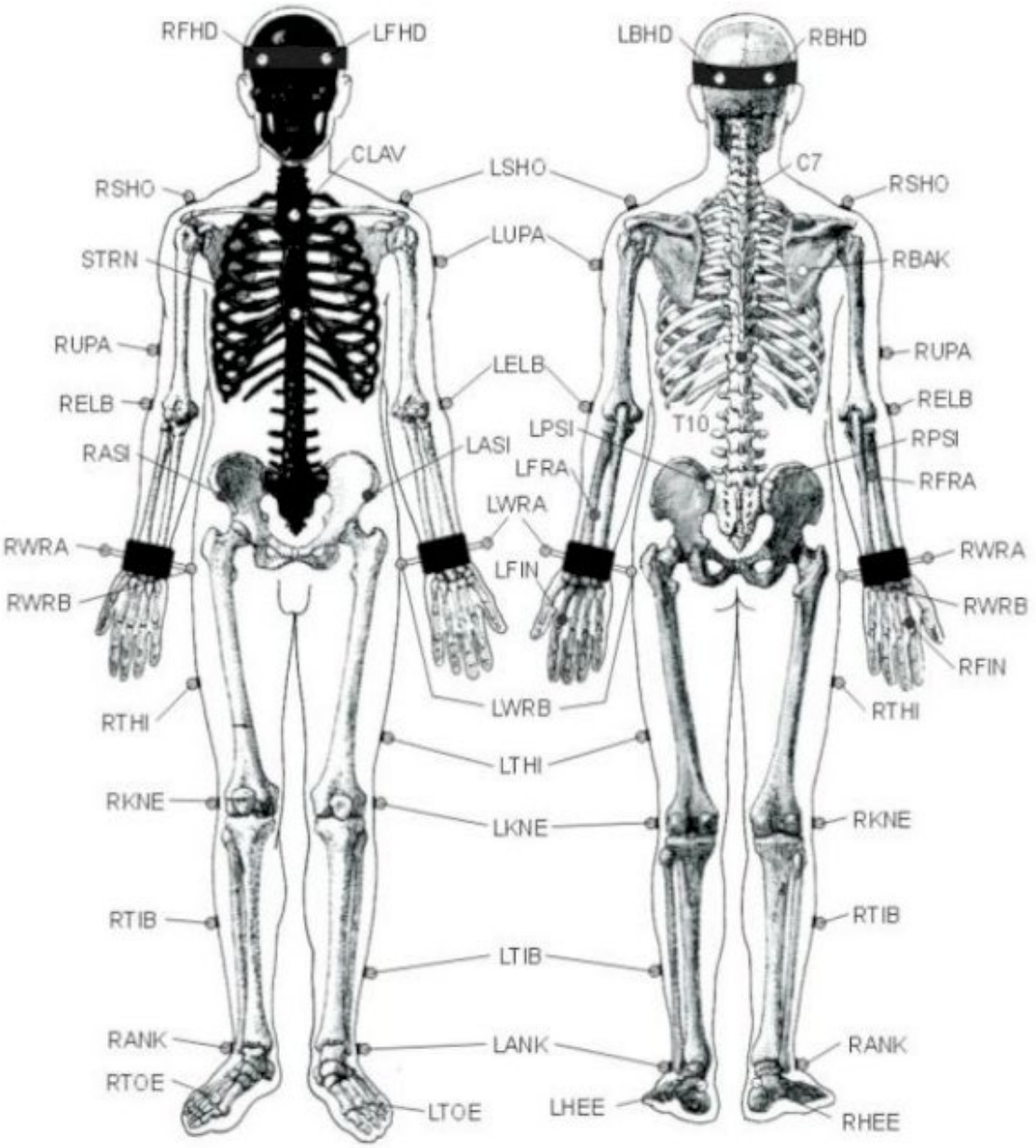

Figure 7.4 *Marker placements for the Plug-in Gait Marker Model*

PiG requires four assumptions to be met for the model to be successfully applied: 1) Subject parameters are input and the minimum required markers are present for the lower body model (pelvis markers) and the upper body model (thorax markers); 2) Static values of each gait trial are needed for the definitions of the segments; 3) Rigid segment positions are defined on a frame-by-frame basis. Each segment is defined by an origin in the global coordinate system and three orthogonal axis directions. These local axes are defined from two directions derived from the marker data using a right-handed Cartesian coordinate system, whereby the principal direction is used to establish one of the axes in the segment, the second direction is subordinate to the first and in conjunction with it

defines a plane and the third axis is perpendicular to this plane; 4) Model outputs are calculated based on frame-by-frame positions of the segments.

7.2.3.2.1 Definition of the Trunk and Pelvis

The PiG model defines the trunk in three dimensions using a Cartesian coordinate system (**Figure 7.5**). The Z axis is the primary axis, pointing downwards along the longitudinal axis and perpendicular to the transverse plane. It is calculated from the midpoint between the 7th cervical spinous process (C7 marker) and the sternal notch (CLAV marker) to the midpoint between the 10th spinous process of the thoracic spine (T10 marker) and xiphoid process of the sternum (STRN marker). The secondary direction is the X axis, which points forward along the sagittal axis and is perpendicular to the coronal plane. It is calculated from the midpoint between the C7 marker and T10 marker to the midpoint of the CLAV and STRN markers. The resulting Y axis points leftwards perpendicular to the X and Z axes and also to the sagittal plane. The origin is then calculated from the CLAV marker with an offset of half a marker diameter backwards along the X axis (Vicon, 2017).

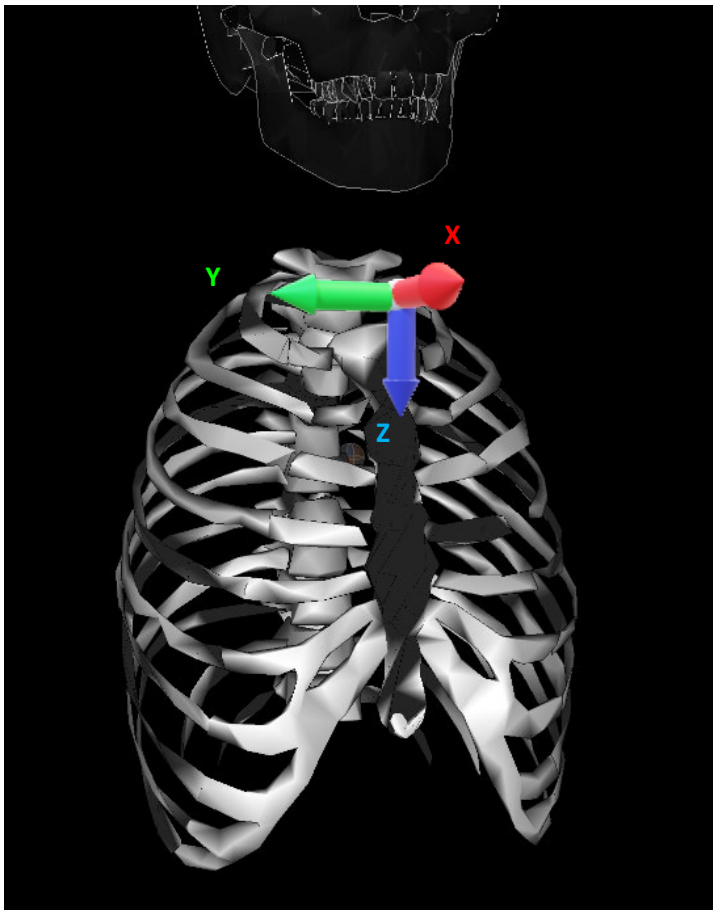

Figure 7.5 *Trunk segment displayed with its local Cartesian coordinate system. Segment created in Vicon Polygon (version 4.4.5, Vicon Motion Systems, Oxford, UK)*

The dominant axis of the pelvis segment is the Y axis, with a direction derived from the RASI marker to the LASI marker. The secondary direction is derived using the mean of the left and right posterior superior iliac spines (LPSI and RPSI markers) to the RASI marker. The position and scale of the pelvis is determined by the LASI and RASI markers, since they also determine the origin of the orientation of the pelvis in the coronal plane (Vicon, 2017). The LPSI and RPSI markers determine the anterior tilt of the pelvis segment. The Z axis is upwards along the longitudinal axis whilst the X axis is forwards. The LASI and RASI markers are used to calculate the lateral positions of the hip joint centres within the pelvis segment (Vicon, 2017). The origin of the pelvis is initially taken as the midpoint between the left and right anterior superior iliac spines (LASI and RASI markers), which is then shifted to the midpoint of the hip joint centres once they are defined (**Figure 7.6**). A high degree of accuracy is required when positioning the LASI and RASI markers since they affect the determination of the femur segments which impacts on the angles of the hip and knee joints (Vicon, 2017).

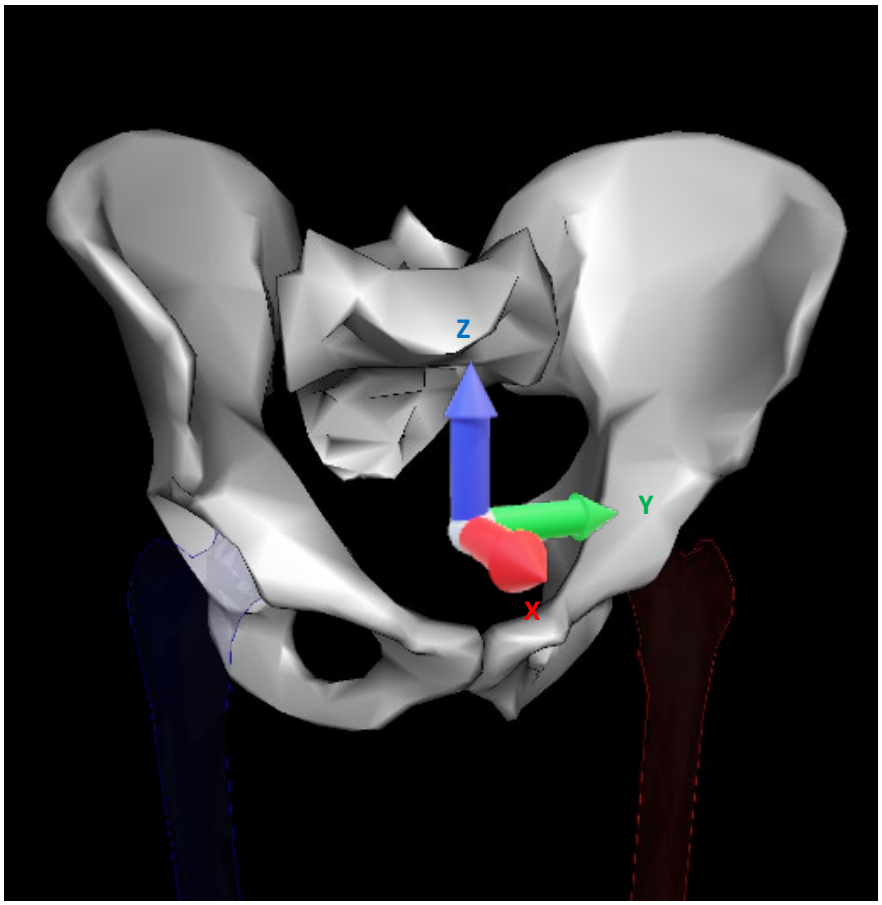

Figure 7.6 *Pelvis segment displayed with its local Cartesian coordinate system. Segment created in Vicon Polygon (version 4.4.5, Vicon Motion Systems, Oxford, UK)*

7.2.3.2.2 Kinematic Model

Segments were modelled as rigid bodies. Three non-colinear points were used to define the orientation of a segment. Using the relative orientation of two segments, joint kinematics were calculated from Cardan/Euler angles (YXZ) (Kadaba, Ramakrishnan and Wootten, 1990). PiG uses Euler angles to calculate joint angles allowing clinical descriptions of motion (Vicon, 2017). Euler angles are typically used to describe relative rotations of one segment with respect to another reference segment in three-dimensions (Chao et al., 1983). These angles are a set of three finite rotations that describes the sequence of rotations used to achieve the final orientation of a segment from the reference orientation (Kadaba, Ramakrishnan and Wootten, 1990). The rotation order sequence was flexion/extension followed by abduction/adduction then internal/external rotation, which has been recommended as standard reporting of joint actions (Cole et al., 1993). Euler angles can be represented as absolute rotations relative to the global reference frame (laboratory axes) and as relative rotations (Vicon, 2017). To calculate relative Euler angles, a set of orthogonal embedded axes need to be defined in the moving segment and the reference segment (Kadaba, Ramakrishnan and Wootten, 1990). PiG uses embedded axes to calculate joint kinematics in the coronal and transverse planes (Vicon, 2017). Pelvis and trunk-G kinematics were expressed in the global reference frame, whilst trunk-P kinematics were expressed as the relative angle between the trunk and pelvis in the pelvic reference frame. Anterior/posterior movement of the trunk-G and pelvis referred to latero-lateral rotation resulting in tilting in the sagittal plane. Flexion/extension described the sagittal plane motion of the trunk-P. Lateral tilt and obliquity referred to trunk-G and pelvic movements in the coronal plane, respectively. Lateral flexion referred to trunk-P movement in the coronal plane where ipsilateral flexion described movement towards the reference limb (i.e. first limb to contact the force plate) and contralateral flexion described movement away from the reference limb toward the contralateral limb. Finally, axial rotations of the trunk-G, trunk-P and pelvis occurred in the transverse plane. Motion in this plane was described as either protraction or retraction, where protraction was defined as rotation away from the reference limb whilst retraction was rotation toward the reference limb (Sartor et al., 1999).

7.2.3.2.3 Kinetic Model

Net joint moments (NJMs) were calculated using an inverse dynamics approach to solve the equations of motion. Kinetic calculations were based on external forces (GRF), kinematic data and segmental inertial properties, such as a segment's mass, moment of inertia and radii of gyration. The assumptions for NJM calculations where gravity and force plate measurements were the only external forces in the system and segment masses, COG and radii of gyration were known (Dempster, 1955). NJMs order

sequence was flexion/extension, lateral flexion then axial rotation. Analogous to Sadeghi, Allard and Duhaime (2000), mechanical joint powers were expressed in the three anatomical planes rather than as the scalar product of joint moments and angular velocities. Whilst mathematically power is a scalar quantity, expressing power as three components enables a better understanding of the biomechanical actions of the trunk musculature. For example, when expressed as a scalar it is assumed that most of the power generated at a joint will be in the plane corresponding to the direction of COM progression. As trunk kinetics are not well characterised during normal walking gait, expressing joint powers as 3-D vector quantities may highlight otherwise obscured complex interactions and be more meaningful physiologically. The kinetic hierarchy started from the foot as this segment is in contact with the force plate (**Appendix p**).

7.2.3.3 Walking Protocol

7.2.3.3.1 Static Trials

Prior to dynamic trials, static calibration trials were completed in three different poses. Participants held a static pose for three seconds in the anatomical, fundamental and T-pose positions (**Figure 7.7**). To ensure all markers had been correctly assigned, marker labelling in Nexus software was manually performed following static calibration trials for each participant. This also enabled live tracking during dynamic gait trials which in turn allowed data quality to be visually inspect during the session (i.e. marker drop-out and ghost markers).

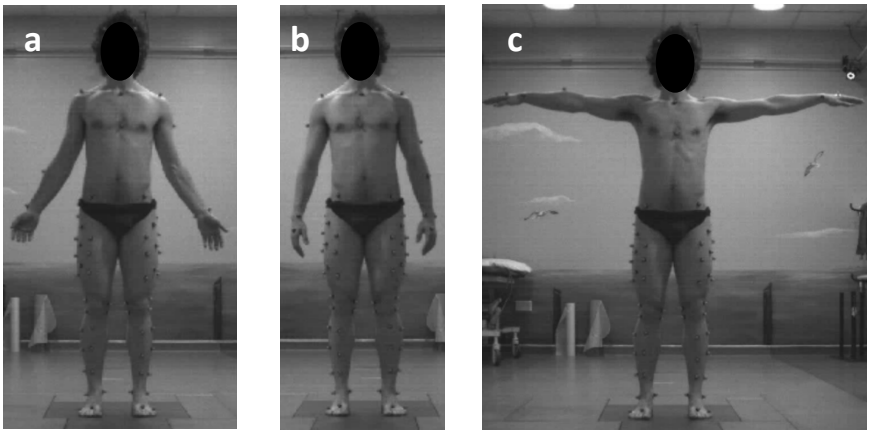

Figure 7.7 *Static calibration poses. (a) Fundamental position, (b) anatomical position and (c) T-pose position*

7.2.3.3.2 Dynamic Trials

Participants performed a series of practice trials prior to data collection trials. Each participant was instructed to walk at their usual walking speed using a specific phrase to ensure consistency between participants: "Walk as if you were walking to the shops". A successful trial was defined as a habitual steady state walk with complete force plate strikes for two consecutive foot contacts. Habitual steady state walking was determined by: 1) participants did not accelerate or decelerate from the start to finish of the walking trial and 2) each trial was similar in linear walking speed (within 5% of mean). The first foot contact with a force plate was also required to only contact the first force plate. To ensure this, participants were asked to adjust their starting position using the coloured lines at the start of the walkway (**Figure 7.2a**). This also helped to minimise force plate targeting as the purpose of start position adjustments was not obvious. A research assistant visually inspected foot contacts on the force plates, providing real-time feedback on whether any part of the foot had contacted the ground outside of the force plate area. If a participant's foot made contact with the ground outside of the force plates or the first foot contact with force plate one was outside of its area, the trial was discounted. Once participants had established their optimal starting position and successfully completed three consecutive practice trials, data collection trials were performed. Using the same criteria as the practice trials, three successful trials were required to complete data collection for each participant. If a marker fell off during a trial, the trial was discounted and the marker replaced. Static calibrations were also repeated at the end of the data collection to ensure replaced markers were accurately repositioned.

7.2.3.4 Data Processing

All gait trials were processed in Vicon Nexus. Standard Vicon Nexus operations were used to perform preliminary marker reconstructions and auto-labelling. Data quality (i.e. unused markers, total gaps, markers labelled and correct labelling) for each trial was inspected. Manual labelling was performed where necessary. Marker trajectory gap filling was initially performed using the Woltring quintic spline filter on up to five samples joined by linear interpolation. Gaps in marker trajectories of the head, thorax and pelvis segments were filled using the Rigid Body function on up to 25 frames. Remaining gaps of up to 10 frames were interpolated using the Pattern Fill function, which uses the shape of another trajectory of a similar motion without a gap to fill the selected gap. Marker trajectory and force plate data were then smoothed using a low-pass 4^{th} order zero-lag Butterworth filter with a 10 Hz cut-off frequency.

Careful consideration was taken to select the optimal cut-off frequency. Whilst choosing the optimal filter is typically an interactive trial and error process which is best accomplished through visual inspection (van den Bogert, 1996), attempts were made to support decisions numerically. Therefore, power spectral density analysis of marker trajectories was performed to examine the cumulative content of the signal in the frequency domain. Based on the recommendations of Sinclair and colleagues (Sinclair et al., 2013a; Sinclair, Taylor and Hobbs, 2013), optimal cut-off frequency was chosen as the frequency at which 99% of the signal power was contained below (Sinclair et al., 2013b, 2013a). The analysis was performed using a custom-built MATLAB script (R2019a version 9.6.0.1072779, The MathWorks, Inc., Natick, MA, US) (**Appendix q**). 10 Hz was identified as the optimal cut-off frequency across marker trajectories for each participant. This was supported by the original decision for a cut-off of approximately 10 Hz based on visual determination. Furthermore, this cut-off frequency is able to attenuate noise without distorting high-frequency marker movement at ground contact (Sinclair, Taylor and Hobbs, 2013). Analogue GRF data were filtered using the same recursive Butterworth filter. Whilst the necessity to filter force plate data is less than marker position data as force data is not differentiated in the inverse dynamics equations, several researchers have indicated that artefacts are created when different cut-off frequencies are used to filter force and position data in inverse dynamics (Bisseling and Hof, 2006; Kristianslund, Krosshaug and Van den Bogert, 2012; Bogert and Koning, 1996). To avoid artefacts in the NJM curves, raw GRF data were also filtered with the same cut-off frequency of 10 Hz. An example of the effect of filtering is shown in **Appendix r**.

Following data filtering, GC events (IC and TO) were detected using a sub-routine in Nexus. This operation checks for the vertical GRF crossing the threshold value, which was applied at 20 N, when the ankle and toe markers are within the force plate boundary. Trials were visually inspected using the video recordings to verify the correct timings of these events. GC events were manually adjusted where necessary. Where IC occurred off a force plate (i.e. second heel contact of the first limb), GC events were manually identified using visual inspection of the video data. GC parameters were calculated following this process. The “Process Dynamic PiG Model” pipeline was then run to generate the PiG biomechanical model, which outputs joint kinematics and kinetics and defines them in terms of their order and sign conventions (Vicon, 2017). Each trial was manually truncated to include one full GC for each limb before the trial (C3D and VSK files) was saved and data were exported in C3D format for analysis in Polygon (Version 4.4.5, Vicon Motion Systems, Oxford, UK).

7.2.4 Data Analysis

Analysis of processed gait trials was performed in Vicon Polygon and data were exported as an ASCII file (CSV format) for further analysis in Microsoft Excel (Microsoft® Excel ® for Office 365, version 1908, Tokyo, Japan). All outcomes were normalised to one GC (100%) using linear interpolation to 101 data samples in Polygon. Each trial consisted of a left and right limb GC and three trials were processed for each participant. Therefore, kinematic and kinetic peak values as well as spatiotemporal parameters were averaged across six GCs for each participant. Ensemble-averages were also generated across trials and participants for trunk, pelvis and lower back kinematics and kinetics as well as GRFs.

7.2.4.1 Spatiotemporal Parameters

Usual gait parameters generated in Polygon included: Cadence (steps·min^{-1}), Step Time (s), Stride Time (s), Single Support Time (s), Double Support Time (s), Foot Off (% of GC – referred to as % from here), Opposite Foot Contact (%), Opposite Foot Off (%), Step Length (m), Stride Length (m), Step Width (m). Walking speed (m·s^{-1}) was calculated using the Brower timing gates. The timing gates were positioned 4 m apart. This distance was then divided by time taken to walk through the timing gates to derive walking speed. Normalised Step Length and Normalised Stride Length were calculated in Microsoft Excel by dividing mean Step Length and Stride Length of each participant by their height (m). Spatiotemporal parameters with definitions are provided in **Table 7.5**.

Table 7.5 *Spatiotemporal parameters with definitions*

Spatiotemporal Parameter	Definition
Cadence	Number of steps taken per minute
Step Time	Time between contralateral and the following ipsilateral foot contact
Stride Time	Time between successive ipsilateral foot strikes
Single Support Time	Time from contralateral foot off to contralateral foot contact
Double Support Time	Time from ipsilateral foot contact to contralateral foot off plus time from contralateral foot contact to ipsilateral foot off
Foot Off	Percentage of the gait cycle of ipsilateral foot off
Opposite Foot Contact	Percentage of the gait cycle of contralateral foot contact
Opposite Foot Off	Percentage of the gait cycle of contralateral foot off
Walking Speed	Distance between timing gates divided by the time taken to walk between them
Step Length	Distance from ipsilateral toe marker position to contralateral toe marker position
Normalised Step Length	Step length normalised to height
Stride Length	Distance from ipsilateral toe marker position at first and second ipsilateral foot contacts
Normalised Stride Length	Stride length normalised to height
Step Width	Distance from contralateral toe marker position onto the first and second ipsilateral foot contacts

7.2.4.2 Kinematic Outcomes

Joint kinematics were generated for the trunk and pelvis segments relative to the global coordinate system in the sagittal plane (anterior/posterior tilt), coronal plane (obliquity) and transverse plane (axial rotation). The relative rotations between the pelvis and trunk segments were also calculated to provide joint kinematics for the lower back in the sagittal (flexion/extension), coronal (lateral flexion) and transverse (internal/external rotation) planes. Mean peak minima and maxima joint and segment angles were obtained during the GC for each participant. Mean ROM was also obtained for the trunk, pelvis and lower back throughout the GC for each participant.

7.2.4.3 Kinetic Outcomes

Lower back joint moments reflected the NJMs between the pelvis and trunk (Vicon, 2017). PiG derives NJMs from the local coordinate frame of the distal segment in the hierarchical kinetic chain (Vicon, 2017). This meant that Vicon Polygon calculated joint moments in the external perspective. The net external moment is counterbalanced by the net internal moment produced by the muscles predominantly. Mathematically, the internal moment is equal and opposite to the external moment

(Derrick et al., 2020). Whilst there is no fundamental difference in adopting an internal or external perspective, the decision should be made based on the researcher's view of the source of the moments (Derrick et al., 2020). Furthermore, describing joint moments in the external reference frame is not commonplace in biomechanical analysis. Therefore, in accordance with biomechanical convention and the fact that NJMs were considered to be the result of muscular actions, an internal perspective was adopted, and moment data were adjusted accordingly. Ensemble averages for lower back NJMs were calculated in the sagittal, coronal and transverse planes. Mean peak flexion/extension lower back moments were obtained at Loading Response (LR), Midstance (MS), Terminal Stance (TS), Pre-Swing (PSw), Initial Swing (ISw) and Terminal Swing (TSw). Mean peak minima and maxima lateral flexion and axial rotation moments were also obtained during the stance and swing phases.

Mechanical joint powers for the trunk-P were calculated in Microsoft Excel from the dot product of the moment vector (M) and the joint angular velocity vector (ω). Trunk-P powers were expressed in the three Cardinal planes to make the results more meaningful physiologically. Therefore, joint powers in the trunk-P were given by the following equations:

$$P_x = M_x \omega_x$$

$$P_y = M_y \omega_y$$

$$P_z = M_z \omega_z$$

Ensemble-averages were calculated for flexion/extension, lateral flexion and rotational lower back power. From these curves, key instances of power generation/absorption were identified. In the sagittal plane, S1 denoted a peak of lower back power generation during LR, S2 represented peak power absorption during MS and S3 was identified as the power generation peak during PSw. In the coronal plane, S4 was peak power absorption during LR, S5 was peak power generation at PSw and S6 represented peak power absorption during swing phase. No key instances were identified for axial rotation power. The power curve exhibited local minima and maxima, however, amplitudes were very low indicating that the contribution of lower back rotational power was negligible. Finally, the time integral of the power curves (i.e. work) (**Figure 7.8**) was calculated in Microsoft Excel using the Trapezium Rule. Total positive work was calculated as the sum of positive areas under the curve contained by the X axis. Total negative work was the sum of negative areas above the curve contained by the X axis. Total positive and negative work was obtained in all three planes. Joint moments, powers and work were normalised to body mass of the participant.

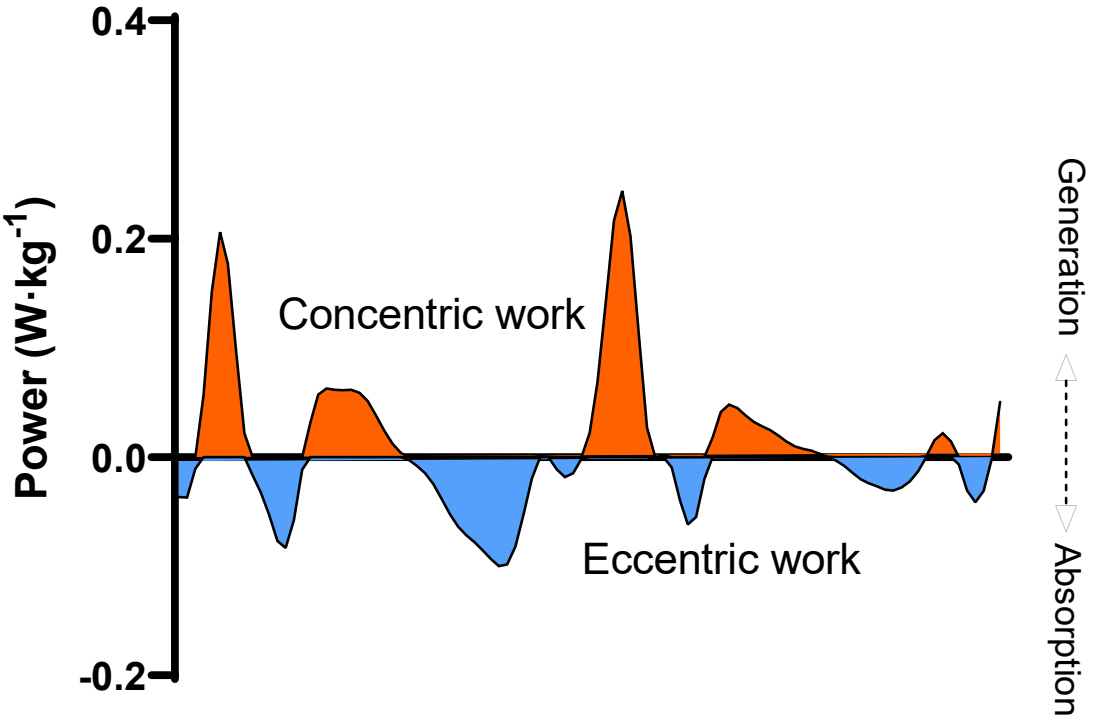

Figure 7.8 *Graphical representation of the power curve time integral. Orange area represents positive work performed and the blue area represents negative work performed. The sum of positive areas gave total positive work during the gait cycle, whilst the sum of negative areas gave total negative work across the gait cycle.*

FD of the lower back was calculated in Microsoft Excel by dividing the joint moment during gait by the peak isokinetic MVC moment (see **Chapter 6**). FD was calculated in the sagittal plane only as trunk isokinetic dynamometry was limited to flexion/extension movements. The mean peak flexion/extension moments of the lower back during LR, MS, TS, PSw, ISw and TSw were identified on the moment-phase curves for each participant. These curves were synchronised with the corresponding power-phase curves to determine the neuromuscular action of the lower back (e.g. extensors concentrically activated) during LR, MS, TS, PSw, ISw and TSw. The moment-phase curves for each participant were then matched up to the corresponding angular velocity-phase curves to determine the angular velocity of the lower back when the identified peak moments occurred. This information was then used to select peak moments from the most appropriate isokinetic condition (i.e. the isokinetic test condition that mimicked the lower back neuromuscular action and movement during the corresponding phase of the GC). For example, if a participant's lower back extensor muscles were generating power during LR at $30°\cdot s^{-1}$, the concentric extension isokinetic test at $30°\cdot s^{-1}$ was sought. It should be noted that $15°\cdot s^{-1}$ and $30°\cdot s^{-1}$ were the most typical trunk angular velocities and this did not differ by age. The participant's peak moment during LR was then divided by their peak

isokinetic moment and expressed as a percentage. Therefore, if the participant produced a moment of 1 Nm·kg^{-1} at LR and their peak isokinetic moment was 4 Nm·kg^{-1}, the FD would be 25 %. If the demand and capacity were equal, the FD would be 100 %. For each of the identified phases (LR, MS, TS, PSw, ISw and TSw), individual FD values were calculated and averaged across age groups. FD was given by the following equation:

$$Functional\ Demand = \frac{Moment\ produced\ during\ the\ movement\ in\ the\ gait\ cycle}{Maximum\ available\ isokinetic\ moment}$$

7.2.5 Statistical Analysis

Statistical analyses were performed using SPSS (SPSS® for Windows Version 24.0, IBM Corp, Armonk, NY, US) and MATLAB (R2019a, version 9.6.0.1072779, The MathWorks, Inc., Natick, MA, US). Graphical presentation was performed using GraphPad Prism (Version 8.3.1, San Diego, CA, US). Data are presented as means with standard deviations (mean ± SD) unless otherwise stated.

7.2.5.1 Discrete Variables

For spatiotemporal parameters, ROM data, kinematic and kinetic peaks, independent samples t-tests were performed to compare statistical differences between the OG and YG. Muscle morphology and PA covariates were assessed using univariate ANCOVA. Potential covariates were mean MFI and total NMV across all of the measured paravertebral muscles as well as VPA. For each discrete variable, differences between groups were analysed and reported using the ANCOVA test if a significant covariate was found. Zero-order and partial correlations, controlling for age group, were also performed between ROMs in the trunk-G, trunk-P and pelvis. Alpha level was set at 5% for all statistical tests and effect sizes (Cohen's d) calculated where appropriate. All data were normally distributed, as assessed by Shapiro-Wilks test ($p > .05$). Where the assumption of homogeneity of variances was violated, as assessed by Levene's Test of Equality of Variances ($p < .05$), the Welch-Satterthwaite correction was used.

7.2.5.2 Continuous Variables

Age group differences in kinematic and kinetic waveforms were compared using Statistical Parametric Mapping (SPM) (Friston et al., 2007), which was performed in MATLAB R2019a (v. 9.6.0) and implemented using the open-source one-dimensional Statistical Parametric Mapping code (spm1D-

package, version 0.4.3, http://spm1d.org/index.html) (**Appendix s**). Rather than basing age-related differences in gait on discrete information (i.e. peaks and ROM), SPM provides a novel method to investigate the phase-specific effect of age during the GC. Specifically, SPM two-tailed independent t-tests were used to examine whether mean trunk, lower back and pelvis angle waveform patterns differed significantly ($\alpha = 0.05$) between the age groups. Lower back joint moment and power waveforms as well as GRF waveforms were also compared between groups. For each SPM t-test, a statistical parametric map (SPM{t}) was created by calculating the SPM{t} test statistic separately at each individual point in the normalised time series (Pataky, Vanrenterghem and Robinson, 2017). To test the null hypothesis, Random Field Theory determined the critical threshold at which 5 % of smooth random curves would be expected to traverse. Field smoothness was based upon estimates of temporal gradients of the data residuals to determine statistical significance (Friston et al., 2007; Pataky, Vanrenterghem and Robinson, 2017). If SPM{t} crossed the critical threshold, a supra- or infra-threshold cluster depicted by grey shading indicated a significant difference ($p < .05$) between groups at a specific phase in the GC. For all age-group comparisons, SPM{t} inference and cluster properties were provided.

7.3 Results

7.3.1 Spatiotemporal Parameters

Opposite foot contact occurred significantly later in the GC for the YG than the OG ($t(22) = 2.22$, $p = .037$) and stride length was significantly shorter in the OG ($t(22) = 2.11$, $p = .047$), however the difference in stride length between age groups became non-significant when it was normalised to participants' height ($p > .05$). The difference between the OG's mean and YG's mean was large for opposite foot contact (Cohen's d = 0.91) (**Table 7.6**). The difference in walking speed between the OG and YG was moderate and not significant ($p > .05$).

Table 7.6 *Spatiotemporal parameters (mean ± SD) for the younger and older groups during normal gait*

Parameter	Young group (n = 12)	Old group (n = 12)	Independent t-test	Cohen's d
Rhythm				
Cadence (steps·min^{-1})	113.9 ± 5.8	113.0 ± 7.2	$t(22) = 0.34$, $p = .74$	0.14
Step Time (s)	0.53 ± 0.03	0.53 ± 0.04	$t(22) = -0.40$, $p = .70$	0.16
Stride Time (s)	1.06 ± 0.05	1.07 ± 0.07	$t(22) = -0.41$, $p = .68$	0.17
Single Support Time (s)	0.41 ± 0.02	0.42 ± 0.02	$t(22) = -0.66$, $p = .51$	0.27
Phases				
Double Support Time (s)	0.22 ± 0.03	0.22 ± 0.04	$t(22) = -0.12$, $p = .91$	0.05
Foot Off (%)	59.78 ± 1.00	60.16 ± 1.16	$t(22) = -0.85$, $p = .41$	0.35
Opposite Foot Contact (%)*	50.39 ± 0.32	50.16 ± 0.17	$t(22) = 2.22$, $p = .037$	0.91
Opposite Foot Off (%)	11.46 ± 0.93	10.87 ± 1.59	$t(22) = 1.10$, $p = .28$	0.45
Pace				
Walking Speed (m·s^{-1})	1.45 ± 0.19	1.33 ± 0.16	$t(22) = 1.70$, $p = .10$	0.69
Step Length (m)	0.76 ± 0.08	0.71 ± 0.06	$t(22) = 1.85$, $p = .08$	0.76
Normalised Step Length	0.43 ± 0.05	0.41 ± 0.03	$t(19.7) = 1.43$, $p = .17$	0.59
Stride Length (m)*	1.53 ± 0.15	1.41 ± 0.12	$t(22) = 2.11$, $p = .047$	0.86
Normalised Stride Length	0.86 ± 0.09	0.81 ± 0.06	$t(22) = 1.64$, $p = .12$	0.67
Base of Support				
Step Width (m)	0.14 ± 0.03	0.16 ± 0.03	$t(22) = -1.22$, $p = .23$	0.50

* significant age effect

7.3.2 Kinematic Parameters

There were significant age-effects in trunk-G kinematics, predominantly in the transverse plane. The YG had significantly greater trunk-G ROM than the OG in the sagittal (2.96 ± 0.88° vs 1.99 ± 0.39°, $t(22) = 3.49$, $p = .002$) and transverse planes (6.87 ± 2.21° vs 5.04 ± 1.29°, $t(17.7) = 2.48$, $p = .024$) during the GC. Peak trunk-G protraction and retraction were also significantly greater in the YG compared to the

OG (**Table 7.7**). All significant differences were large according to effect size estimates (Cohen's d = 0.9 – 1.4). Whilst non-significant, age-group differences in maximum and minimum trunk-G anterior tilt, flexion/extension and axial rotation ROM in the trunk-P, peak trunk-P protraction and peak pelvic axial rotations were moderate to large (Cohen's d = 0.54 – 0.84).

Table 7.7 *Trunk and pelvic kinematic peaks (mean ± SD) for the young and old groups during normal gait*

Parameter	Young group (n = 12)	Old group (n = 12)	Independent t-test	Cohen's d
Trunk-G (°)				
Antero-posterior Tilt ROM**	2.96 ± 0.88	1.99 ± 0.39	$t(22) = 3.49$, p = .002	1.43
Max Anterior Tilt	5.08 ± 3.21	7.69 ± 6.10	$t(22) = -1.31$, p = .20	0.54
Min Anterior Tilt	2.12 ± 3.34	5.70 ± 6.29	$t(22) = -1.74$, p = .10	0.71
Lateral Tilt ROM	4.78 ± 2.28	4.95 ± 2.82	$t(22) = -0.16$, p = .88	0.06
Contralateral Flexion	2.30 ± 1.18	2.38 ± 1.44	$t(22) = -0.15$, p = .88	0.06
Ipsilateral Flexion	-2.48 ± 1.11	-2.56 ± 1.39	$t(22) = 0.16$, p = .87	0.07
Axial Rotation ROM*	6.87 ± 2.21	5.04 ± 1.29	$t(17.7) = 2.48$, p = .024	1.01
Protraction Rotation*	3.42 ± 1.06	2.50 ± 0.58	$t(16.9) = 2.64$, p = .017	1.08
Retraction Rotation*	-3.45 ± 1.19	-2.54 ± 0.74	$t(22) = -2.25$, p = .035	0.92
Trunk-P (°)				
Flexion/Extension ROM	2.85 ± 1.00	2.23 ± 0.69	$t(22) = 1.76$, p = .09	0.72
Max Extension	-6.45 ± 5.32	-4.87 ± 7.79	$t(22) = -0.58$, p = .57	0.24
Min Extension	-3.61 ± 5.31	-2.64 ± 7.82	$t(22) = -0.35$, p = .73	0.14
Lateral Flexion ROM**	14.31 ± 3.08	10.22 ± 3.29	$t(22) = 3.15$, p = .005	1.29
Ipsilateral Flexion**	7.19 ± 1.50	5.19 ± 1.64	$t(22) = 3.10$, p = .005	1.27
Contralateral Flexion**	-7.13 ± 1.59	-5.03 ± 1.66	$t(22) = -3.18$, p = .004	1.30
Axial Rotation ROM	12.51 ± 3.85	9.55 ± 3.20	$t(22) = 2.05$, p = .053	0.84
Protraction Rotation	-6.21 ± 1.99	-4.78 ± 1.56	$t(22) = -1.97$, p = .062	0.80
Retraction Rotation*	6.30 ± 1.88	4.77 ± 1.65	$t(22) = 2.11$, p = .047	0.86
Pelvis (°)				
Antero-posterior Tilt ROM	2.31 ± 0.75	2.28 ± 0.65	$t(22) = 0.11$, p = .91	0.05
Max Anterior Tilt	9.72 ± 4.00	11.60 ± 4.96	$t(22) = -1.02$, p = .32	0.42
Min Anterior Tilt	7.41 ± 4.04	9.32 ± 4.97	$t(22) = -1.03$, p = .31	0.42
Obliquity ROM***	9.83 ± 2.45	5.64 ± 1.72	$t(22) = 4.85$, p < .001	1.98
Upward Tilt***	4.87 ± 1.23	2.84 ± 0.81	$t(22) = 4.77$, p < .001	1.95
Downward Tilt***	-4.96 ± 1.23	-2.80 ± 0.94	$t(22) = -4.83$, p < .001	1.97
Axial Rotation ROM	11.92 ± 4.35	8.62 ± 3.82	$t(22) = 1.98$, p = .061	0.81
Protraction Rotation	6.01 ± 2.25	4.39 ± 1.93	$t(22) = 1.90$, p = .071	0.77
Retraction Rotation	-5.91 ± 2.12	-4.23 ± 1.93	$t(22) = -2.03$, p = .054	0.83

Age effect significance values * p < .05, ** p < .01, *** p < .001; ROM = range of motion

Significant age-related differences in trunk-P kinematics were predominantly in the coronal plane although rotational kinematics also showed large differences (Cohen's d = 0.8 – 0.9). Peak trunk-P retraction was significantly greater ($t(22) = 2.11$, $p = .047$) in the YG (6.30 ± 1.88°) compared to the OG (4.77 ± 1.65°). The YG also demonstrated greater peak protraction and rotational ROM than the OG although not statistically significant. However, the magnitudes of these differences were large and comparable to that of internal rotation (**Table 7.7**). During stance, the YG (7.19 ± 1.50°) exhibited significantly greater ($t(22) = 3.10$, $p = .005$) peak contralateral flexion compared to the OG (5.19 ± 1.64°). The YG (-7.13 ± 1.59°) also demonstrated significantly greater ($t(22) = -3.18$, $p = .004$) peak ipsilateral flexion than the OG (-5.03 ± 1.66°) during swing phase, resulting in the YG possessing a lateral flexion ROM 40% greater than the OG (14.31 ± 3.08° vs 10.22 ± 3.29°, $t(22) = 3.15$, $p = .005$). Similar differences were found in pelvic obliquity where peak upward and downward tilt were significantly greater in the YG compared to the OG (**Table 7.7**). Pelvic obliquity ROM as a result was also significantly greater ($t(22) = 4.85$, $p < .001$) in the YG compared to the OG by 74%. Significant age-related differences in trunk-P lateral flexion and pelvic obliquity were very large according to effect size estimates (Cohen's d = 1.3 – 2.0).

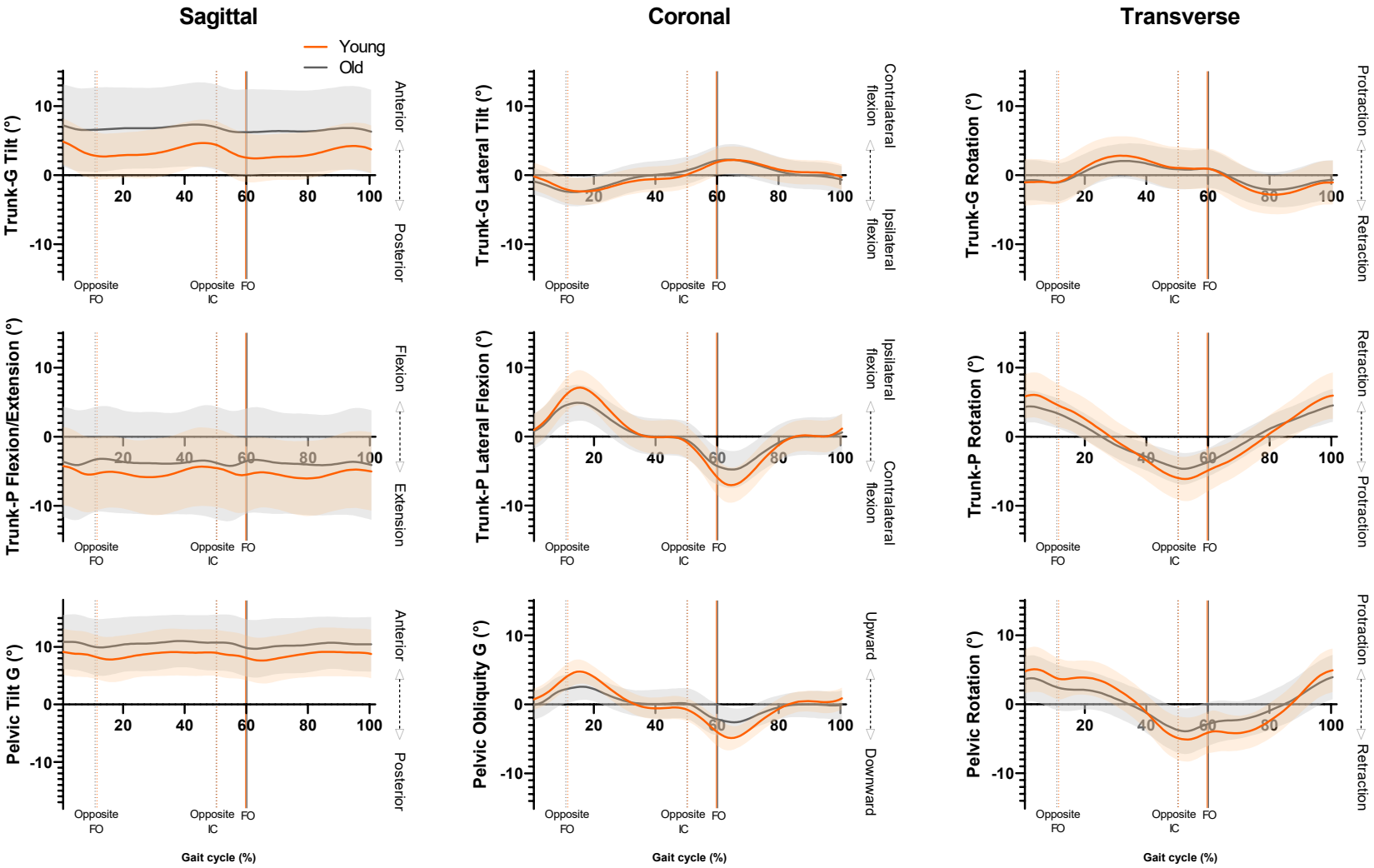

Figure 7.9 *Ensemble averages for Trunk-G (trunk relative to global reference frame), Trunk-P (trunk relative to pelvic reference frame) and Pelvis kinematics. Orange line = YG mean, grey line = OG mean, orange shaded area = YG SD, grey shaded area = OG SD*

There were numerous significant interplanar and intersegment correlations in ROMs (**Table 7.8**). However, after controlling for age group, only significant partial correlations remained between trunk-P lateral flexion ROM and trunk-G lateral tilt ROM ($r(21) = .43$, $p = .043$), trunk-P lateral flexion ROM and pelvic obliquity ROM ($r(21) = .56$, $p = .005$) and trunk-P axial rotation ROM and pelvic axial rotation ROM ($r(21) = .63$, $p = .001$). There were no significant partial correlations in the sagittal plane.

Trunk-G kinematic waveform patterns were highly similar between the older and YG throughout the GC (**Figure 7.9**). No significant SPM phase differences were revealed between groups (**Figure 7.10**).

Table 7.8 *Zero-order correlation coefficients for ROM between the trunk-P, trunk-G and pelvis in all Cardinal planes*

		Trunk-P ROM			**Trunk-G ROM**			**Pelvic ROM**		
		Flexion/ Extension	Lateral flexion	Axial rotation	Tilt	Lateral tilt	Axial rotation	Tilt	Obliquity	Axial rotation
Trunk-P ROM	Flexion/Extension	r = 1.0	r = .43*	r = .37	r = .03	r = .14	r = .41*	r = .23	r = .37	r = .27
	Lateral flexion		r = 1.0	r = .25	r = .46*	r = .33	r = .44*	r = .29	r = .73***	r = .06
	Axial rotation			r = 1.0	r = .12	r = .14	r = .45*	r = .02	r = .37	r = .69***
Trunk-G ROM	Tilt				r = 1.0	r = -.02	r = .25	r = .13	r = .46*	r = .14
	Lateral tilt					r = 1.0	r = .31	r = .01	r = -.13	r = -.14
	Axial rotation						r = 1.0	r = -.30	r = .39	r = .44*
Pelvic ROM	Tilt							r = 1.0	r = .22	r = -.02
	Obliquity								r = 1.0	r = .43*
	Axial rotation									r = 1.0

* p < .05, ** p < .01, *** p < .001

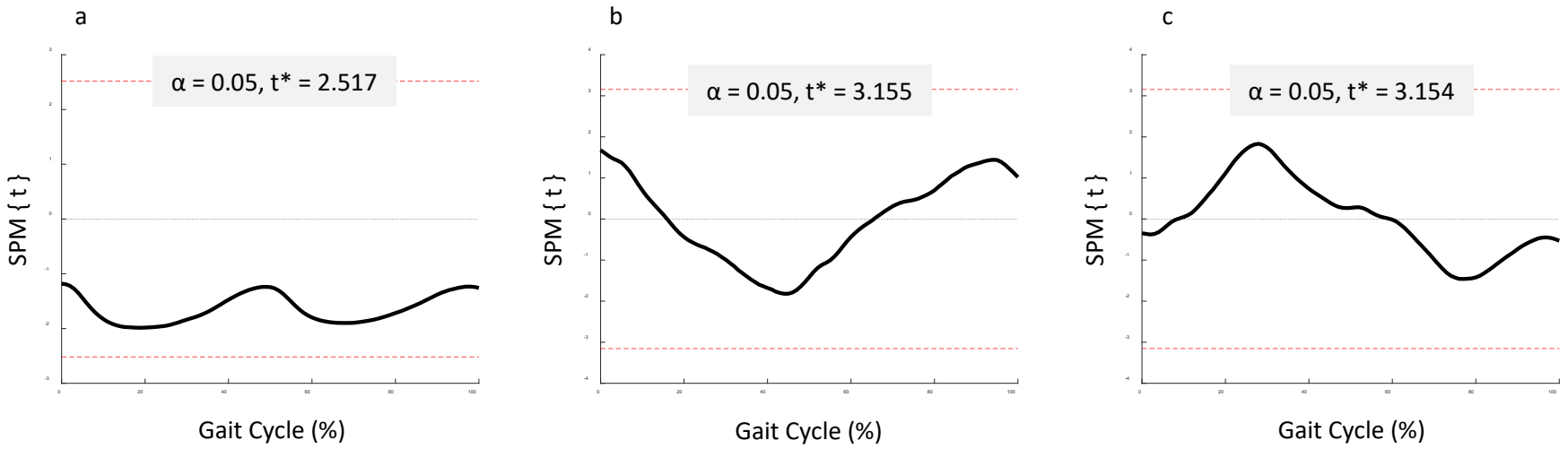

Figure 7.10 *Statistical Parametric Mapping (SPM) output for Trunk-G kinematics. a = anterior tilt, b = lateral tilt, c = rotation*

SPM phase differences between age groups were not significant for trunk-P flexion/extension (**Figure 7.11a**). Two clusters were identified for trunk-P kinematics in the coronal plane (**Figure 7.11b**). A supra-threshold cluster (13.3 – 17.7%) and an infra-threshold cluster (62.5 – 67.2%) exceeded the critical threshold indicating that lateral flexion in the YG was significantly greater during midstance and initial swing than in the OG ($t(22) = 3.247$, $p = .039$; $t(22) = 3.247$, $p = .038$, respectively). In the transverse plane, one infra-threshold cluster (66.1 – 76.9%) exceeded the critical threshold of $t(22) = 3.346$ as the YG exhibited significantly greater axial rotation than the OG during swing phase ($p = .004$) (**Figure 7.11c**).

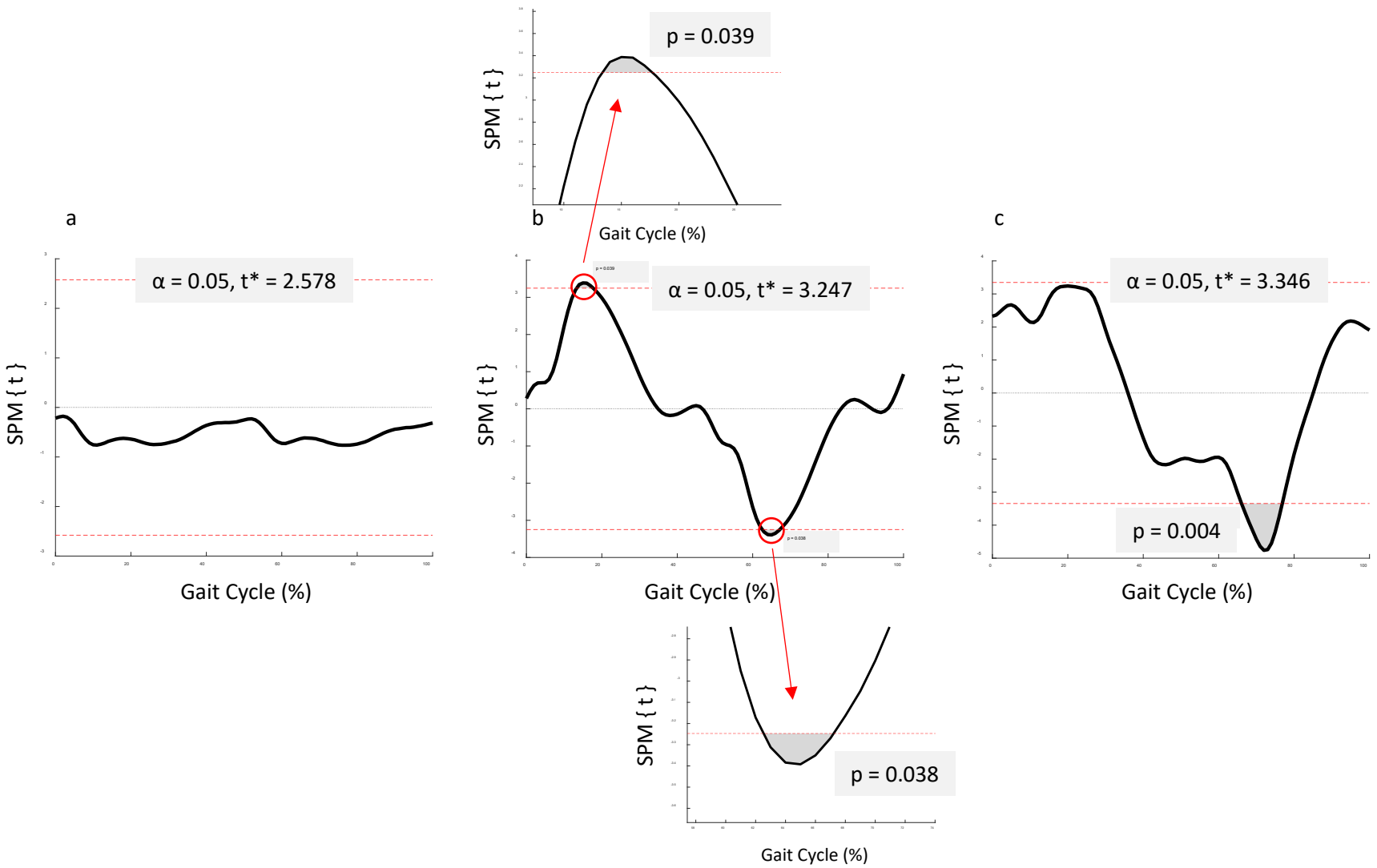

Figure 7.11 *Statistical Parametric Mapping (SPM) output for Trunk-P kinematics. a = flexion/extension, b = lateral flexion, c = axial rotation*

No significant differences were revealed for pelvic tilt between groups (**Figure 7.12a**). Two clusters were identified for pelvic obliquity (**Figure 7.12b**). A supra-threshold cluster (10.3 – 24.6%) and an infra-threshold cluster (59.4 – 73.8%) exceeded the critical threshold indicating that pelvic obliquity in the YG was significantly greater during midstance and initial swing than in the OG ($t(22) = 3.274$, $p = .003$; $t(22) = 3.247$, $p = .002$, respectively). One infra-threshold cluster (67.8 – 76.3%) exceeded the critical threshold of $t(22) = 3.222$ as the YG exhibited significantly greater pelvic rotation than the OG during swing phase ($p = .023$) (**Figure 7.12c**).

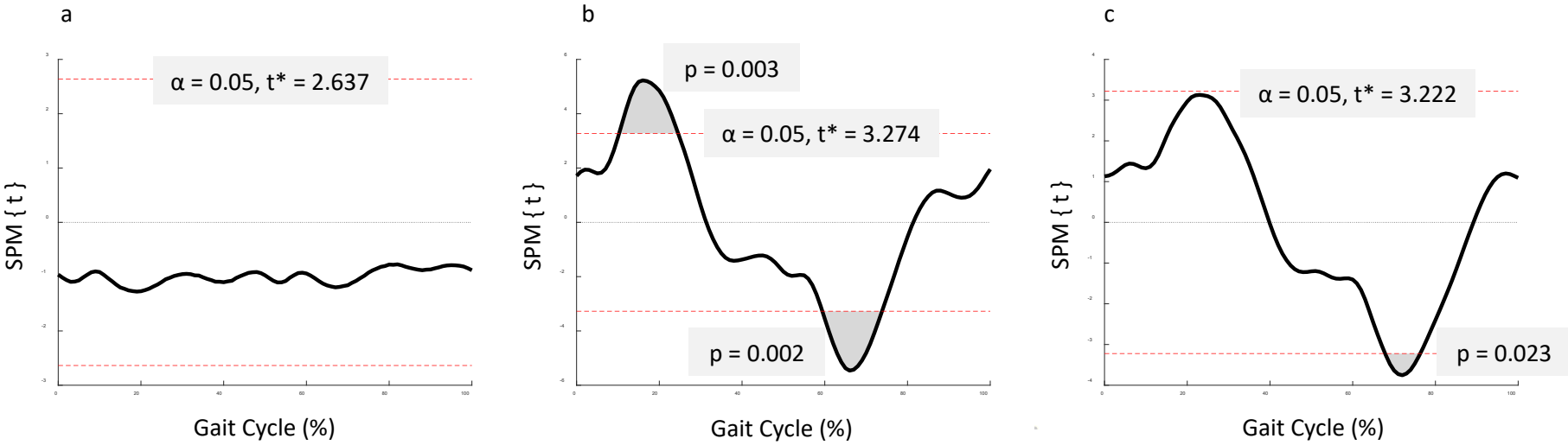

Figure 7.12 *Statistical Parametric Mapping (SPM) output for Pelvis kinematics. a = pelvic tilt, b = pelvic obliquity, c = pelvic rotation*

7.3.3 Kinetic Parameters

There were significant age effects for peak extension moment during MS ($t(22) = 2.28$, $p = .032$), peak flexion moment during TSw ($t(22) = -2.16$, $p = .042$) and peak contralateral moment during swing phase ($t(22) = -2.65$, $p = .015$). Peak spinal extension moments were on average 54.8% greater in the YG (0.96 ± 0.45 $Nm \cdot kg^{-1}$) compared to the OG (0.62 ± 0.24 $Nm \cdot kg^{-1}$) during MS, whereas during swing phase the OG produced significantly greater peak spinal flexion (1.05 ± 0.37 $Nm \cdot kg^{-1}$) and contralateral flexion moments (0.34 ± 0.11 $Nm \cdot kg^{-1}$) than the YG (0.74 ± 0.34 and 0.22 ± 0.12 $Nm \cdot kg^{-1}$, respectively). The difference in these outcomes between groups was large (Cohen's d = 0.9 – 1.1). Whilst the mean difference in peak external rotation moments did not reach significance ($t(22) = -2.02$, $p = .055$), the difference between groups was also large (Cohen's d = 0.8) (**Table 7.9**).

Table 7.9 *Lower back kinetic peaks (mean ± SD) for the young and old groups during normal gait*

Parameter	Young group (n = 12)	Old group (n = 12)	Independent t-test	Cohen's d
Peak Moments (Nm·kg^{-1})				
Flexion/Extension				
Extension – LR	0.78 ± 0.39	0.74 ± 0.26	$t(22) = 0.31$, p = .76	0.13
Extension – MS*	0.96 ± 0.45	0.62 ± 0.24	$t(22) = 2.28$, p = .032	0.93
Flexion – TS	0.63 ± 0.27	0.70 ± 0.41	$t(22) = -0.50$, p = .62	0.20
Extension – PSw	0.97 ± 0.34	0.99 ± 0.25	$t(22) = -0.15$, p = .89	0.06
Extension – ISw	0.73 ± 0.29	0.63 ± 0.20	$t(22) = 1.01$, p = .33	0.41
Flexion – TSw*	0.74 ± 0.34	1.05 ± 0.37	$t(22) = -2.16$, p = .042	0.88
Lateral Flexion				
Ipsilateral Stance	0.24 ± 0.11	0.29 ± 0.12	$t(22) = -0.92$, p = .37	0.37
Contralateral Stance	-0.37 ± 0.10	-0.43 ± 0.09	$t(22) = 1.59$, p = .13	0.65
Ipsilateral Swing*	0.22 ± 0.12	0.34 ± 0.11	$t(22) = -2.65$, p = .015	1.08
Contralateral Swing	-0.26 ± 0.12	-0.33 ± 0.15	$t(22) = 1.28$, p = .21	0.52
Axial Rotation				
Retraction Stance	0.17 ± 0.05	0.18 ± 0.07	$t(22) = -0.75$, p = .46	0.31
Protraction Stance	-0.14 ± 0.05	-0.10 ± 0.06	$t(22) = -2.02$, p = .055	0.83
Retraction Swing	0.14 ± 0.06	0.11 ± 0.08	$t(22) = 1.26$, p = .22	0.51
Protraction Swing	-0.17 ± 0.05	-0.19 ± 0.07	$t(22) = 0.76$, p = .46	0.31
Peak Powers (W·kg^{-1})				
Flexion/Extension				
S1 (Generation)	0.29 ± 0.22	0.18 ± 0.16	$t(22) = 1.32$, p = .20	0.54
S2 (Absorption)	-0.21 ± 0.09	-0.16 ± 0.11	$t(22) = -1.05$, p = .31	0.43
S3 (Generation)	0.32 ± 0.27	0.28 ± 0.19	$t(22) = 0.44$, p = .67	0.18
Lateral Flexion				
S4 (Absorption)	-0.33 ± 0.16	-0.25 ± 0.13	$t(22) = -1.35$, p = .19	0.55
S5 (Generation)	0.20 ± 0.10	0.30 ± 0.20	$t(22) = -1.57$, p = .13	0.64
S6 (Absorption)*	-0.14 ± 0.10	-0.05 ± 0.04	$t(22) = -2.82$, p = .010	1.15
Axial Rotation				
Generation	0.07 ± 0.04	0.05 ± 0.03	$t(22) = 1.16$, p = .26	0.47
Absorption	-0.08 ± 0.05	-0.07 ± 0.04	$t(22) = -0.89$, p = .38	0.36
Peak GRF (% BW)				
Mediolateral				
Medial	5.3 ± 1.5	5.9 ± 1.1	$t(22) = -1.13$, p = .27	0.46
Lateral	-2.3 ± 0.9	-2.8 ± 1.2	$t(22) = 1.09$, p = .29	0.45
Anteroposterior				
Braking	-21.2 ± 4.9	-18.2 ± 4.0	$t(22) = -1.59$, p = .13	0.65
Propulsive*	24.8 ± 5.1	21.1 ± 3.1	$t(18.0) = 2.14$, p = .046	0.88
Vertical				
Passive	114.9 ± 14.4	111.6 ± 6.7	$t(15.5) = 0.72$, p = .48	0.29
Active	119.2 ± 13.3	114.1 ± 5.3	$t(22) = 1.23$, p = .23	0.50

* significant age effect; LR = loading response, MS = midstance, TS = terminal stance, PSw = pre-swing, ISw = initial swing, TSw = terminal swing

The only between-group difference in peak spine powers was at S6 ($t(22) = -2.82$, $p = .010$), where the lateral flexor muscles in the YG (-0.14 ± 0.10 $W \cdot kg^{-1}$) absorbed significantly more power than in the OG (-0.05 ± 0.04 $W \cdot kg^{-1}$) during swing phase. The effect size for age-group difference at S6 was large (Cohen's d = 1.2). Whilst non-significant, mean differences in peak powers between age groups were generally moderate (**Table 7.9**). There was also a significant and large age effect for propulsive GRF ($t(18) = 2.14$, $p = .046$, Cohen's d = 0.9) where the YG (24.8 ± 5.1 %BW) produced an average 17.5% greater peak GRF along the anteroposterior axis than the OG (21.1 ± 3.1 %BW).

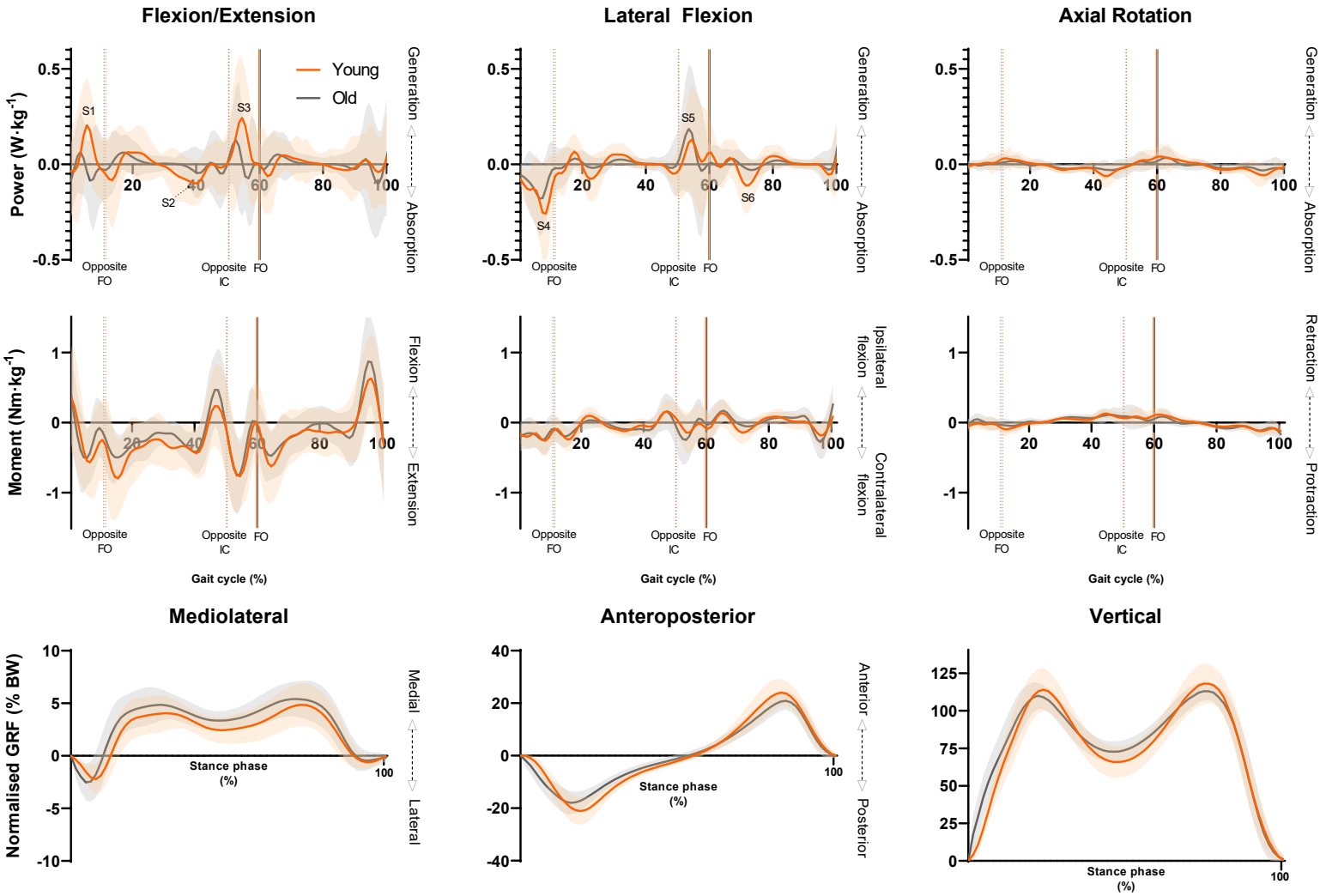

Figure 7.13 *Ensemble averages for trunk-P joint powers and moments and GRFs. Orange line = YG mean, grey line = OG mean, orange shaded area = YG SD, grey shaded area = OG SD*

The YG displayed significantly greater lower back extensor joint power than the OG during loading response (**Figure 7.13**). More precisely, the YG exhibited power generation in the extensor muscles whilst the OG exhibited power absorption during this period. This was shown by a supra-threshold cluster (4.9 – 5.4%) exceeding the critical threshold of $t(22) = 3.966$ (**Figure 7.14a**). The probability that a supra-threshold cluster of this size would be observed in repeated random samplings was $p = .045$. Lower back joint power waveforms in the coronal and transverse planes were similar between groups (**Figure 7.13**), resulting in a lack of significance (**Figure 7.14b and c**).

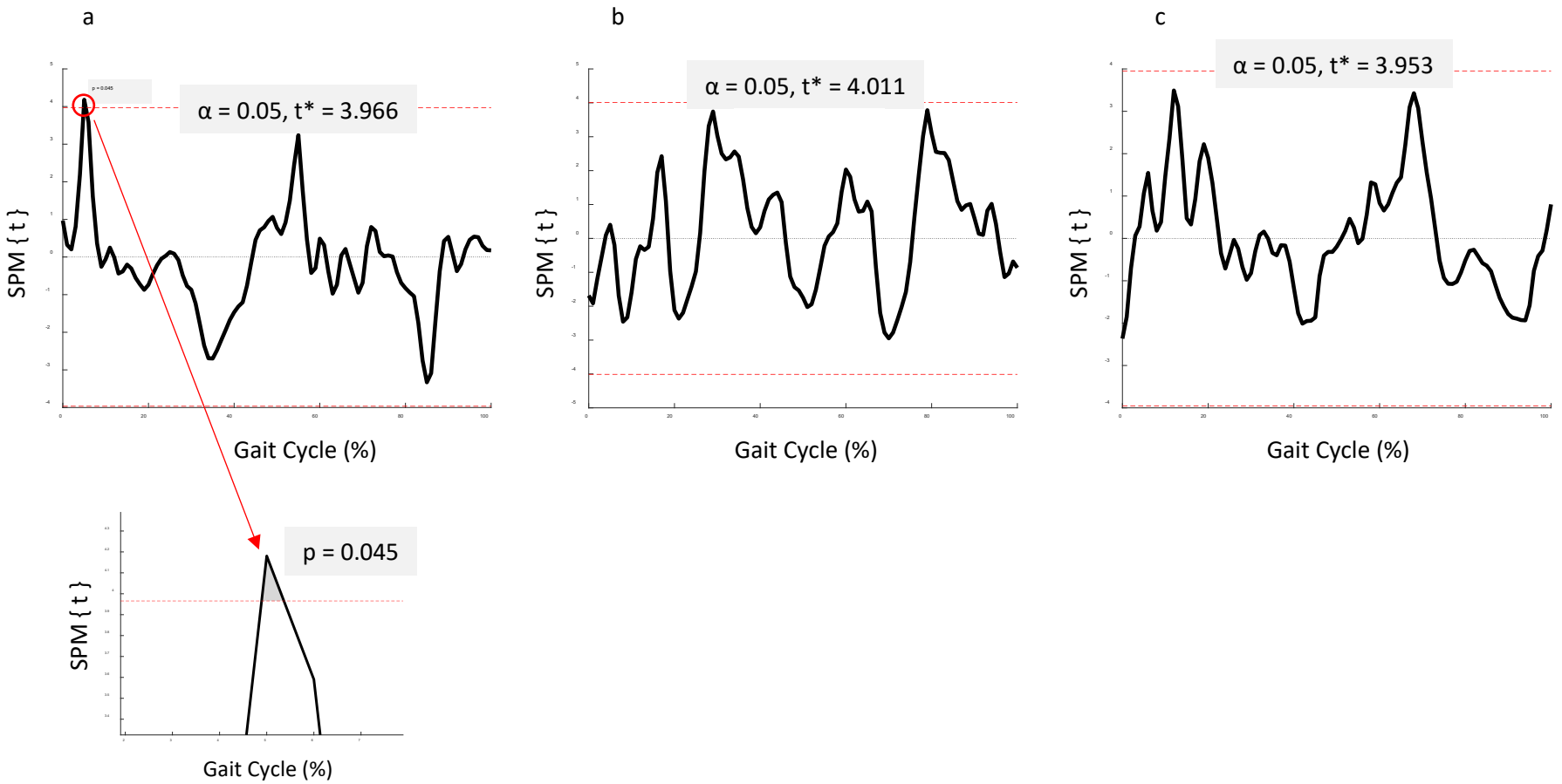

Figure 7.14 *Statistical Parametric Mapping (SPM) output for lower back joint power. a = flexion/extension joint power, b = lateral flexion joint power, c = rotational joint power*

Mean lower back joint moments were highly similar between the older and YG in all planes throughout the GC. No significant SPM phase differences were found (**Figure 7.15**).

Whilst the YG performed more positive and negative work than the OG in all three cardinal planes ($F(6, 17) = 1.53$, $p = .23$; Wilk's $\Delta = 0.65$, $\eta_p^2 = 0.35$), the only significant difference was in total negative work in the coronal plane ($F(1,22) = 5.95$, $p = .023$, $\eta_p^2 = 0.21$, $1-\beta = 0.65$). The YG (-0.051 ± 0.019 J·kg^{-1}) performed on average 49.8% more eccentric work in lateral flexion throughout the GC than the OG (-0.034 ± 0.013 J·kg^{-1}) (**Figure 7.16**).

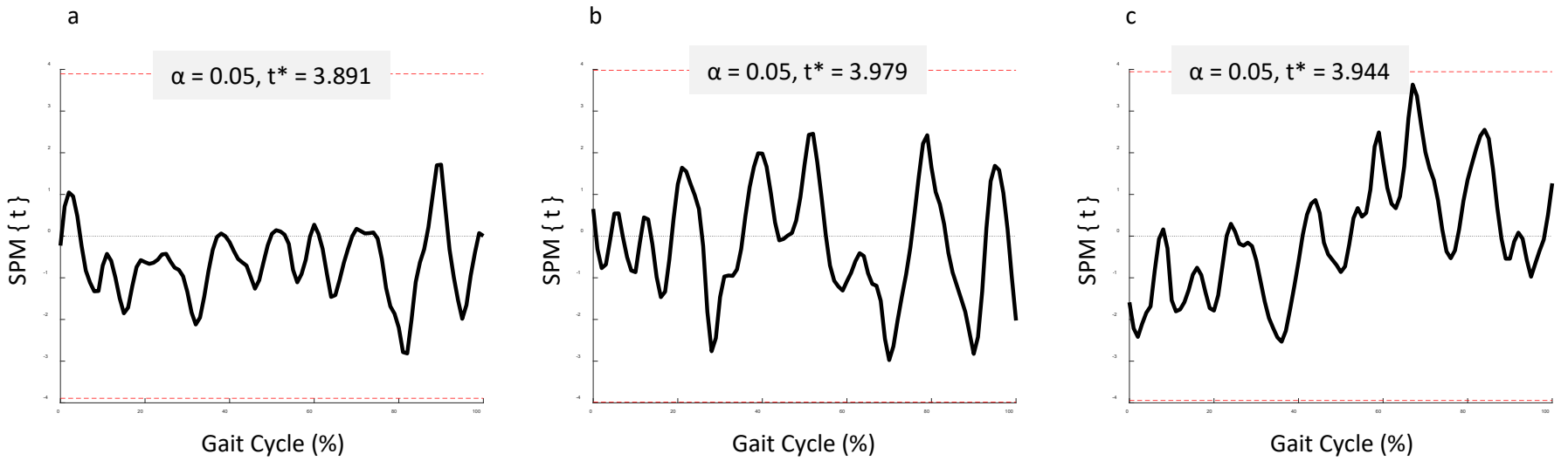

Figure 7.15 *Statistical Parametric Mapping (SPM) output for lower back joint moments. a = flexion/extension joint moments, b = lateral flexion joint moments, c = rotational joint moments*

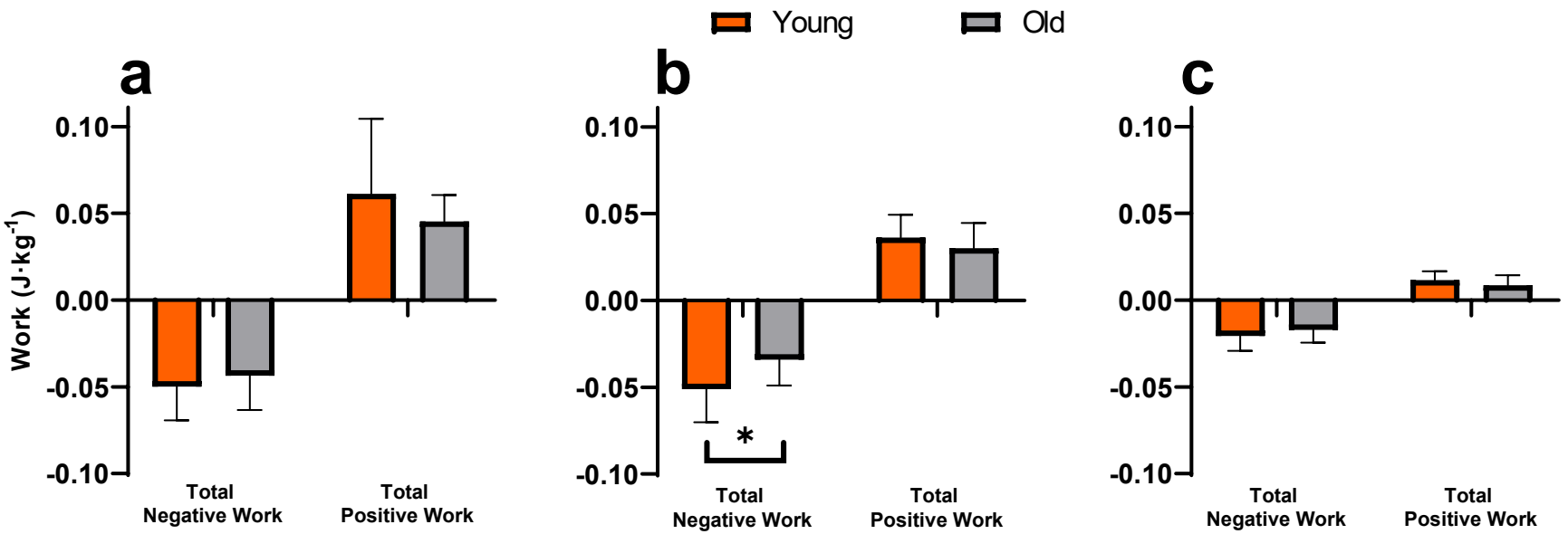

Figure 7.16 *Total positive and negative work performed during the gait cycle. a = work performed in the sagittal plane, b = work performed in the coronal plane, c = work performed in the transverse plane*

7.3.3.1 Functional Demand

The difference in FD between age groups was approaching statistical significance ($F(6, 17) = 2.40$, $p = .073$; Wilk's $\Delta = 0.54$, $\eta_p^2 = 0.46$), however, there were no significant differences in lower back FD between the YG and OG for any individual phase (**Figure 7.17**). FD was generally higher in the OG, except during MS where the YG's mean peak extensor moment was closer to their MVC. The difference in FD at TSw was greatest between groups and was approaching statistical significance ($t(22) = -1.97$, $p = .062$). The FD during TSw was on average 42% greater in the OG (34.8 ± 13.6%) compared to the YG (24.5 ± 12.1%). Across the phases shown in **Figure 7.17**, mean FD in the lower back was 20% greater in the OG.

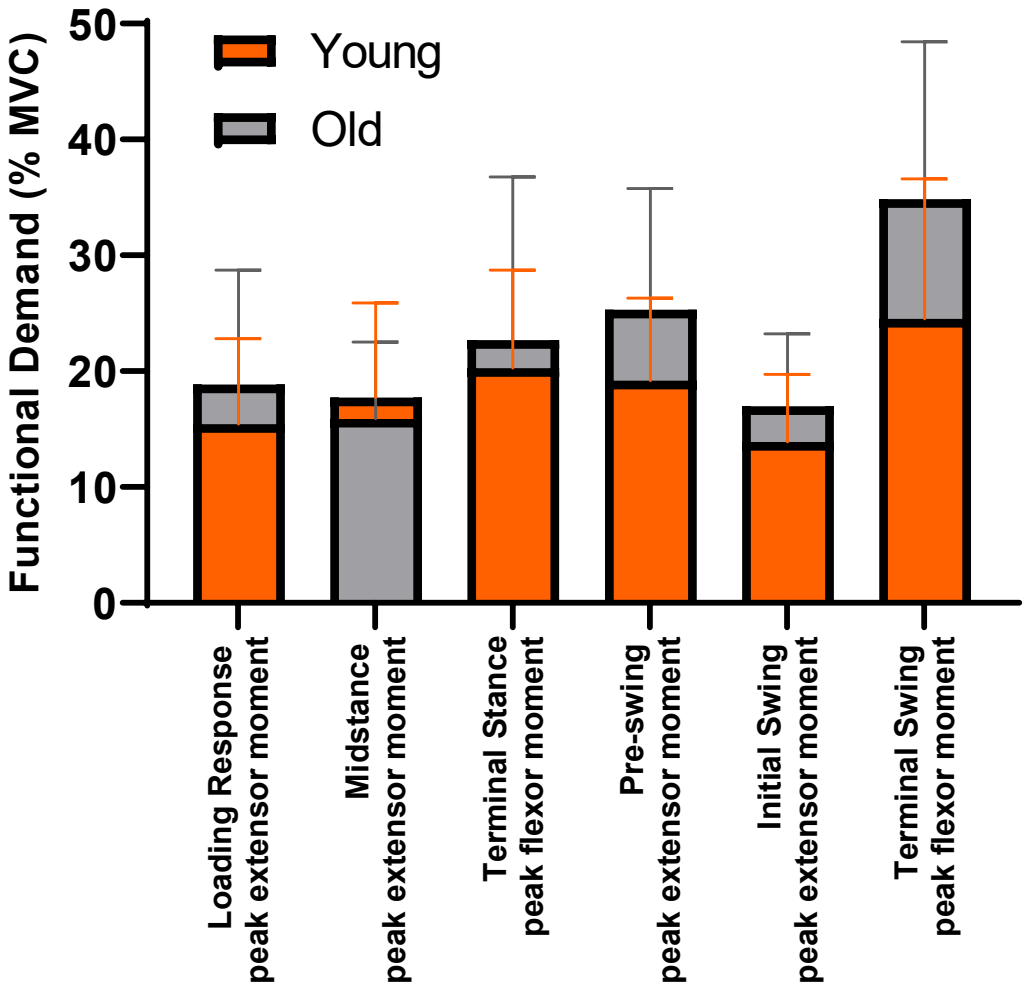

Figure 7.17 *Functional demand of the lower back joint in the sagittal plane, expressed as joint moment relative to maximum voluntary contraction (MVC) joint moment, at key instances during the gait cycle*

7.3.3.2 Moderating Effect of Muscle Morphology and Physical Activity

Mean MFI and total NMV of the LPMs were not significant covariates for any of the discrete spatiotemporal, kinematic or kinetic outcome variables. However, VPA was a significant covariate for total negative work performed in the transverse plane ($p = .034$), pelvic ROM in the transverse plane ($p = .047$) and peak pelvic retraction ($p = .020$). After controlling for these covariates, pelvic ROM ($F(1,19) = 4.38$, $p = .050$, $\eta_p^2 = .187$, $1-\beta = .510$) and peak retraction ($F(1,19) = 5.22$, $p = .034$, $\eta_p^2 = .215$, $1-\beta = .582$) in the transverse plane became significant. The difference in total negative axial rotation work performed remained non-significant ($p > .05$).

7.4 Discussion

Age related changes in 3-D trunk kinematics and kinetics are not fully understood. As the trunk plays crucial roles in mobility and postural support during ADLs (Panjabi, 1992; Hicks et al., 2005b), it is important to understand normal age-related changes in trunk movements as well as internal kinetics that drive these movements. It is also unknown how functionally demanding gait is on the lumbar spine for younger and older adults. Therefore, the current study sought to investigate age-related differences in biomechanical function of the trunk during normal gait. The results of this study extend our knowledge of ageing biomechanics in walking gait, highlighting kinematic and kinetic changes in the trunk in healthy older men. The main finding was that trunk-P kinetics are similar between healthy younger and older men during normal walking gait, whilst age-related differences in trunk-P kinematics are most apparent in the coronal and transverse planes. Furthermore, the initial periods of single limb support were identified as important phases during the GC capable of differentiating between young and older adult trunk kinematics. This suggests that age-related differences in the trunk are most apparent when dynamic balance is more compromised during initial single limb support phases. For the first time, the age-response was also elucidated for FD in the lower back. These observations stress the importance of upper body dynamics in gait and have potentially detrimental implications on walking efficiency and falls risk in older adults.

7.4.1 Age-related Differences in the Sagittal Plane

Sagittal plane kinematic waveforms were similar between the OG and YG across the GC for the trunk and pelvis segments, resulting in similar waveform patterns for the trunk relative to the pelvis. However, the OG adopted a more anteriorly tilted trunk position throughout the GC (**Figure 7.9**) and exhibited less ROM compared to the YG. Whilst waveforms were similar for the trunk and pelvis, there was a delayed phase-shift in the YG resulting in altered phase-specific flexion/extension movements of the trunk relative to the pelvis. However, these differences were small and non-significant.

Previous studies have reported comparable trunk flexion/extension kinematics despite a variety of modelling techniques (Thurston, 1985; Stokes, Andersson and Forssberg, 1989; Krebs et al., 1992; Taylor, Goldie and Evans, 1999; Whittle and Levine, 1999; Cromwell et al., 2001; McGibbon and Krebs, 2001; Vogt, Pfeifer and Banzer, 2002; Leteneur et al., 2009; Chung et al., 2010; Leardini et al., 2013; Hendershot and Wolf, 2014; Aminiaghdam et al., 2017; Crawford et al., 2018; Sartor et al., 1999; Crosbie, Vachalathiti and Smith, 1997). A study using the same marker model reported a similar ROM for the trunk relative to the global and pelvic reference frames (< 2°) yet a much greater mean trunk angle in an extended position (Chung et al., 2010). It is unlikely this difference was due to sampling

variance as participants were healthy and similar in age to the current study's YG. Marker placement errors are also unlikely to explain the difference as anterior pelvic tilt was also similar (~10°). Disparities were most likely caused by postural differences. Participants may have adopted a backward leaning position (Chung et al., 2010) compared to the participant's in the current study, which has been shown to result in more extended spinal angles during gait (Leteneur et al., 2009). More importantly, trunk flexion and extension were opposite in phase to those reported in the current study and previous studies (Thorstensson et al., 1982; Sartor et al., 1999; Van Emmerik et al., 2005). The model used by Sartor et al. (1999) was also similar; modelling the trunk relative to the pelvic reference frame. As the pelvis was tilted anteriorly throughout the GC, this would predispose the trunk-P to be in an extended position (Sartor et al., 1999). In both the YG and OG, the trunk was extended relative to the pelvis throughout the GC (**Figure 7.9**). Given the consistency with previous studies and from an anatomical perspective, the current results suggest that the modelling approach was indicative of spinal motion and not artefacts of the procedure.

Trunk-G tilt exhibited a biphasic oscillation, corresponding to one flexion/extension cycle for each step. These findings are supported by other gait studies assessing trunk movements in the sagittal plane (Thorstensson et al., 1982; Sartor et al., 1999; Crosbie, Vachalathiti and Smith, 1997), and substantiated by EMG studies that have shown peaks of ES and MF electrical activity at early midstance and around FO (Lamoth et al., 2004; Callaghan, Patla and McGill, 1999). Indeed, ES muscle activity precedes corresponding kinematics indicating that the paravertebral muscles drive trunk movement by anticipating propulsive phases in walking (Ceccato et al., 2009). However, others have reported different kinematics (Chung et al., 2010; Hendershot and Wolf, 2014), which may be due to modelling assumptions. More importantly, the current results indicate that trunk-P kinematics are largely unaffected by ageing likely due to the similarity in pelvic anterior/posterior tilt between age groups. However, the OG adopted a more forward tilted trunk-G and had significantly less trunk-G ROM.

Less trunk-G ROM during normal walking may simply reflect the reduction in total ROM in the lumbar spine with ageing (Yukawa et al., 2019; Intolo et al., 2009; Sullivan, Dickinson and Troup, 1994); a plausible explanation given that the LPMs become stiffer with age (Vazirian et al., 2016) and ROM is negatively correlated with muscle stiffness (Miyamoto et al., 2018). Reduced trunk-G ROM may also be indicative of a more conservative gait strategy to prevent larger destabilising forces in the sagittal plane (Van Emmerik et al., 2005), despite this being less efficient as more muscular work would be required for horizontal displacement of the COM. According to Chung et al. (2010), trunk movement in the sagittal plane counterbalances the cyclic motion of the lower limbs during swing phase. It was expected that the OG may extend the trunk to drive the lower limb into swing due to reduced ankle

power generation. However, upon approaching FO trunk extension relative to the pelvic and global reference frames occurred earlier in the GC than for the YG and at a lower amplitude. This suggests that the OG prepared earlier for single limb support which may be indicative of a more cautious gait strategy. To account for a conservative gait and significantly lower propulsive GRF, the OG may have adopted a forward leaning trunk position. A similar strategy has been observed in lower-extremity amputees (Goujon-Pillet et al., 2008) assisting in forward progression to compensate for a lack of propulsive force (Hendershot and Wolf, 2014). According to Leroux, Fung and Barbeau (2002), tilting the trunk forward is the best strategy to generate greater forward propulsion during walking gait. Therefore, the OG may have adopted a greater anterior trunk-G angle to facilitate their forward progression. Tilting the trunk anteriorly may also play an injury prevention role; reducing lower limb stress by damping COM oscillations (Krebs et al., 1992).

7.4.1.1 Modified Lower Back Moments in Older Age

Additional demands may be placed on the trunk musculature in older adults as a consequence of flexed postures (Hendershot and Wolf, 2014). Greater internal extension moments generated by the paravertebral muscles are required to balance the external flexion moment about the lower back caused by an anterior shift in the COM of the upper body (Le Huec et al., 2018). However, this was not seen in the current results (**Figure 7.13**). Generally, the OG produced lower extension moments throughout the GC and performed less negative and positive work in the sagittal plane. Flexion/extension moment waveform patterns were similar between groups. Despite the OG having a lower strength reserve (see **Chapter 6**), differences in FD were also non-significant between the age groups. However, the OG generally operated nearer their maximal capacity to generate and absorb flexion/extension moments during gait, particularly near the end of the GC (**Figure 7.17**). This was highlighted by the significantly greater flexion moment absorbed by the OG during TSw compared to the YG. Whilst this eccentric activity in the trunk flexors may be a stability mechanism to reduce posterior translation of the body's COM over its BOS in preparation for impact at IC, it is likely that the resulting motion of extension aids in forward progression of the body's COM (Sartor et al., 1999). Indeed, trunk extension may also act as a stabilising mechanism for hip extensor activity, particularly during LR when the hip extensors control a large external flexor moment (Sartor et al., 1999).

Trunk extension was demonstrated by the YG during LR whilst the OG exhibited altered neuromuscular control of the trunk during this phase (**Figure 7.9**). The OG fluctuated between extensor power generation and absorption whilst the YG predominantly generated extensor power during LR. Peak power generation was also smaller in the OG. This age-related disparity was replicated in the second

period of double limb support during PSw. Other studies support this finding that locomotor function of the trunk is altered in older age during gait (McGibbon and Krebs, 2001). It is possible that younger adults contract the lower back muscles concentrically during double limb support to adopt a pelvic-leading gait strategy which is more efficient. In older adults, the trunk-leading strategy results in greater mechanical energy expenditure of the lower back musculature due to greater eccentric activation (McGibbon and Krebs, 2001). Consistent with the current findings, McGibbon and Krebs (2001) found that older adults have a greater reliance on eccentric control of the LPMs to mediate energy transfer during double limb support phases. Furthermore, lower power generation in the trunk may be indicative of the OG reducing concentric muscle activity to minimise energy transferred proximally to the trunk during periods of less stability such as single limb support (McGibbon and Krebs, 2001). The current results exemplify this as the YG generated significantly greater peak extensor moments during MS compared to the OG. This mechanism is supported by EMG studies showing lower muscle activity in the ES and psoas muscle groups of older adults compared to younger adults during walking (Ceccato et al., 2009; Schloemer et al., 2017).

The moment waveforms for the OG and YG are similar to those produced in previous studies (Hendershot and Wolf, 2014; Leteneur et al., 2009), despite differences in biomechanical models. The most notable features are the extensor moment peaks produced during MS (YG = 0.96 ± 0.45 $Nm \cdot kg^{-1}$ and OG = 0.62 ± 0.24 $Nm \cdot kg^{-1}$) and PSw (YG = 0.97 ± 0.34 $Nm \cdot kg^{-1}$ and OG = 0.99 ± 0.25 $Nm \cdot kg^{-1}$), which were similar to previously reported values of approximately 0.5 – 1.1 $Nm \cdot kg^{-1}$ in healthy adults (Leteneur et al., 2009; Hendershot and Wolf, 2014). Fernandes et al. (2016) reported lower extension moments of 0.23 $Nm \cdot kg^{-1}$, although they also reported substantially lower NJMs in the ankle, knee and hip than typically reported (Sadeghi, Allard and Duhaime, 2000; Chen, Kuo and Andriacchi, 1997; DeVita and Hortobagyi, 2000; Cofré et al., 2011). Therefore, peak extension moments in the lower spine were also likely underestimated. Peak extension moments in the trunk appear to be generally less than 1.0 $Nm \cdot kg^{-1}$ in healthy younger and older men and are therefore lower than peak moments typically seen in the lower limb joints during normal gait. However, Leteneur et al. (2009) observed that peak moments in the lumbar spine (~ 1.1 $Nm \cdot kg^{-1}$) were slightly larger than those developed in the hip (~ 0.8 $Nm \cdot kg^{-1}$) and knee during gait. Given the range of methods, it is difficult to conclude whether the trunk plays a more important role than the lower limbs during walking. Based on typical lower body joint moment patterns and the current results, it is likely that the dominant role changes between the trunk and lower limb joints throughout the GC. However, older age may not influence the contribution of lower back flexion/extension moments during the GC as phase-specific differences were not observed between age groups in this study. Researchers have suggested that low levels of loading, spinal motion and muscular activation during walking gait present a low risk of injury

(Leteneur et al., 2009; Callaghan, Patla and McGill, 1999), which also makes walking a suitable exercise for general back exercise and rehabilitation programmes in older adult populations (Callaghan, Patla and McGill, 1999).

7.4.2 Age-related Differences in the Coronal Plane are a reflection of Pelvic Movement

Trunk motion in the coronal plane was less variable than in the sagittal plane, consistent with the literature (**Table 7.2**). Movement patterns were similar between age groups. The OG and YG both demonstrated approximately 1° of trunk-P lateral flexion towards the ipsilateral limb at IC. The trunk relative to the pelvis continued to flex laterally over the reference stance limb until it reached a peak during early midstance. The trunk-P then flexed laterally towards a neutral position and maintained a neutral position throughout late midstance. During PSw the trunk-P flexed away from the reference limb in the coronal plane and reached its contralateral peak during early swing. Contra- and ipsilateral flexion peaks both occurred as single-limb support phases commenced, in agreement with previous findings (Chung et al., 2010). Trunk-P lateral flexion angle then returned to neutral from midswing until the subsequent IC.

Results from the SPM indicate that there is a phase-specific age effect in trunk-P coronal plane kinematics during early midstance (13.3 – 17.7%) and early swing phase (62.5 – 67.2%). This is supported by movement amplitudes being significantly greater in the YG than the OG. Given that trunk movements relative to the global reference frame were similar in phase and amplitude between groups, and pelvic obliquity exhibited similar significant differences to trunk-P lateral flexion; it is likely that trunk-P motion in the coronal plane was simply a reflection of pelvic movement. This is supported by Crosbie, Vachalathiti and Smith (1997) who state that spinal movements associated with walking are linked to the primary motions of the pelvis. Furthermore, Krebs et al. (1992) found that greater trunk ROM in the pelvic reference frame was due to independent pelvis motions moving out of phase with the trunk. Indeed, several researchers have indicated that lumbar spinal motion is affected by pelvic motion (Callaghan, Patla and McGill, 1999; Feipel et al., 2001; Stokes, Andersson and Forssberg, 1989; Crosbie, Vachalathiti and Smith, 1997; Vogt and Banzer, 1999).

Decreased trunk-P motion may partially explain why falls are more prevalent in older adults (Sharif et al., 2018). According to Balaban and Tok (2014), increased lateral trunk flexion and elevation of the hip elicits improved foot clearance in stroke patients. Lower lateral flexion peaks in the OG may lead to reduced foot clearance (Prince et al., 1997) and consequently increase risk of falling (Robinovitch et al., 2013). However, older adults may prevent impact injuries by decreasing trunk movement in the coronal plane. Reducing peak lateral flexion during the early stages of single-limb support allowed the

OG to decrease trunk-P angular velocity when it started to move towards the opposite side (**Figure 7.18**), which may be an effort to reduce impact from heel strike (Chung et al., 2010). Indeed, the trunk acts to reduce angular velocity toward the contralateral side (Chung et al., 2010). However, pelvic obliquity plays an important role in creating a more energy efficient gait pattern by reducing COM vertical oscillations (Saunders, Inman and Eberhart, 1953). The reduced pelvic ROM with age may therefore impact upon the OG's gait stability and efficiency during normal walking (Saunders, Inman and Eberhart, 1953; Van Emmerik et al., 2005).

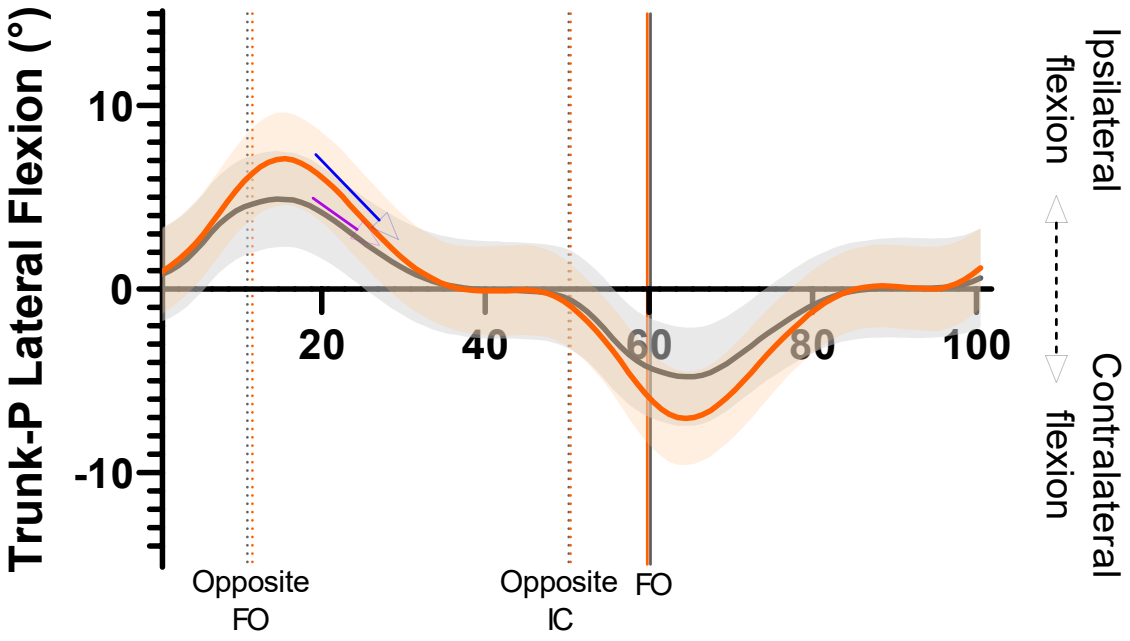

Figure 7.18 *Trunk-P motion in the coronal plane. The difference between the young group's slope (blue arrow) and old group's slope (purple arrow) represents the difference in angular velocity.*

For both groups, trunk ROM in the coronal plane was greater relative to the pelvis than in the global reference frame. These results support the assertions of Saunders, Inman and Eberhart (1953) that pelvic motion in the coronal plane is particularly important for reducing trunk-G oscillations, which could excessively displace COM and cause lateral instability during walking. It should be noted that trunk-P ROM was related to both pelvic and trunk-G ROM in the coronal plane after controlling for age, which may be an indication of the out-of-phase rotations between these segments. Instability during the GC may be minimised by permitting a relatively larger ROM in the trunk relative to the pelvis than in the global reference frame, through the independent motions of the pelvis. The reduced pelvic and thus trunk-P ROM in the OG may decrease stability during gait, although relatively small

trunk lateral flexion ROM is required to transfer weight and assist foot clearance in the swinging leg during gait (Krebs et al., 1992). This may explain why the OG, despite reduced trunk-P and pelvis movement, exhibited similar trunk-G ROM to the YG.

Moment waveforms were similar between the YG and OG, indicating no phase-specific differences throughout the GC (**Figure 7.15b**). However, significant differences were observed in peak values. Comparable peak lateral flexion moments have been previously reported (Fernandes et al., 2016; Hendershot and Wolf, 2014), increasing confidence in the current results. Peak ipsilateral flexion moment during swing phase was greater in the OG than the YG. Peak power absorption during swing phase however was significantly larger in the YG. These findings suggest that during swing younger adults rely more on angular velocity to conserve trunk momentum in the coronal plane whilst older adults rely more on muscular force to absorb trunk-P powers. This would increase the demands of walking in older adults, compounded by dysfunctional muscular activation patterns which further increases mechanical energy demands (McGibbon and Krebs, 2001). However, the YG performed significantly more negative work in the coronal plane than the older adults (**Figure 7.16b**). This is likely to be the result of greater trunk-P excursions, which contribute to larger lateral flexion moments (Hendershot and Wolf, 2014). As older adults are known to operate nearer their maximum physical capacity during ADLs compared to younger adults (Reeves et al., 2008), and a large amount of muscle activity is required to maintain balance in the coronal plane during walking (O'Connor and Kuo, 2009), the lower amount of negative work performed may mask the FD of walking in older adults. For example, it has been observed that decreased step lengths may result from declining trunk kinematics which cost an inordinate amount of energy to be expended to maintain erect posture and forward progression (Sartor et al., 1999). Given the differences in peak moments and powers and the phase at which they occurred between groups, it appears that older adults may solicit their trunk muscles to produce instances of high lateral flexion moment providing vertical support. Whereas younger adults appear to perform more work in the coronal plane to maintain forward progression (Sartor et al., 1999), similar to the contributions of the lower limb muscles during gait (Kepple, Siegel and Stanhope, 1997).

7.4.3 Age-related Kinematics in the Transverse Plane

Similar to kinematics in the coronal plane, pelvic and trunk movement patterns in the transverse plane were more consistent than in the sagittal plane. These results concur with the literature (**Table 7.2**). In the transverse plane, age-related differences in pelvic rotation peaks and ROM were approaching significance and became significant when VPA was controlled for. There was a significant age effect in

peak trunk-P retraction, with differences in peak trunk-P protraction and ROM also approaching significance between groups. Trunk-G axial rotation showed the greatest age-related differences; the YG demonstrated significantly greater trunk-G retraction and protraction peaks as well as a greater ROM. Whilst it appears that trunk-G axial rotation was most affected in older age based on peak values and ROM, no phase-specific differences were observed in the trunk-G. Furthermore, coinciding phase-specific differences between groups were identified in the pelvis (68 – 76% of GC) and trunk-P (66 – 77% of GC) during swing phase. Coupling between the trunk-P and pelvis was further supported by the significant correlation in their ROMs in the transverse plane. These results indicate that age-related differences in the transverse plane involve a complex interrelationship between the trunk and pelvis. It appears that instances of peak rotational excursion during gait, immediately following heel strikes, are affected by age-related kinematic changes in the trunk-G. Whereas during single limb support, age-related differences in the pelvis appear to be more influential to trunk-P motion. However, significant age-related decrements in peak trunk-G motions may be misleading and possibly mask the relative contribution of the pelvis to the coordination of the upper body during walking gait.

Concurrent with other research (Krebs et al., 1992; Thurston, 1985; Chung et al., 2010), trunk-P curve reversals in the transverse plane were observed immediately after each heel strike. This was likely caused by the contravening rotation of the pelvis continuing after the rotation of the trunk-G had stopped (Krebs et al., 1992). This highlights that discrete kinematic values may be insufficient to understand how ageing affects the complex relationship between pelvis and trunk axial rotations during gait. The movement pattern of the trunk-P in the transverse plane is governed by the out-of-phase rotations of the trunk-G and pelvis. Simply put, the trunk-G and pelvis rotate in opposite directions about the vertical axis relative to each other during the GC. This movement pattern has been consistently shown (Whittle and Levine, 1999; Chung et al., 2010; Leardini et al., 2013; Bruijn et al., 2008; Titus et al., 2018). At slow gait speeds the trunk and pelvis may demonstrate synchronous axial rotations (Lamoth et al., 2002), although this is equivocal (Chung et al., 2010). As walking speed in the OG was not significantly slower than in the YG, it is unsurprising that kinematic waveforms in the transverse plane were similar. However, this should be considered with caution. It is highly contested whether changes in walking speed affect trunk movements during gait (Van Emmerik et al., 2005; Stephan, Sutin and Terracciano, 2015; Kavanagh, 2009; Taylor, Goldie and Evans, 1999). Bruijn et al. (2008) and Van Emmerik et al. (2005) suggested that rotational motion of the trunk is important to adapting to changes in walking speed. However, others have suggested that trunk motion is more dependent on loss of strength and flexibility in older age than slower walking speeds (Kang and Dingwell, 2008). When interpreting these findings, it should be noted that the difference between age groups in walking speed was moderate to large despite being non-significant.

The out-of-phase rotation between the trunk-G and pelvis is essential in reducing the momentum of the trunk-P to conserve angular momentum (Chung et al., 2010; Crosbie, Vachalathiti and Smith, 1997) and maintain stability during gait (Van Emmerik et al., 2005). There are a few reasons that may explain why ROM was decreased in the OG, such as increased rigidity in the trunk (Van Emmerik et al., 1999). This is questionable however as others have indicated that trunk-P kinematics in the transverse plane are not influenced by trunk stiffness (Prins et al., 2019). Another mechanism concerns decreasing ROM to maintain stability. Larger axial rotations in the trunk are indicative of instability in immature and pathological gait (Ledebt and Bril, 2000; Winter, 1995). The OG may have attempted to maintain stability by reducing trunk ROM, in agreement with previous findings (Van Emmerik et al., 2005). Finally, this may have been a strategy to increase energy in the trunk to compensate for reduced propulsive force (McGibbon, Krebs and Puniello, 2001). However, the current findings do not support this as trunk-P powers in the transverse plane were relatively small and there were no significant differences with age. Furthermore, there were no significant phase-specific or peak differences in rotational moments between the OG and YG, indicating that age-related differences in trunk kinematics were not caused by changes in kinetics. This suggestion is substantiated by the relatively small amount of rotational work performed in the transverse plane and non-significant difference between age groups.

7.4.4 Age Effect on Interplanar Motions

Older age appears to alter the coupling between motions in different planes. Zero-order correlations showed that there were numerous significant relationships between the ROM of the trunk-P, trunk-G and pelvis in different planes. For example, the range of flexion/extension in the trunk-P was moderately correlated with lateral tilt ROM. After controlling for age, only three significant correlations remained, all exclusive to their respective planes. These were between trunk-P ROM and trunk-G ROM, and between trunk-P ROM and pelvic ROM; both relationships in the coronal plane. The other significant partial correlation was between trunk-P and pelvic axial rotation ROMs. These results suggest that in older age, rotations of the upper body in a given plane become independent of rotations in orthogonal planes. Whilst this finding has never been reported before, others have shown that coronal and transverse plane trunk motions are inter-connected in younger adults (Chung et al., 2010; Whittle and Levine, 1999) and that older adults exhibit reduced compensatory coordination between trunk and pelvis movements during walking (Van Emmerik et al., 2005).

In older age, disassociation of trunk and pelvic movements in the Cardinal planes may increase the energetic demands of walking. To highlight this concept, more obvious examples can be drawn upon.

For example, upper body movements such as punching require axial trunk rotation to transfer kinetic energy from the lower to upper extremities. The interplanar relationship between the lower and upper extremities as well as the trunk is designed to minimise energy cost and increase horizontal punching force (Tong-Iam, Rachanavy and Lawsirirat, 2017). Therefore, walking is likely to be more energetically demanding in older adults if angular momentum cannot be conserved from motions in orthogonal planes to assist in forward progression of the body's COM.

It has been suggested that the vector of the spinal muscles may be responsible for the association between coronal and transverse plane trunk movement (Chung et al., 2010). MRI studies using diffusion-tensor techniques have shown that ageing affects the orientation of muscle fibres (Yoon et al., 2018; Sinha et al., 2015; Farrow et al., 2020). Therefore, it is likely that degeneration of paraspinal muscle structure and function is in part responsible for age-related uncoupling of trunk and pelvic kinematics in the Cardinal planes. Emerging evidence from ongoing research as part of this PhD project has shown that fractional anisotropy of the LPMs is greater in the OG than the YG. Whilst the exact mechanisms for the age-related disassociation of interplanar motions in the trunk are unknown, this ongoing research may provide a useful first step.

7.4.5 Clinical and Practical Applications

Understanding age-related differences in trunk biomechanics could assist clinical decision making and public health strategies with regards to older adults incorporating motor skills training into exercise interventions and physical activity programmes. Exercise interventions generally focus on promoting strength, endurance, balance and flexibility. Whilst useful, some evidence indicates that their impact on physical function and mobility is modest in older adults; increasing self-selected walking speed by up to 13% (Wolf et al., 2006; Pahor et al., 2006; Buchner et al., 1997; Bean et al., 2004). Incorporating such training into exercise interventions may potentially improve physical function and mobility to a greater extent. The current findings could provide a targeted approach by highlighting the movement patterns in the trunk which should be prioritised in exercise interventions to reduce gait inefficiencies in older age. This study found that trunk movements were most affected in older age in the coronal and transverse planes. Therefore, concentrating efforts on these planes of motion during motor skills training may reduce the impact of ageing on trunk movements during walking gait. This would consequently reduce the mechanical demand on the trunk musculature and possibly reduce the FD throughout the GC.

In clinical gait analysis the upper body is typically overlooked whilst the lower limbs receive much of the focus. This study highlights the need to investigate trunk biomechanics; indicating that during low-

impact everyday activities such as walking the trunk muscles are challenged more in older age. Routinely analysing the trunk will increase the evidence base and establish normative data in younger and older adults, which could be used to identify abnormal movement patterns higher up the kinetic chain that may not be apparent in the lower limbs. Furthermore, the PiG model would be appropriate in clinical settings where a balance of accuracy and practicality is needed. Therefore, clinicians and biomechanists analysing the trunk should look to use the full-body PIG model, which would allow sufficiently detailed analysis of the trunk in 3-D whilst having a negligible impact on data collection time.

7.4.6 Limitations

There were limitations in this study that should be acknowledged. Namely, it was inferred that the floating axis about which the pelvis and trunk segments rotated approximated the lumbar spine. PiG has been an effective and widely used modelling approach in biomechanics to analyse human gait (Ramanujam, Forrest and Sisto, 2008; Cockcroft, Louw and Baker, 2016), however, other modelling approaches that use the same marker set may be superior when analysing 3-D kinematics and kinetics of the lumbar spine (Stambolian, Asfour and Eltoukhy, 2014). Given that the current results are highly similar to kinematics and kinetics of the lumbar spine reported in previous studies, more complex models may not provide a positive additional information to computational time balance. This is more pertinent for musculoskeletal models, where the benefit of physiological accuracy may not outweigh the computational time to solve the inverse dynamics and optimisation problems (Shourijeh, Mehrabi and McPhee, 2017). Furthermore, musculoskeletal models are still susceptible to modelling assumptions and may therefore produce results that no more translatable to real-world situations than the PiG model. Indeed, validation of lumbar spine musculoskeletal models has typically included only one participant (Bassani et al., 2017; Raabe and Chaudhari, 2016). Comparison of PiG against the Full-Body Lumbar Spine Model (OpenSim) (Raabe and Chaudhari, 2016) and Lumbar Spine Model (AnyBody Technology) (de Zee et al., 2007) is a necessary first step in establishing concurrent validity between these approaches. If kinematics is the primary outcome, musculoskeletal modelling would be unnecessary. Future research should look to include more markers on the lumbar spine to better represent the movement of the individual vertebral bodies. However, researchers should be aware that this approach may lead to marker placement errors and additional inter-marker soft tissue artefacts. Confidence in the current results is high since the movement patterns and peak values in all planes were highly comparable to a study using indwelling bone pins to assess 3-D motion of the lumbar spine during gait (MacWilliams et al., 2013), which is considered the gold-standard approach.

Another limitation concerns the ecological validity of the testing procedure. The laboratory environment, barefoot walking and attachment of markers to participants' bodies, may have influenced habitual gait patterns (Franklin et al., 2015). Also, the sample comprised of healthy active young and older adult men. It has been shown that trunk kinematics are affected by disease and physical impairment such as stroke (Titus et al., 2018). Therefore, caution should be taken when generalising the findings of the current study to populations other than healthy men. This study was also specific to walking gait biomechanics. Age-related differences may be more pronounced for more challenging movements such as stair negotiation and sit-to-stand.

Finally, it was not possible to calculate FD for lateral flexion and axial rotation of the lumbar spine in the current study. Therefore, it is unknown whether walking was more functionally demanding for the OG in the coronal and transverse planes compared to the YG. Given that the only significant age-related difference in work performed was in the coronal plane, obtaining isokinetic strength data for lateral flexion of the trunk may reveal useful findings about the FD of walking in the coronal plane. It should also be noted that FD is a relative measure of an individual's maximal capacity to produce muscular force. Whilst it is an intuitive measure and provides useful information, it does not account for the fatigue resistant capabilities of the muscle group being assessed. Understanding the FD of the trunk during gait across longer distances or time periods may reveal other age-related mechanisms associated with neuromuscular fatigue. Since older adults are arguably more susceptible to the detrimental effects of muscle fatigue than younger adults (Kent-Braun, 2009; Sundberg et al., 2018; Allman and Rice, 2002), it is likely that the trunk musculature would not respond as well to the demand for prolonged force production. This may have greater consequences in older populations, limiting their ability to perform longer duration walking.

7.5 Conclusion

Older age appears to alter trunk kinematics primarily in the coronal and transverse planes, indicative of a more conservative gait strategy. Kinematic changes with age were characterised by reductions in peak amplitudes and ROM in all planes of motion. Few age-related differences existed in trunk kinetics, although midstance and swing were identified as phases where older age alters flexion/extension and lateral flexion trunk moments. The mechanical demands of gait also seem to be largely unaffected in older age based on the total amount of work performed, although FD was generally greater in the OG compared to the YG. In the current sample of healthy men, older age appears to only significantly reduce the amount of negative work performed in the coronal plane during walking gait. Furthermore, muscle degeneration in the lumbar spine did not covary with

kinematic or kinetic outcomes. VPA also had little effect on age-related differences in trunk biomechanics, although it moderated the effect of age on pelvic motion in the transverse plane. Future research should look to calculate FD of the trunk during gait in the coronal plane. Further research in a range of healthy age groups and diseased populations should be undertaken to increase understanding of normal biomechanical trunk function during gait with ageing.

Table 7.10 Thesis Map

Chapter and Study	Problem Statements		Outcomes
Chapter 3 Assessment of Variables that may covary with Age-related Differences in Muscle Morphology, Strength and Function	• Physical activity level, body composition, handgrip strength and functional disability varies greatly with age and the values of each domain are highly individualised • These variables are known to influence measures of muscle mass, strength and function	Aim	• To establish whether there were significant differences in physical activity level, whole body composition, handgrip strength and functional disability between the older and younger groups
		Key findings	• The younger group were significantly more active regarding vigorous physical activity than the older group • Dominant and non-dominant handgrip strength was significantly greater in the younger group compared to the older group • Appendicular lean mass was significantly greater in the younger group, whilst whole-body fat mass was greater in the older group
		Implications	• Vigorous physical activity level should be included as a potential covariate in statistical models comparing muscle morphology, spinal muscle strength and physical function between the age groups • The moderating effect of body composition measures and handgrip strength should be explored in statistical models assessing the effect of older age on trunk muscle strength
Chapter 4 Age-related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-level Meta-regression	• A quantitative analysis on the association between healthy ageing and morphological degeneration of the lumbar paravertebral muscles has not been performed to date • It is unknown how the muscles in the lumbar spine change in size and composition with healthy ageing in older adults. Understanding this phenomenon may elucidate mechanisms related to functional decline. • Studies use a wide range of methods to evaluate the lumbar musculature. A statistical model is needed to include each variable as a potential moderator to account for heterogeneity amongst studies • Multiple effects are typically reported by a single study. Meta-analyses typically adopt a reductionist approach by aggregating effect sizes. To adopt an integrative approach, a novel statistical model is needed to account for interdependency amongst effect sizes	Aims	• To perform a quantitative analysis of the literature to establish the relationship between normal ageing and lumbar paravertebral muscle degeneration • A secondary aim was to identify important methodological parameters that moderate the relationship between ageing and degeneration of paravertebral muscle morphology
		Key findings	• The lumbar paravertebral muscles experience significant atrophy and fat infiltration with ageing • Degeneration is muscle-, level- and sex-specific • Fat infiltration appears to be more effectual than atrophy with ageing in the lumbar musculature • Imaging modality significantly influences the relationship between ageing and paravertebral muscle atrophy • There is a considerable amount of between-study heterogeneity, although methodological factors explain a substantial amount of explainable variance
		Implications	• Use high-resolution imaging modalities (e.g. MRI/CT) to image to spinal musculature • Volumetric measures covering multiple lumbar levels are superior to cross-sectional measures taken at single levels • Measurements should be obtained for each of the main paravertebral muscles in the lumbar to better represent the degenerative effects of ageing
Chapter 5 Age-related Differences in Lumbar Paravertebral Muscle Morphology in Healthy Younger versus Older Men	• Studies investigating muscle degeneration with ageing have typically focused on the appendicular muscles • There is increasing recognition for the importance of the lumbar paravertebral muscles in maintaining health and mobility in older age	Aims	• To investigate age-related differences in LPM morphology • A secondary aim was to investigate the age-response on fat infiltration and volume of the different lumbar muscles (i.e. MF, ES, QL and PS) • An additional aim was to explore other predictors of lumbar paravertebral muscle degeneration

	• Few studies have characterised features of age-related degeneration in the lumbar musculature • Few studies have provided volumetric information on all of the paravertebral muscles using high-resolution imaging modalities	Key findings	• Older age negatively affected all paravertebral muscles, although some showed greater degenerative changes than others • Age-related fat infiltration has a global effect across the lumbar musculature, whereas atrophic changes appear to be muscle-specific • Only the QL and ES showed significant age-related declines in muscle volume • All muscles showed age-related declines in muscle quality (i.e. increase in intramuscular adipose tissue) • The MF was most susceptible to compositional changes with age, whilst the QL was most vulnerable to reductions in muscle volume • Physical activity did not influence age-related differences in muscle degeneration in the lumbar spine • Non-dominant handgrip strength was a predictor of muscle atrophy in the lumbar musculature
		Implications	• The QL and ES appear to be most affected in older age since they exhibited declines in size and quality • When investigating the effects of ageing on lumbar muscle function, macroscopic changes in the paravertebral muscles should be considered • Structural changes, resulting in a loss of contractile tissue, may reduce muscle function in the lumbar spine • Convenient and easily administered measures such as handgrip strength may be able to predict muscle atrophy in the lumbar spine
Chapter 6 Age-related Differences in Concentric and Eccentric Isokinetic Trunk Strength in Healthy Older versus Younger Men	• Dynamic trunk strength in older adults has not been fully explored • Studies have typically investigated age-related strength loss using handgrip dynamometry or lower limb isokinetic dynamometry • Majority of studies have used clinical assessments which may not be appropriate to assess maximal trunk strength • No study has assessed eccentric trunk strength in older adults and contractile modes are typically limited • The findings from chapter 5 have also influenced the need for this study. Research investigating how muscle morphology degeneration in the lumbar spine impacts on trunk extensor strength is warranted	Aims	• To investigate age-related differences in dynamic trunk strength • The secondary aim was to explore the moderating effect of muscle morphology degeneration on extensor muscle strength
		Key findings	• Age had a significant and negative effect on peak concentric trunk extensor torque across all angular velocities • The difference in concentric extensor torque between the older and younger group increased with increasing angular velocity indicating that the lumbar extensor muscles express a slower phenotype with ageing • Peak concentric torque of the trunk flexor muscles decreases in older age but not significantly • Peak eccentric torque of the extensors and flexors in the trunk is preserved in older age • Concentric strength of the trunk extensor muscles is negatively associated with age, but not paravertebral muscle morphology • Eccentric strength of the trunk is primarily related to quadratus lumborum muscle quality, but not age
		Implications	• Loss of trunk strength in older age is contractile mode- and muscle- specific • Training interventions should target the extensor trunk muscles using concentric exercises to improve strength in older adults • Improving paravertebral muscle quality may further preserve eccentric strength of the trunk extensors

			• Internal trunk moments produced during daily tasks should be combined with the peak values measured in this study to determine how functionally demanding these tasks are on the trunk musculature
Chapter 7 Age-related Differences in Trunk Biomechanics during Walking Gait in Healthy Younger versus Older Men	• Studies investigating the relationship between ageing of the lumbar spine and loss of physical function have typically used clinical assessments • Clinical assessments and performance batteries are not specific to the lumbar spine • Few studies have investigated age-related changes in trunk kinematics during gait, and fewer still have investigated kinetic changes in older age. Therefore, the effects of age on trunk movements and kinetics during gait are not well known • Functional demand is a measure that has been applied to the lower limbs to investigate how biomechanically demanding everyday activities are. However, functional demand has never been applied to the trunk • There is a need to understand how biomechanical function of the lumbar spine is related to muscle morphology degeneration and strength loss in older age	Aims	• To investigate age-related differences in trunk biomechanics during normal walking gait • A secondary aim was to determine the functional demand of the trunk during normal walking and investigate how it is affected in older age • A further aim was to investigate the relationship between morphological degeneration of the lumbar musculature and biomechanical outcomes
		Key findings	• Trunk range of motion and peaks were reduced in all planes of motion with age • Age-related differences in trunk kinematics were most apparent in the coronal and transverse planes • Age-related differences in lateral flexion of the trunk appeared to be due to pelvic motions, whilst axial rotation reductions with age were due to trunk and pelvic alterations • Midstance and initial to mid-swing were identified as significant age-related phase-specific differences in trunk and pelvic kinematics • Controlling for age reduced the number of interplanar relationships between range of trunk and pelvis motions, suggesting that in older age trunk and pelvic movements are uncoupled during gait • Trunk moment and power waveform patterns were similar between the young and old groups • The younger group performed more negative work during the gait cycle in the coronal plane than the older group • Functional demand was greater in the older group, albeit the difference with the younger group was not significant

		• Muscle morphology does not moderate the effect of age on trunk kinematics or kinetics
	Implications	• Exercise interventions should look to target coronal and transverse plane ranges of lower back and pelvic motion • Improving ROM in older age may be beneficial although this may make walking a more functionally demanding task in the presence of lower trunk strength reserves • Reducing muscle atrophy or improving composition of the LPMs may not improve trunk biomechanics in older adults during normal gait, however, improving trunk strength may make walking less functionally demanding on the trunk

Chapter 8 General Discussion

Sarcopenia is a major health concern and socioeconomic burden (Pinedo-Villanueva et al., 2019; Janssen et al., 2004). As the musculoskeletal system declines with advancing age (Cruz-Jentoft and Sayer, 2019) and the global population is ageing (Department of Economic and Social Affairs Population Division, 2019), the prevalence of sarcopenia is increasing and already affects a large proportion older adults (Sobestiansky, Michaelsson and Cederholm, 2019). Despite the importance of the LPMs in maintaining physical function and reducing adverse health risks in older age (Katzman et al., 2012; Hicks et al., 2005b), sarcopenia research typically focuses on the appendicular muscles (Cruz-Jentoft et al., 2019). As a result, age-related declines in muscle morphology, strength and biomechanical function are not well characterised in the lumbar spine. This thesis addresses the gaps in the literature; exploring the effects of age on lumbar spine specific measures of sarcopenia. This was achieved through a series of experimental chapters, each with specific aims and objectives, which are outlined in the thesis map (**Table 7.10**). In order to analyse the complex data, a range of novel analytical techniques were applied (e.g. 3-level meta-regression hierarchical clustering, cumulative density power spectrum analysis andSPM), which added to the novelty of this thesis. A graphical overview of the main data collection methods and interrelationships between the primary measures are presented in **Figure 8.1**.

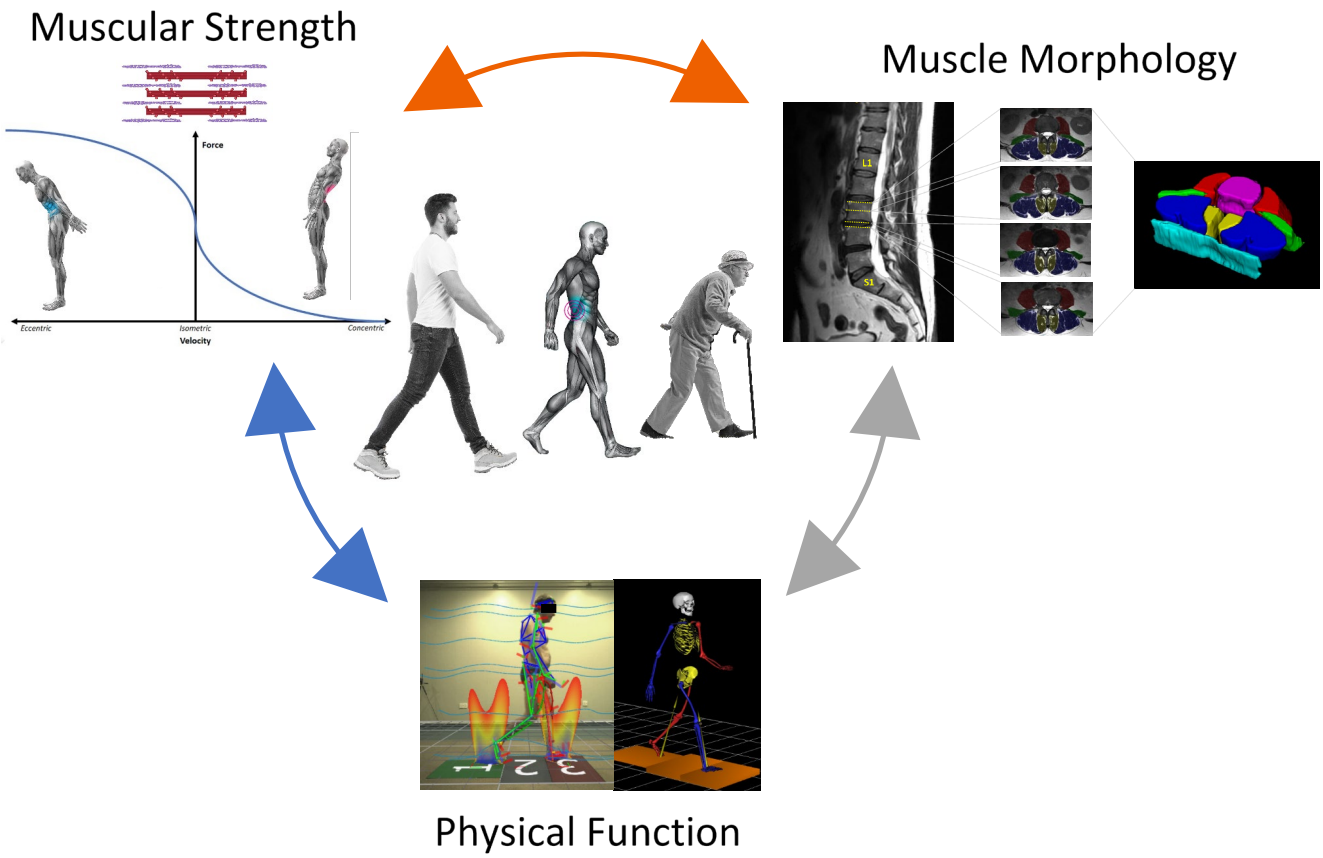

Figure 8.1 *The graphical overview represents the three main measures obtained during the project that were specific to the lumbar spine: muscular strength of the trunk assessed using isokinetic dynamometry, MRI derived morphology of the lumbar paravertebral muscles and biomechanical function of the trunk during gait using 3-D motion analysis. The current findings suggest that relationships exist between each variable, although the magnitudes of the effects varied considerably, and the directions could not be determined due to the cross-sectional study design*

8.1 Thesis Synthesis

The work in this thesis fills a substantial gap in the literature and represents an original contribution related to the ageing musculoskeletal system. For the first time, specific measures of muscle morphology, strength and biomechanics of the lumbar spine have been investigated in relation to healthy ageing. Experimental chapters focused on one of these specific aspects and whilst each of these were original as standalone studies, the thesis draws together findings to provide new insights and a holistic view on spinal sarcopenia. Using analytical techniques in novel ways also revealed important insights that have not been previously observed using conventional indices (e.g. gait speed and Short Physical Performance Battery to measure physical function, handgrip strength to measure

muscle function, and BIA to measure ALM). The nature of this research is diverse; sarcopenia encompasses three distinct areas that are interrelated. Applying the approach undertaken in the current thesis to the spinal musculature was challenging and pushed the boundaries of knowledge on this topic. Rather than simply using performance batteries, clinical strength assessments, ultrasound devices and traditional meta-analytical methods, gold-standard approaches were adopted to address some of the limitations of previous research whilst also increasing the fidelity of the data presented. Covariates were also considered and controlled for in each study to isolate the effect of older age on each of the key outcomes. This represents an additional contribution as prior work has not tended to appropriately consider confounders in its examination of this topic. This thesis therefore provides novel insights into this ageing phenomenon that extends knowledge on sarcopenia and furthers the establishment of the concept 'spinal sarcopenia'.

It has been shown that muscles in the lumbar spine undergo degenerative remodelling (i.e. atrophy and fat infiltration) as a normal response to ageing. This finding, presented in **Chapter 4,** represents a significant contribution to the ageing musculoskeletal system literature. The systematic review and meta-analysis was the first to establish and quantify the relationship between healthy ageing and morphological degeneration of the LPMs. However, the considerable methodological variation amongst studies and lack of consistency regarding the term 'healthy' made it difficult to distinguish between features of age-related and pathological degeneration, as well as artefacts of the methods used. This warranted the need for further research in a healthy population.

A unique aspect of **Chapter 4** was the meta-regression model. Rather than aggregating effect sizes from interdependent study populations, which is the conventional approach, a novel 3-level model was applied to account for statistical dependency. This allowed several moderators to be accounted for simultaneously without the need to aggregate data and perform multiple sub-group analyses. The results showed that the LPMs undergo age-related degeneration in healthy adults with muscle, lumbar level and sex-specific responses. The meta-analysis also revealed that high-resolution imaging modalities should be used to analyse LPM atrophy and fat infiltration. These were important findings in the context of the thesis as they were used to inform methodological decisions in **Chapter 5**.

MRI analysis was used to measure fat infiltration and muscle volume in the LPMs based on findings from the meta-analysis. Unlike the appendicular muscles, there are numerous factors that make MRI analysis in the LPMs challenging. Primarily, these muscles are deeper and poor imaging resolution can result in inaccurate segmentation. Despite advancements in imaging technologies and analysis techniques, the intricate geometry of the back muscles meant that automated procedures were not possible. Indeed, no automated methods have been successfully reported in the literature for

segmentation of the LPMs, which is one of the reasons why studies report CSAs at a single representative slice and combine the LPMs; it is time consuming and not always practical to provide volumetric data for each individual muscle. Whilst automated segmentation of the PS, QL, ES and MF was initially attempted, this approach was inaccurate as muscle boundaries were indistinguishable using thresholding and geometric techniques. Manual segmentation using T2-weighted axial slices was therefore performed to identify muscle boundaries, following the guidance of leading researchers (Berry et al., 2018; Crawford et al., 2017).

Using known image analysis techniques (Valentin, Licka and Elliott, 2015; Gibbons et al., 1997; Kim et al., 2019), the results were comparable to other studies that had analysed LPM size and fat composition. Furthermore, the findings were consistent with the literature (**Chapter 4)**, indicating that age predominantly modifies the fat composition of the muscles in the lumbar spine (i.e. increased fat infiltration) and causes atrophy of the ES and QL. Whilst this finding was not novel on its own, this study was the first to provide volumetric and fat infiltration data for all of the main LPMs; making the dataset unique. A surprising finding was that VPA did not covary with age-related muscle degeneration. The independence of PA on LPM morphology adds to the growing conflicting evidence questioning the effect of exercise interventions for spinal muscles, particularly those involving less volition like the shorter, deeper fascicles of the MF (Dahlqvist et al., 2017; Lee et al., 2017; Anderson et al., 2013). Following the results of this chapter, there was a need to understand how changes in muscle morphology related to changes in muscle function. Therefore, the next step in the thesis was to understand how this data related to age-related differences in dynamic trunk strength.

Loss of strength is a well-known characteristic of ageing (Keller and Engelhardt, 2013), however, there is a paucity of available literature that has examined the age-related loss of dynamic trunk strength. **Chapter 6** contributes original findings to the literature; no study has previously investigated the effect of age on eccentric strength in the LPMs. Isokinetic dynamometry was applied to the trunk using a range of angular velocities and both concentric and eccentric contractile modes. The range of test conditions was chosen to reflect the kinematics of the trunk and contractile function of its muscles during ADLs. The lumbar spine musculature is crucial to the performance of everyday activities (Hicks et al., 2005a; Ikezoe et al., 2015; Panjabi, 1992; Cholewicki, Panjabi and Khachatryan, 1997), therefore loss of strength may have an inordinately great impact in older adults manifesting as increased falls risk, and loss of mobility and independence (Suri et al., 2009; Granacher et al., 2013).

In this chapter, the OG exhibited significant reductions in peak concentric extensor torque and the difference with the YG generally increased with faster movements. This is unsurprising given the shift towards a slower muscle phenotype in older age (Evans and Lexell, 1995; Mitchell et al., 2012; Unhjem

et al., 2015). There was also an apparent preservation of eccentric strength in older age, consistent with research in the lower limbs. A key part of this thesis was to understand how age-related differences in trunk strength related to functional movement. The application and method used to derive FD was unique. Whilst FD has been calculated before, it has only been achieved in the lower limbs and typically derived from isometric strength tests that lack functional relevance. The methods used in this thesis are more representative of the trunk muscles' function during ADLs and represent an original contribution to the literature.

Another unique aspect of this chapter was relating strength changes in the trunk extensor muscles to the morphology data in **Chapter 5**. Fat infiltration was related to the loss of eccentric trunk strength, despite eccentric strength being somewhat preserved in older age. However, neither atrophy nor fat infiltration was able to explain loss of concentric trunk extensor strength with age. Since this mode of contraction was where the largest declines in strength were observed, these findings are of clinical importance in understanding how to offset strength losses in the lower back. It may therefore be possible to identify and even predict degradation/loss of function early through the combination of MRI and isokinetic data; such multi-modal analysis is not considered clinically. Furthermore, the results suggest that changes in eccentric extensor strength may be a modifiable feature of strength loss in the trunk, unlike concentric extensor strength. Other age-related mechanisms were likely responsible for the loss of concentric extensor torque such as neuropathic processes (Hunter, Pereira and Keenan, 2016; Roos et al., 1997). Exercise interventions should therefore target the back muscles, not necessarily to improve their macroscopic structure, but to improve force producing capacity through other mechanisms which concomitantly decline with age (e.g. neural drive) (Hunter, Pereira and Keenan, 2016; Roos et al., 1997). The practical implications of this are particularly useful in improving the effectiveness of training programmes in older populations. Rather than using traditional resistance-based exercise interventions focused on muscle hypertrophy and slow movement speeds (Katula, Jack and Marsh, 2008), future research should explore the efficacy of power-based training on improving trunk strength and function in older adults. Indeed, power training has been recommended for older adults (Donnelly et al., 2009; Miszko et al., 2003) as the high-speed movements elicit positive responses in physical function (Bean et al., 2009) more effectively than conventional strength training (Rice and Keogh, 2009). These high-speed movements decrease fast-twitch motor unit recruitment thresholds and increase firing rates (Van Cutsem, Duchateau and Hainaut, 1998). Such neuromuscular adaptations are likely to bring about improvements in power production and strength, which are positively related to improved functional capacity in older adult individuals (Häkkinen et al., 2001; Kyröläinen et al., 2005; Aagaard et al., 2010). It is not only

interesting, but important, to establish whether non-conventional training interventions could elicit similar positive responses in the trunk and in turn physical function in older adults.

Independent living in older age relies on maintaining a sufficient level of physical functioning (Vaughan et al., 2016). As the LPMs are important in maintaining physical function in older age (Ikezoe et al., 2015; Hicks et al., 2005b), it was important in the final experimental chapter to investigate how biomechanics of the lumbar spine is affected by age as well as how changes in muscle morphology and strength moderate the effect. Only one study has applied the conceptual framework of sarcopenia to investigate age-related changes in the lumbar spine (Shahtahmassebi et al., 2017). However, Shahtahmassebi et al. (2017) assessed physical function using a performance battery (i.e. Berg Balance Scale), clinical assessments (e.g. Timed Up and Go Test) and spaciotemporal measures (i.e. gait speed). Whilst such measures provide a general understanding of sarcopenia, they are not robust enough or have the fidelity to make specific conclusions on the age-effect in the lumbar spine.

In this thesis, 3-D motion analysis was used to investigate biomechanical function of the lumbar spine. No study has analysed age-related differences in movement data, moments, powers, and work performed in the trunk during walking gait; therefore, the experimental approach in this chapter is a novel contribution to the literature. SPM was also used to compare age-related differences across entire waveforms which provided detailed information on phase-specific age-responses. Typically, studies have analysed discrete values rather than considering the entire movement waveform. Longitudinal ageing studies with large cohorts from a general population, such as the English Longitudinal Study of Ageing (Weber, 2016) and Baltimore Longitudinal Aging Study (Jerome et al., 2015), have primarily focused on spaciotemporal parameters in gait analysis. This approach may have advantages when analysing large amounts of data, however, the opportunity to uncover important phase-specific age-effects may be missed. Discrete spaciotemporal measures may not be sensitive enough to distinguish the age-effect in healthy older versus younger adults.

The results of this chapter indicated that older adults adopt conservative upper body strategies in the coronal and transverse planes of motion, whilst maintaining forward progression through increased anterior trunk tilt. Reductions in coronal plane trunk movement in the OG were predominantly due to pelvic obliquity, whilst decreases in axial rotation were due to decrements in the out-of-phase relationship between the trunk and pelvis in the transverse plane. Indeed, peak movement amplitudes were reduced in all planes of motion in the OG. This was indicative of a conservative gait strategy, which was substantiated by lower peak functional trunk moments and powers in the OG.

Data derived from 3-D motion analysis were used in combination with the isokinetic data in **Chapter 6** to calculate the FD of walking gait in the trunk. NJMs of the lower back were matched on contractile

mode and angular velocity to the corresponding isokinetic condition, providing an accurate interpretation of FD. Despite having lower peak flexion/extension moments during the GC the OG found walking more functionally demanding than the YG, albeit not significantly. This was likely an artefact of lower trunk strength in the OG. Over longer durations, the cumulative effects of greater FD in the trunk may make everyday activities such as walking challenging in older age. Furthermore, uncoupling of interplanar motions in the trunk and pelvis with age may make walking less efficient and further increase its biomechanical challenge. This is a relatively novel assertion and mechanisms substantiating it have not yet been established.

Similar to the relationship with strength measures, muscle morphology variables were unable to explain changes in trunk biomechanics with age. Age however significantly influenced the loss of dynamic trunk strength and trunk kinematics during gait. Although speculative, this suggests that neuropathic processes may alter muscle function in older age more so than changes in muscle morphology. Efforts should focus upon improving physical function and strength in older age through other means (e.g. neuromuscular adaptations to exercise interventions), rather than concentrating primarily on improving skeletal muscle size and fat composition through conventional resistance-based interventions. The findings from each chapter illustrate that healthy older men experience age-related declines in LPM morphology and trunk strength, but these have modest impacts on physical function (assessed by walking gait). Therefore, reduced trunk movement during gait may be a selective conservative strategy in older age, rather than a consequence of muscle degeneration and strength loss, that reduces internal loading whilst maintaining a similar walking speed to healthy younger men.

8.2 Thesis Limitations and Future Research

Whilst the work in this thesis was rigorous and efforts were made throughout to ensure the highest standards of research were maintained, there were general limitations that should be acknowledged.

8.2.1 Sample Population

As with all studies, there is a compromise between measurement complexity, sustainability and resources. A fundamental understanding of spinal sarcopenia was sought in this thesis. Therefore, high data fidelity and measurement complexity was chosen over a larger dataset meaning some of the analyses were underpowered due to low sample size. The diverse nature of this thesis and unique focus of each chapter meant that it was not possible to select a sample size that would be sufficiently

powered for all outcomes within the limits of this project. Including 24 participants was the maximum sample size permissible due to the time constraints of a 3-year studentship and availability of hospital facilities. Efforts were made to match the groups as closely as possible, based on criteria most likely to affect skeletal muscle related variables (i.e. sex, PA level and ethnicity). Despite sample size being potentially low, the matching procedure increased confidence in the findings. The high level of homogeneity within and between groups meant that observed differences were likely due to ageing and not artefacts of high variability or confounding variables.

The sample lacked individuals more representative of the general population as participants were highly selected for health. Given that the aim of the thesis was to investigate normal age-related differences in musculoskeletal outcomes, this is not necessarily a limitation. However, the participants included can only reflect a specific proportion of older adults with any degree of confidence and therefore the current findings should only be generalised to healthy older white men. In addition, most participants were highly active which further limits generalisation of the findings beyond this subgroup. The generally high strength and activity levels of the OG may also explain why only a few biomechanical variables differed significantly with the YG. Whilst lower than the YG, muscle strength in the OG may not have passed the threshold of low physical performance (Cruz-Jentoft et al., 2019).

Although many confounding factors were controlled for, other covariates such as dietary information were not considered in this thesis. Protein supplementation has been shown to attenuate sarcopenic effects in older adult populations (Robinson, Cooper and Aihie-Sayer, 2012; Yanai, 2015), therefore differences in nutritional status may have influenced observations. Future studies should investigate whether nutritional status covaries with changes in LPM morphology, trunk strength and biomechanical function with ageing.

8.2.2 Protocol

Trunk strength was assessed in the sagittal plane. Measurement of trunk lateral flexion and axial rotation strength was not possible using the equipment available. Although isokinetic dynamometers have been used to measure torque in these other planes (Ellenbecker and Roetert, 2004; Huang and Thorstensson, 2000), there is currently no standardised approach to this form of assessment and the fidelity of prior works' data are questionable. Future research should look to measure trunk strength in the coronal and transverse planes using valid procedures, which would also allow FD to be calculated in these planes of motion. Secondly, normal walking gait was the only functional task that was assessed. Due to the healthy and active status of the participants, age-related differences in trunk function were not as pronounced as expected. Indeed, walking speed did not significantly differ

between the groups and it is generally considered one of the most sensitive indicators of age-related gait dysfunction (Pirker and Katzenschlager, 2017). More challenging ADLs, such as negotiating stairs, may have revealed greater functional differences in the trunk between the YG and OG. Relationships between muscle morphology, trunk strength and trunk function may have also been more apparent in more challenging scenarios. Therefore, future research should assess a range of tasks that are representative of ADLs. The compromise between external and internal validity must also be acknowledged as a limitation. To obtain a high level of internal validity, laboratory-based assessments were conducted which may have reduced the real-world application of strength and function measures. An example of this is seen in **Chapter 7**, where walking in minimal clothing, barefoot, with markers attached and in a laboratory is not representative of real-world environments. To increase external validity, future studies may consider the use of inertial measurement units which can be seamlessly worn as individuals go about their daily activities. However, these methods have inherent problems such as cumulative drift (Fong, Ong and Nee, 2008) which may affect the fidelity of the data.

Whilst efforts were made to control for PA participation prior to attending testing sessions, it cannot be confirmed that participants refrained from strenuous and atypical PA prior to undergoing the assessments. Similarly, it cannot be confirmed that participants did not consume food or caffeine prior to testing. Participants provided written and verbal confirmation that they had not performed strenuous PA, consumed caffeine or food prior to undergoing assessments. Collecting oxygen saturation, blood lactate and glucose levels through capillary blood sampling methods may have provided an objective means of screening for strenuous PA, caffeine and food ingestion.

8.2.3 Group Analysis

Participants were dichotomised into older and younger age groups. Whilst this approach is common in the literature to observe the effects of ageing (Valentin, Licka and Elliott, 2015; Hiepe et al., 2015; Anderson et al., 2013; Ikezoe et al., 2012; McGibbon and Krebs, 2001; Schmid et al., 2017) and provides a useful starting point, a greater age range and representation of ages between 30 and 60 years is needed to observe the continuous age effect. Since this approach would have required more funding and time, it was not feasible within the scope of this project. However, the variables that showed the greatest age-related differences should be focused on in future studies including a greater range of ages and specifically covering the fourth to sixth decades of life. Between the ages of 30 and 60 years has been identified as an important period in which the musculoskeletal system starts to decline (Doherty, 2003). Therefore, this age group should receive particular attention regarding age-related changes in LPM morphology, trunk strength and biomechanical function.

8.2.4 Study design

Using a cross-sectional design allowed important age-related differences to be revealed. A longitudinal design would have been superior in assessing age-related changes in muscle morphology, trunk strength and gait biomechanics. However, conducting a longitudinal study of sufficient time would have been outside the remit of this thesis. Research on this topic is still emerging, therefore the data collected is particularly valuable as it incrementally increases the knowledge base in a field that is largely unexplored. Whilst the study design precluded examination of causal relationships, it was an important first step that enabled initial investigations to be performed. Future studies should adopt a longitudinal design, exploring the causative effect of ageing on the most important variables highlighted in this thesis.

8.3 Conclusion

There are wider implications of the work in this thesis. It furthers our understanding of sarcopenia, laying the first steps in establishing the age effect on muscle morphology, strength and biomechanical function in the lumbar spine. This will also allow identification of pathological deviations and sarcopenia in the lumbar spine as the current data are representative of a healthy population. This is highly valuable in clinical settings where knowledge of the lumbar musculature in relation to ageing has not been established. Whilst each experimental chapter contributes an original methodological approach or finding, the thesis as a whole extends the concept of 'spinal sarcopenia' which may help to distinguish it as a separate condition from sarcopenia going forward. Also, the findings provide compelling evidence that can help to focus targeted interventions in older adult populations. Given the important role of the LPMs in ADLs, improving trunk strength and biomechanical function may have considerable benefits in older age such as reducing falls risk and preserving independence. This may also have a positive impact at a societal level as associated primary, secondary and tertiary healthcare costs would be expected to reduce.

In summary, this thesis has explored an emerging and multifactorial phenomenon using gold-standard approaches and novel techniques to make unique contributions to the literature. Research aiming to establish normative and sarcopenic features of age-related degeneration in the LPMs, as well as randomised-controlled trials aiming to combat the pervasive nature of these changes, will be of increasing importance as the proportion of older adults continues to rise in the UK and indeed globally.

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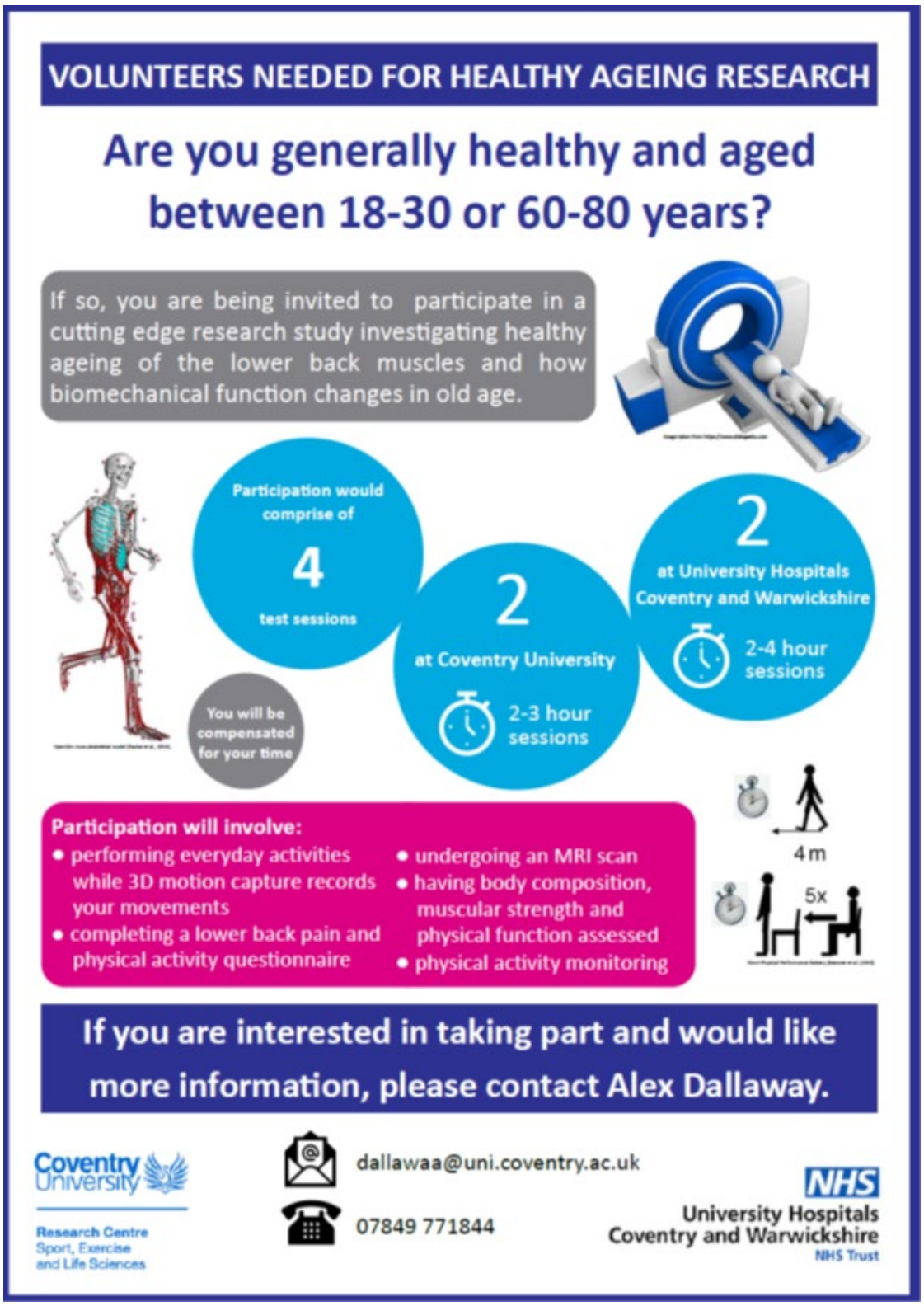
Appendices

Appendix a *Recruitment Poster*



Appendix b *Participant Information Sheet*

Coventry University

Research Centre
Sport, Exercise
and Life Sciences

NHS
University Hospitals
Coventry and Warwickshire
NHS Trust

Participant Information Sheet

Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in Older Adults

You are being invited to take part in this research study. Before you decide, it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is unclear, or if you would like more information please do not hesitate to contact us.

Contents

ABOUT US

Principal Investigator

Alexander Dallaway *(PhD Student, Centre for Sport, Exercise and Life Sciences, Coventry University)*

Other researchers working on the study

Professor Michael Duncan *(Professor of Sport and Exercise Science, Coventry University)*
Dr John Hattersley *(Head of Human Metabolism Research Unit, University Hospitals Coventry & Warwickshire)*
Dr Jason Tallis *(Senior Lecturer in Biomechanics, Coventry University)*
Professor Derek Renshaw *(Professor of Translational Physiology, Coventry University)*

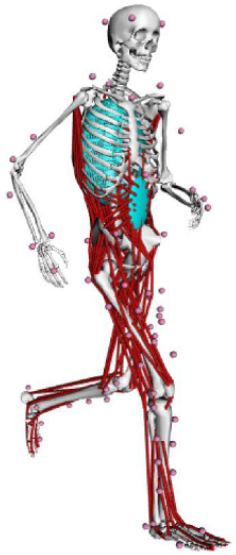
ABOUT THE STUDY

What is the study about?

This study will investigate the age-related changes that occur in the muscles surrounding the lumbar spine and how these changes impact upon physical function and loading in the lumbar spine.

Why is this study important?

The muscles in the lower back provide postural stability and allow movement of the trunk. As we age these muscles may go through degenerative changes that compromises physical function. Research has shown that ageing leads to a general decline in the ability to perform everyday tasks and that the size and quality of muscles deteriorates. However, the relationship between physical function and healthy ageing of the lower back muscles has not been fully explored. To investigate this relationship, accurate measurements of the muscles in the lower back and how people move during everyday tasks is essential. This project will provide a comprehensive insight into how and why age-related degeneration of the lower back muscles affects the ability of individuals to perform activities of daily living. This understanding will inform efforts to extend the period the elderly are able to live independently.



What is our aim?

The aim of this study is to investigate how the muscles surrounding the lumbar spine change in old age, and how these age-related changes impact upon physical function and lower back pain.

ABOUT YOU

Why have I been chosen?

We are looking for…

- *12 individuals who are aged between 60 - 80 years*

and…

- *12 individuals who are aged 18 - 30 years*

Unfortunately, you will <u>not</u> be eligible to participate in this study if…

- *You are not within the age ranges 18 - 30 years or 60 - 80 years*
- *Your BMI is not between 18.5 - 30 kgm^{-2}*
- *You are dependent on others to perform everyday activities*
- *You are unable to undergo MRI based on the MRI screening questionnaire*
- *You are a current or ex-smoker who has ceased smoking less than 6 months ago*
- *You consume alcohol daily*
- *You have existing or a past medical history of cancer; diabetes; liver, kidney, vascular, pulmonary, digestive (Coeliac disease) or thyroidal diseases; osteoporosis or history of falls*
- *You have a neuromuscular disorder/injury or physical impairment that limits your normal daily activities*
- *You are unable to provide informed consent and follow study procedures*

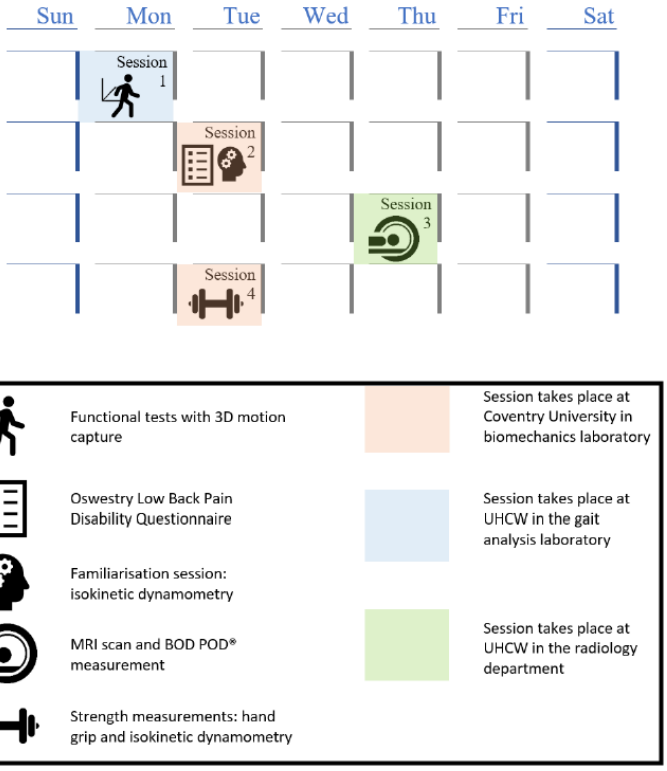
We will check your eligibility to participate in this study by asking you to complete a health and lifestyle questionnaire. If you are in doubt about whether or not you are eligible, please do not hesitate to contact the principal investigator.

Do I have to take part?

It is up to you to decide whether or not to take part. If you decide to take part, you are still able to withdraw at any time without having to give a reason. There are no future implications if you choose not to take part or withdraw from the study at any time. If you are interested in taking part but still undecided, an expression of interest does not commit you to participating in the study. If you have any questions about the study, please contact the principal investigator who will be happy to answer them.

What do I have to do?

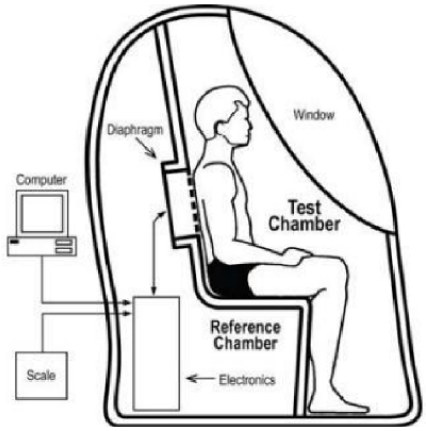
You will be required to complete four testing sessions. The sessions will comprise of body composition measurements, strength testing, biomechanical analyses and the completion of self-reported questionnaires. Measurements are not restricted to a specific order as there is no intervention. The only exception is that the motion capture testing must take place before the strength testing. An example of what you will undergo is shown below. The order and timings of these testing sessions is flexible, although it is better to complete the sessions within a relatively small time frame to avoid variations that may occur due to changes in health status and body composition.



Example of order of testing and familiarisation sessions. *note that testing sessions may take place on other days in the week except for MRI scanning as the hospital has an allocated research slot on a Thursday afternoon each week.

Body composition/ muscle morphology measurements

You will undergo an MRI scan of the lumbar spine at University Hospitals Coventry and Warwickshire. The scan will last approximately 50 minutes during which you will be lying down comfortably on your back. You will also undergo a body composition measurement in a BOD POD® machine, lasting around 15 minutes. You will be seated and asked to wear tight clothing (e.g. swimming costume) and a swimming cap.



Participant sitting position in the BOD POD®. Image adapted from http://ybefit.byu.edu/portals/88/documents/how%20does%20the%20bod%20pod%20work.pdf

Physical function tests and 3D motion analysis

You will be asked to perform a range of activities that mimic everyday tasks. These activities include: walking a distance of 4 metres, carrying a load whilst walking a distance of 4 metres, standing up from a seated position and sitting down again, balancing with both feet in different positions and balancing on one foot. You may be required to perform a test multiple times. Whilst performing these activities you will have markers attached to your skin or skin-tight clothing. Hypoallergenic medical grade tape will be used to reduce skin irritation. Cameras will be positioned around you so that your movements can be recorded. Your performance will be scored and analysed for each task. The entire motion capture process is expected to take approximately 4-5 hours.

Strength measurements

You will perform maximum voluntary contractions whilst connected to an isokinetic dynamometer. This machine will measure the force you produce at your ankle, knee, hip and lumbar spine for flexion and extension movements. At least 10 days prior to testing, you will be required to complete a familiarisation session so that you are comfortable and understand the testing process. On the day of the test, you will perform one more familiarisation session. For the test you will be asked to perform maximum voluntary contractions for each of the joints specified. Restraints will be used to restrict the unwanted movement of other body parts contributing to the force production of the muscle group being tested.

Your handgrip strength will be measured using a handgrip dynamometer. You will perform 6 maximal effort trials, alternating between hands. For each trial you will grip the dynamometer and squeeze for a brief period.

Physical activity monitoring

You will be asked to complete the International Physical Activity Questionnaire (short form) to give a measure of your physical activity level. You will also wear an accelerometer on your wrist for 7 consecutive days which will record data that can used to determine how active you have been.

Self-reported lower back pain

You will be asked to complete the Modified Oswestry Low Back Pain Disability Questionnaire. This is to evaluate how much lower back pain affects your physical function.

Will my details be confidential?

Your personal data including non-sensitive information (e.g. name, address and contact details) will be handled only by the researchers identified in this study. All data will be treated as strictly confidential and stored in a locked filing cabinet within a restricted access room or on a password protect computer in which only the identified researchers have access. Upon enrolment onto the study, you will be assigned a randomly-generated 3-digit number. All data and results related to the study will be coded with this number to ensure your identity remains anonymous.

Is there a financial cost to me if I want to take part?

There is no financial cost to you if you decide to take part in this study. All travel and parking expenses will be covered or reimbursed promptly.

RISKS AND BENEFITS

Are there any risks associated with this project?

***MRI** - There are no known risks associated with MRI. MRI does not expose the patient to radiation, unlike other imaging techniques such as X-rays and CT scans. You should be able to undergo an MRI scan safely unless you are excluded from doing so based on the MRI safety questionnaire. A UHCW radiologist will review the MRI images obtained during this study. If the radiologist is concerned by the appearance of any images, your GP will be informed*

***BOD POD®** - Bod Pod® uses air displacement plethysmography to assess body composition meaning it is non-invasive.*

***Physical performance tests** - The movements performed will mimic your everyday activities (e.g. walking at your usual pace, sitting down on a chair). There should be minimal risk in performing these tests and additional support will be provided in case you should need it.*

***Activity monitoring** - Accelerometers will be used to record your physical activity. They are*

small, lightweight and non-invasive.

***Strength measurements** - There is a risk of musculoskeletal injury when performing any maximal strength test. However, competent operators will adhere to institutional working practices to ensure the equipment is used in a safe manner. Sufficient warm-ups will also be given to reduce the risk of musculoskeletal injury.*

***3D motion capture** - Medical grade double-sided tape will be used to attach reflective spheres to your skin. This may cause skin irritation and potentially pain when removed. You will be asked in advance whether you have any allergies to certain materials so that a hypoallergenic alternative may be used instead. If necessary, areas of your body may be shaved to reduce skin irritation and pain when the tape is removed. Markers can also be attached to skin tight clothing.*

What are the benefits of taking part?

We cannot guarantee that taking part in this study will benefit you directly. However, by taking part you will receive a comprehensive biomechanical assessment using state-of-the-art equipment, an examination of your lower limb and back strength, and an MRI scan of your spine which will be clinically checked for abnormalities. The results of these assessments, which will be presented to you in a portfolio at the end of the study, will provide valuable information about the health of your muscles and your physical function. All participants who complete the entire study will also receive a £20 Amazon voucher as a thank you.

Who has reviewed this study?

The procedures in this study have been developed by the research team with input from research and clinical staff at University Hospitals Coventry & Warwickshire. This study has been reviewed and approved by the Ethics Committee at Coventry University and by Research & Development at University Hospitals Coventry & Warwickshire NHS Trust.

AFTER YOU HAVE TAKEN PART

What will happen with the results of this study?

The results of this research study will be published in scientific journals and presented at national and international scientific conferences. You will not be identified personally in any output related to this study.

What will I receive?

Upon completion of the study, you will be provided with a summary of the research findings and a free copy of any journal articles that are published where your data has been used. You will also receive an Amazon gift voucher worth £20 if you complete the full protocol.

WHAT HAPPENS NEXT

What if I have more questions?

If you have any further questions please get in touch with Alex Dallaway by email or phone.

dallawaa@uni.coventry.ac.uk

07849 771844

What do I do if I want to be part of this study?

If you would like to take part in this study please contact Alex Dallaway and complete the health and lifestyle questionnaire and return via email to dallawaa@uni.coventry.ac.uk

Thank you for showing interest in this study and for taking the time to read this information sheet.

Appendix c *Informed Consent Form*

Coventry University
Research Centre
Sport, Exercise
and Life Sciences

Informed Consent Form

NHS
University Hospitals
Coventry and Warwickshire
NHS Trust

Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in Older Adults

	Please tick
1. I confirm that I have read and understood the participant information sheet for the above study and have had the opportunity to ask questions.	☐
2. I understand that my participation is voluntary and that I am free to withdraw at anytime without giving a reason.	☐
3. I understand that all the information I provide will be treated in confidence.	☐
4. I understand that I also have the right to change my mind about participating in the study for a short period after the study has concluded (insert deadline here).	☐
5. I agree to be filmed/recorded (delete as appropriate) as part of the research project. Filming/photography will be used for data collection purposes and to illustrate procedures in scientific communications and posters.	☐
6. I agree to take part in the research project	☐

Name of participant: .....

Signature of participant: .....

Date: .....

Witnessed by (if appropriate): .....

Name of witness: .....

Signature of witness: .....

Name of Researcher: .....

Signature of researcher: .....

Date: .....

Appendix d *International Physical Activity Questionnaire – Short Form*

Appendix e *International Physical Activity Questionnaire – Short Form (elderly)*

Some materials have been removed from this thesis due to Third Party Copyright. Pages where material has been removed are clearly marked in the electronic version. The unabridged version of the thesis can be viewed at the Lanchester Library, Coventry University.

Appendix f GafREC Approval Letter

Content removed on data protection grounds

Coventry University
Research Centre
Sport, Exercise
and Life Sciences

NHS
University Hospitals
Coventry and Warwickshire
NHS Trust

HEALTH AND LIFESTYLE QUESTIONNAIRE

The following questionnaire will help us to determine if you are eligible to participate in the research study. Please complete each question in full.

	DAY		MONTH		YEAR			
Date questionnaire completed								

Your Contact details

Title: Forename(s): Surname:	Address: Postcode:
Landline telephone number:	Mobile telephone number:
E-mail address:	Tell us the best way to contact you:

About you:

Date of Birth:		
Age today:		
Approximately how tall are you?		BMI:
What is your approximate weight?		Office use only:

Has your weight changed in the last 6 months? **IF YES, by how much?**

Q1. Do you smoke? If you are an ex-smoker, when did you stop? A. (**if previous smoker**, please give further details of approximate no. of cigarettes smoked per day and no. of years smoked)

BRIEF MEDICAL HISTORY
(please be aware that we do NOT have access to your medical notes and any relevant information you can provide is very helpful)

For the following questions, please tell us **<u>if your GP has diagnosed you</u>** with existing or previous history of:

Q2. Heart disease (e.g. Angina), stroke or any other disease of the circulation (such as Reynaud's disease)? IF yes please give further details, including when diagnosed
Q3. Diabetes (either type 1 or type 2) IF yes please give further details, including when diagnosed
Q4. Cancer IF yes please give further details, including when diagnosed

Q5. A liver or kidney complaint / disease? IF yes please give further details, including when diagnosed
Q6. A digestive disease (such as Chron's, coeliac irritable bowel, colitis)? IF yes please give further details, including when diagnosed
Q7. Neurological disorder, severe muscular injuries or a thyroid condition? IF yes please give further details, including when diagnosed
Q8. Respiratory conditions, including asthma: IF yes please give further details, including when diagnosed
Q9. Any other illness or conditions we need to be aware of? A. (please give further details, including when diagnosed)
Q11. Has your doctor diagnosed you as having high cholesterol? A. (please give further details, including when diagnosed)

Q12. Has your doctor diagnosed you as having high blood pressure? A. (please give further details, including when diagnosed)
Q13. Do you have any known food intolerances (e.g. lactose intolerance), food allergies, or other allergies? A. (please give further details, including when diagnosed)
Q14. Are you claustrophobic?

MEDICATIONS

Q15. Have you been prescribed cholesterol lowering drugs (e.g. statins, simvastatin, atorvastatin)?
Q16. Have you been prescribed blood pressure medications (e.g. alpha blockers, beta blockers, ACE inhibitors, Ca^{2+} channel blocker)

Q17. Have you been prescribed anti-diabetic medications (e.g. insulin, exenatide, liraglutide) or oral hypoglycemic agents / oral antihyperglycemic agents (e.g. glipizide, metformin, pioglitazone)?

Q18. Are you currently on any long-term medication; including aspirin / steroids, antihistamines, anti-inflammatory medication, pain relief?

A. (**if yes**, please give further details of brand, dose, duration of use)

Q19. Do you regularly take non prescribed pain relief, anti-inflammatory, anti-histamines?

A. (**if yes**, please give further details of brand, dose, duration of use)

DIETARY OR SUPPLEMENT USE

Q20. Are you on a therapeutic diet? or '*dieting*' with the intention of substantial weight loss?

A. (**if yes**, please give further details regarding the dietary regime)

Q21. Do you currently consume protein and/or amino acid supplements?

A. (**if yes**, please give further details of brand, dose, duration of use)

Q22. Do you classify yourself as a vegetarian?
Q23. Have you been involved in an **exercise** or **dietary intervention** trial in the last 6 months? If yes, please provide more details.

EXERCISE AND PHYSICAL ACTIVITY

Q24. Do you currently participate in resistance exercise (using weights) regularly or have done within the last 6 months?
Q25. Do you participate in any other forms of regular exercise? If yes please provide details.
Q26. Has your GP told you for any reason why you should not participate in an exercise programme?

OFFICE USE ONLY – OFFICE USE ONLY – OFFICE USE ONLY

Is the volunteer eligible to proceed in the trial?

YES **NO**

Name of scientist / research nurse (PRINT): ……

Signature of the scientist / research nurse: ……

DATE:	**D**	**D**	**M**	**M**	**M**	**Y**	**Y**	**Y**	**Y**

University Hospitals Coventry and Warwickshire NHS Trust

MRI Department Volunteer Screening Questionnaire : Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in Older Adults

NAME:	D.O.B.		
ADDRESS:			
GP NAME & ADDRESS:	**MALE / FEMALE**	**Yes**	**No**
1. Do you have a heart pacemaker or artificial heart valve, or have you had a heart operation? If yes, please give details:			
2. Have you ever had any operations on your head or spine? If yes please give details:			
3. Have you ever had any operations which involved the use of metal implants (e.g. clips, pins or plates) or do you have **ANY** implants anywhere in your body? If yes please give details:			
4. Have you **EVER** had an accident involving metal fragments penetrating your eyes?			
5. Have you any metal fragments (e.g. bullets, gun shot or shrapnel) anywhere in your body? If yes please give details:			
6. Do you have a hydrocephalus shunt? If yes is it programmable? Yes/No			
7. Do you have a skin patch, dentures, body piercing, coloured contact lenses or hearing aids?			
8. Have you ever had an epileptic fit?			
Female participants only:		**Yes**	**No**
9. Is there any possibility you may be pregnant?			
10. Do you have an IUD (coil) or sterilisations clips?			

I hereby consent to undergo a Magnetic Resonance Imaging (MRI) examination. The nature of the examination has been fully explained to me and understood by me.

Volunteer's signature …….. **Date**……

To be completed by MR Staff ONLY

Accepted for scanning by:	Participants Weight:
Trial number :	

Appendix i *MRI Consent Form*

Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in Older Adults– Consent Form

Coventry University

Research Centre
Sport, Exercise
and Life Sciences

NHS
University Hospitals
Coventry and Warwickshire
NHS Trust

Participant Identification Number for this study:

Title of Project: Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in Older Adults

Name of Researcher(s): Mr Alexander Dallaway, Dr. John Hattersley, Prof. Mike Duncan, Dr. Jason Tallis, Prof. Derek Renshaw

Please initial all boxes

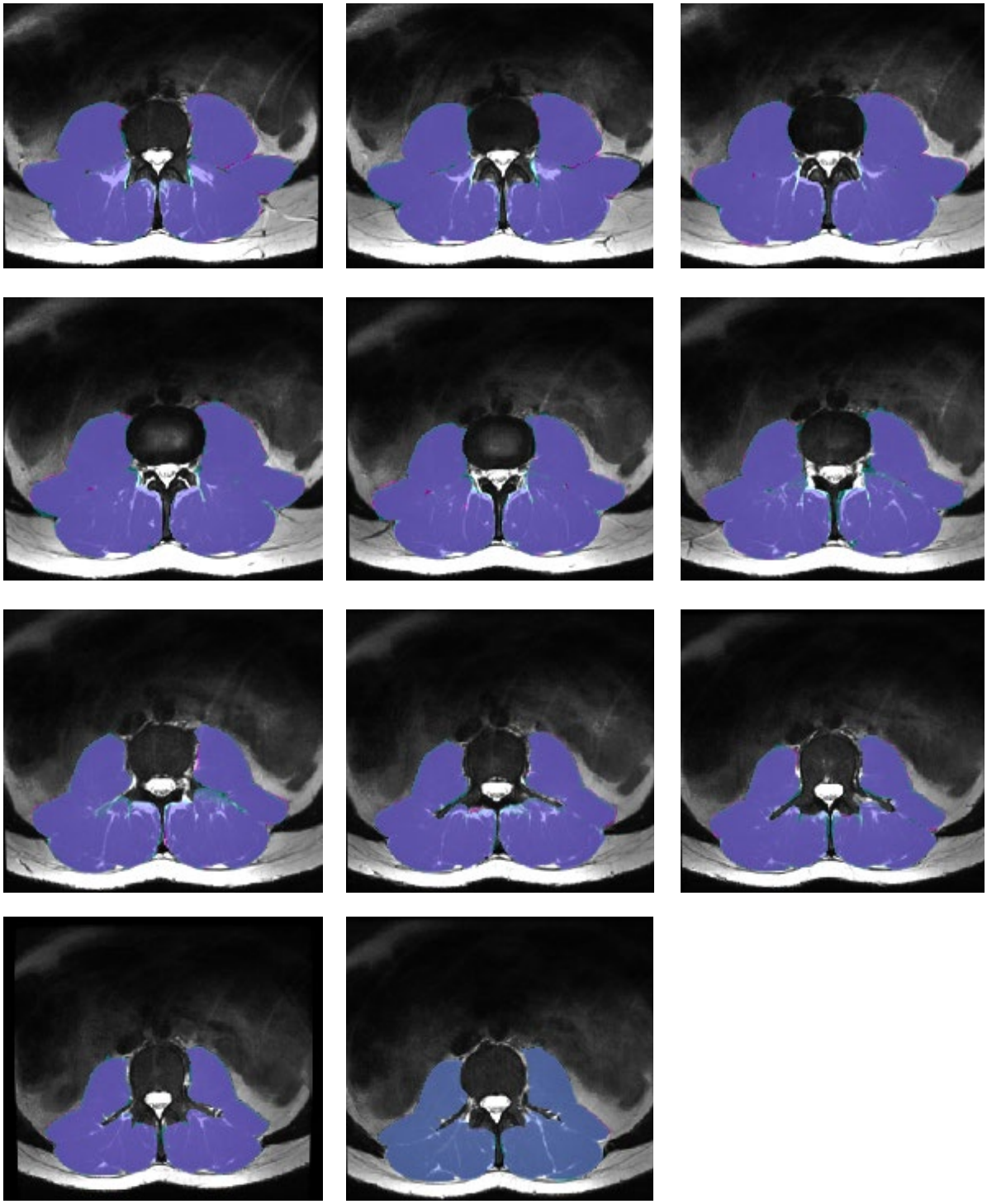
1. I confirm that I have read and understand the information sheet dated **DATE** for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily. ☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason, without my medical care, education, or legal rights being affected. ☐
3. I understand that I have to complete an MRI safety screening questionnaire immediately prior to my MRI scan. ☐
4. I consent to my General Practitioner being contacted in the unlikely event that my MRI scan reveals any unsuspected abnormality. ☐
5. I agree to take part in the above study. ☐

Name of Participant	Date	Signature
Name of Person taking consent	Date	Signature

Appendix j *Graphical Overview of Studies*

Adobe Acrobat Document

Appendix k *Segmentation agreement maps showing the test (blue) and retest (pink) segmentations. Purple areas show the cross-over between test and retest measurements*



Appendix I *Morphology values for the first and second assessments*

		First assessment				Second assessment			
		Participant 1	Participant 2	Participant 3	Participant 4	Participant 1	Participant 2	Participant 3	Participant 4
Volume (cm^3)	**PS**	109.5	101.1	143.8	96.4	109.7	101.2	144.7	96.6
	QL	50.1	43.6	74.2	62.7	51.1	42.1	76.1	64.8
	ES	193.0	150.5	216.8	188.5	191.6	148.8	214.2	189.5
	MF	49.8	64.0	61.8	64.5	48.8	63.8	60.6	65.6
Normalised Volume	**PS**	5.8	5.6	7.3	5.3	5.8	5.5	7.4	5.3
	QL	2.7	2.4	3.8	3.4	2.7	2.3	3.9	3.5
	ES	10.3	8.3	11.0	10.3	10.1	8.2	11.0	10.4
	MF	2.7	3.5	3.1	3.5	2.6	3.5	3.1	3.6
Fat Infiltration (%)	**PS**	12.5	13.8	10.7	12.3	12.7	13.7	11.5	13.3
	QL	11.5	22.7	10.0	11.4	12.1	21.8	11.4	12.7
	ES	18.5	33.1	16.1	14.2	18.7	32.1	16.5	15.3
	MF	24.6	43.6	23.6	17.9	24.6	42.4	24.1	18.9
Vertebral height (cm)		18.76	18.13	19.69	18.31	18.98	18.25	19.46	18.28
Lumbar lordosis angle (°)		42.8	37.3	21.4	28.0	46.5	37.4	21.6	29.9

Appendix m *Participant positioning in the Trunk Module Component*

Appendix n *Anthropometric measurements for the Plug-in Gait Marker Model with descriptions*

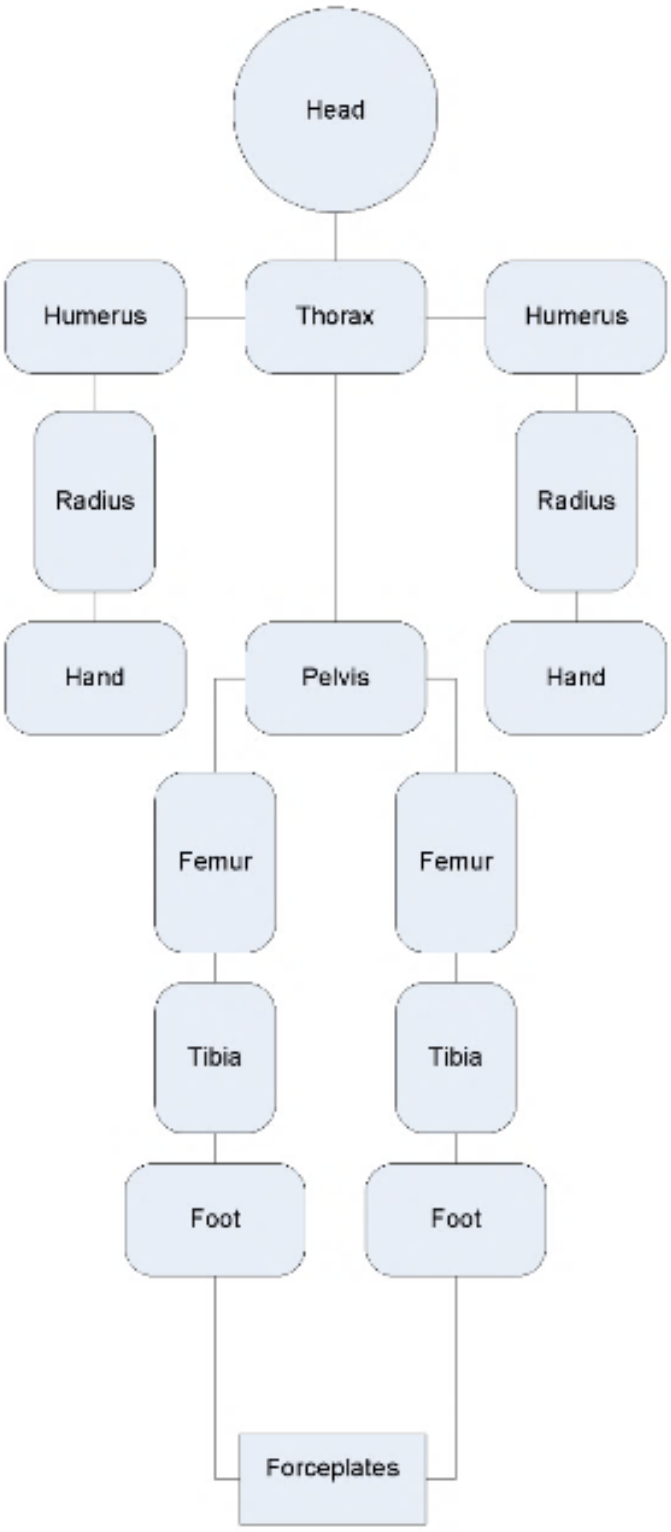
Name	Description	Measure Left	Measure Right
Body Mass	Body mass was measured using SECA scales (Hamburg, Germany) to the nearest 0.1 kg.	_____ kg	
Height	Height was recorded using a SECA stadiometer (Hamburg, Germany) to the nearest 0.01 m.	_____ mm	
Upper Body Measurements			
***Head Offset**	Patient head offset in degrees. Only required if head is not level (calculated after running the Plug-in Gait model)	_____ deg	
Shoulder Offset	Vertical offset from the base of the acromion marker to shoulder joint center.	_____ mm	_____ mm
Elbow Width	Width of elbow along flexion axis (roughly between the medial and lateral epicondyles of the humerus).	_____ mm	_____ mm
Wrist Width	Anterior/Posterior thickness of wrist at position where wrist marker bar is attached.	_____ mm	_____ mm
Hand Thickness	Anterior/Posterior thickness between the dorsum and palmar surfaces of the hand.	_____ mm	_____ mm
Arm Span	Left arm middle finger to right arm middle finger distance when arms are abducted to 90 degrees.	_____ mm	_____ mm
Lower Body Measurements			
***Inter-ASIS distance**	ASIS-ASIS distance is the distance between the left ASIS and right ASIS. This measurement is only needed when markers cannot be placed directly on the ASIS, for example, in obese patients.	_____ mm	
Leg Length	Full leg length, measured between the ASIS marker and the medial malleolus, via the knee joint. Measure with patient standing, if possible. If the patient is standing in the crouch position, this measurement is NOT the shortest distance between the ASIS and medial malleoli, but rather the measure of the skeletal leg length.	_____ mm	_____ mm
***ASIS-Trochanter Distance**	ASIS-greater trochanter distance is the vertical distance, in the sagittal plane, between the ASIS and greater trochanter when the patient is lying supine. Measure this distance with the femur rotated such that the greater trochanter is positioned as lateral as possible.	_____ mm	_____ mm
Knee Width	The medio-lateral width of the knee across the line of the knee axis. Measure with patient standing, if possible.	_____ mm	_____ mm
Ankle Width	The medio-lateral distance across the malleoli. Measure with patient standing, if possible.	_____ mm	_____ mm
***Shank Rotation Offset**	Shank Rotation Offset is automatically inputted by the Plug-in Gait Marker model as zero, as the model assumes the tibia marker is placed exactly in the sagittal plane between the knee and ankle joint centre.	_____ deg	_____ deg
***Sole Thickness Delta**	The difference in the thickness of the sole at the toe and the heel. A positive sole delta indicates that the patient's heel is raised compared with the toe. This value was assumed to be 0 since participants were barefoot.	_____ mm	_____ mm

* not required. PiG model calculates the parameter

Appendix o *Plug-it Gait marker placements and definitions*

Marker label	Definition	Position on patient
Head markers		
LFHD	Left front head	Left temple
RFHD	Right front head	Right temple
LBHD	Left back head	Left back of head (defines the transverse plane of the head, together with the frontal markers)
RBHD	Right back head	Right back of head (defines the transverse plane of the head, together with the frontal markers)
Torso markers		
C7	7th cervical vertebra	On the spinous process of the 7th cervical vertebra
T10	10th thoracic vertebra	On the spinous process of the 10th thoracic vertebra
CLAV	Clavicle	On the jugular notch where the clavicles meet the sternum
STRN	Sternum	On the xiphoid process of the sternum
RBAK	Right back	Anywhere over the right scapula (This marker has no equivalent marker on the left side. This asymmetry helps the autolabeling routine determine right from left on the subject. Placement is not critical as it is not included in the Plug-in Gait model calculations.)
Left upper limb markers		
LSHO	Left shoulder	On the acromio-clavicular joint
*LUPA	Left upper arm	On the upper lateral 1/3 surface of the left arm (Place asymmetrically with RUPA)
LELB	Left elbow	On the lateral epicondyle
*LFRM	Left forearm	On the lower lateral 1/3 surface of the left forearm (Place asymmetrically with RFRM)
LWRA	Left wrist marker A	At the thumb side of a bar attached to a wristband on the posterior of the left wrist, as close to the wrist joint centre as possible. Loose markers can be used but for better tracking of the axial rotations, a bar is recommended.
LWRB	Left wrist marker B	At the little finger side of a bar attached to a wristband on the posterior of the left wrist, as close to the wrist joint centre as possible. Loose markers can be used but for better tracking of the axial rotations, a bar is recommended.
LFIN	Left finger	Just proximal to the middle knuckle on the left hand
Right upper limb markers		
RSHO	Right shoulder	On the acromio-clavicular joint
*RUPA	Right upper arm	On the lower lateral 1/3 surface of the right arm (Place asymmetrically with LUPA)
RELB	Right elbow	On the lateral epicondyle approximating the elbow joint axis
*RFRM	Right forearm	On the lower lateral 1/3 surface of the right forearm (Place asymmetrically with LFRM)
RWRA	Right wrist marker A	At the thumb side of a bar attached symmetrically with a wristband on the posterior of the right wrist, as close to the wrist joint centre as possible
RWRB	Right wrist marker B	At the little finger side of a bar attached symmetrically with a wristband on the posterior of the right wrist, as close to the wrist joint centre as possible
RFIN	Right finger	Just below the middle knuckle on the right hand
Left lower limb markers		
LTHI	Left thigh	Over the lower lateral 1/3 surface of the left thigh
LKNE	Left knee	On the flexion-extension axis of the left knee
LTIB	Left tibia	Over the lower 1/3 surface of the left shank
LANK	Left ankle	On the lateral malleolus along an imaginary line that passes through the transmalleolar axis
LHEE	Left heel	On the calcaneous at the same height above the plantar surface of the foot as the toe marker
LTOE	Left toe	Over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot
Right lower limb markers		
RTHI	Right thigh	Over the upper lateral 1/3 surface of the right thigh
RKNE	Right knee	On the flexion-extension axis of the right knee.
RTIB	Right tibia	Over the upper 1/3 surface of the right shank
RANK	Right ankle	On the lateral malleolus along an imaginary line that passes through the transmalleolar axis
RHEE	Right heel	On the calcaneous at the same height above the plantar surface of the foot as the toe marker
RTOE	Right toe	Over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot

Appendix p *PiG kinetic hierarchy*



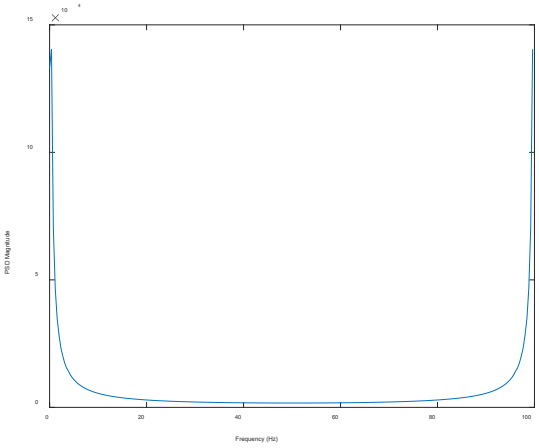
Appendix q *MATLAB code to determine optimal cut-off frequency*

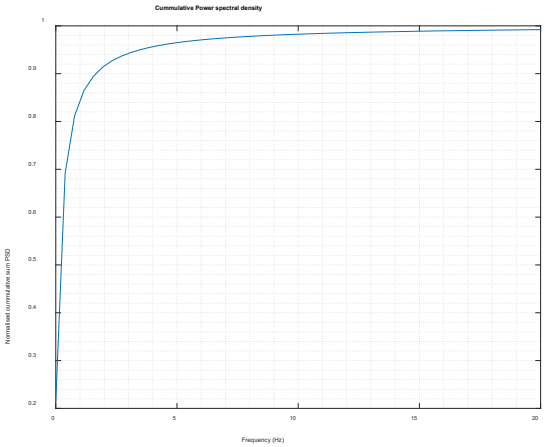
```
Fs = input ('Select signal frequency -> ');       % Signal
x = input ('Select signal vector -> ');       % Signal
    t = length (x);
    [Pxx,F] = periodogram(x,[],t,Fs);
    plot(F,10*log10(Pxx))

% Plot signal in frequency domain
y = fft(x);
f = (0:length(y)-1)*100/length(y);
plot(f,abs(y))
xlabel('Frequency (Hz)')
ylabel ('PSD Magnitude')

% Cummulative power spectrum
CP = cumsum(Pxx);        % Create cummulative power spectral
NormCP = CP/CP(end);      % Normalise
figure; plot (F,NormCP)
title('Cummulative Power spectral density')
xlabel('Frequency (Hz)')
ylabel ('Normalised cummulative sum PSD')
xlim([0,20])
grid minor

% find 99% of cummulative frequency cut-off
f99 = find (NormCP > 0.99);
idx = min(f99);
[cutoff] = F(idx)
```





Appendix r *Effect of a 4th order Zero Lag Low-pass Butterworth filter with cut-off frequency 10 Hz on the marker trajectories of a marker placed on the 10th spinous process of the lumbar vertebrae and GRF data*

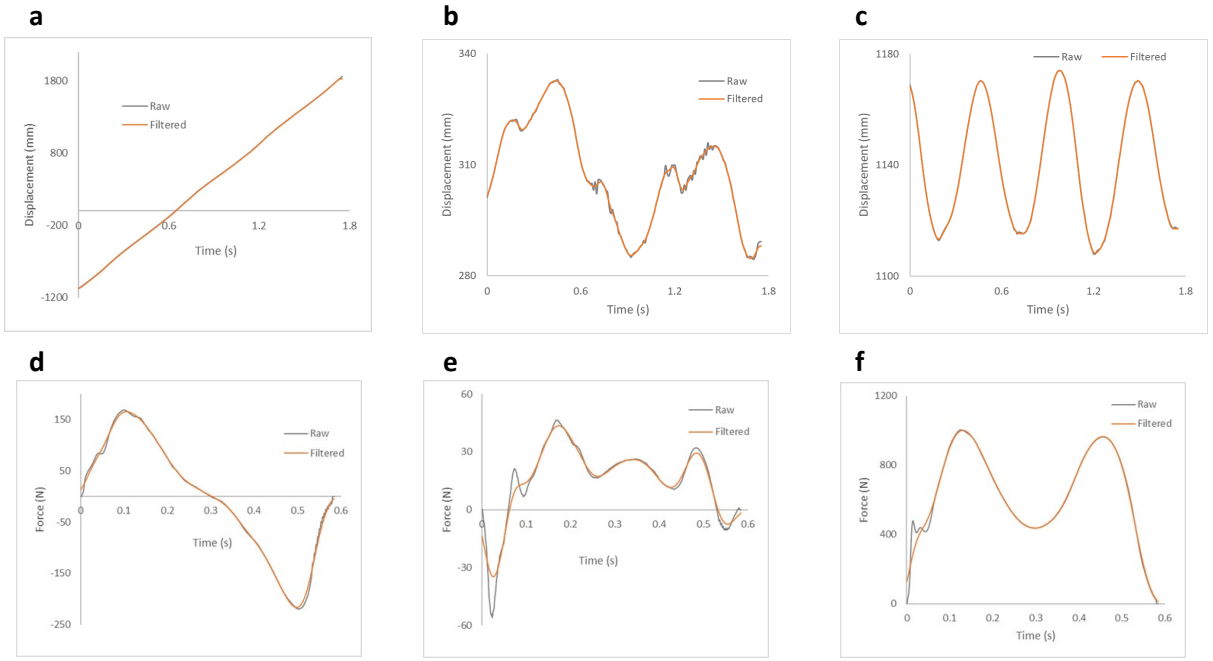


Figure a = anteroposterior displacement, figure b = mediolateral displacement, figure c = vertical displacement, figure d = anteroposterior GRF, figure e = mediolateral GRF, figure f = vertical GRF

```
%% Conduct SPM analysis
spm        = spm1d.stats.ttest2(Young, Old);
spmi       = spm.inference(0.05, 'two_tailed',true, 'interp', true);
disp(spmi)

% Plot SPM results
close all
spmi.plot();
spmi.plot_threshold_label();
spmi.plot_p_values();
spmi.clusters{1,1} % For descriptive information about clusters
```

Certificate of Ethical Approval

Applicant:

Alexander Dallaway

Project Title:

Healthy Ageing of the Lumbar Paravertebral Muscles and Physical Function in an Elderly Community-dwelling Population: a biomechanical approach into the functional consequences of ageing lumbar paravertebral muscle morphology

This is to certify that the above named applicant has completed the Coventry University Ethical Approval process and their project has been confirmed and approved as Medium Risk

Date of approval:

13 September 2018

Project Reference Number:

P70399

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Medium to High Risk Research Ethics Approval Checklist

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Appendix u Dallaway, A., Kite, C., Griffen, C., Duncan, M., Tallis, J., Renshaw, D., and Hattersley, J. (2020) 'Age-Related Degeneration of the Lumbar Paravertebral Muscles: Systematic Review and Three-Level Meta-Regression'. *Experimental Gerontology* 133, 110856

Experimental Gerontology_manuscı

Appendix v Dallaway, A., Hattersley, J., Diokno, M., Tallis, J., Renshaw, D., Wilson, A., Wayte, S., Weedall, A., and Duncan, M. (2020) 'Age-related Degeneration of Lumbar Muscle Morphology in Healthy Younger versus Older Men'. *The Aging Male* 23 (5), 1583-1597

Aging Male_accepted manu

Appendix w Dallaway, A., Hattersley, J., Tallis, J., Renshaw, D., Griffen, C., and Duncan, M. (2021) 'Age-related changes in concentric and eccentric isokinetic peak torque of the trunk muscles in healthy older versus younger men'. *Journal of Aging and Physical Activity* 30, 1-11

JAPA_manuscript proof.pdf