LUMBOPELVIC MUSCLE FUNCTION DURING LOW IMPACT WEIGHT-BEARING EXERCISE: DEVELOPMENT OF THE FUNCTIONAL RE-ADAPTIVE EXERCISE DEVICE

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LUMBOPELVIC MUSCLE FUNCTION DURING LOW IMPACT WEIGHT-BEARING EXERCISE: DEVELOPMENT OF THE FUNCTIONAL RE-ADAPTIVE EXERCISE DEVICE

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ABSTRACT

The aim of this thesis was to develop our understanding of the Functional Re-adaptive Exercise Device (FRED): a novel prototype exercise device proposed to facilitate the activation the deep paraspinal and anterolateral abdominal wall musculature in a manner consistent with the requirements of motor control training in people with low back pain.

Firstly, the intra- and interday reliability and precision of measurement of ultrasound imaging of the lumbar multifidus (LM) and transversus abdominis (TrA) were established. LM and TrA demonstrated good (ICC $\geq$ 0.75) to excellent (ICC $\geq$ 0.9) intrarater reliability for both intra- and interday measurements of absolute linear muscle thickness across all conditions. Normalised thickness change, expressed relative to resting values, also demonstrated good reliability between days, with ICCs in excess of 0.75 across all conditions.

Secondly, the typical nature of LM and TrA function during this mode of exercise was evaluated in relation to commonly used assessment techniques such as the abdominal drawing-in manoeuvre, active straight-leg raise, and contralateral arm-lift. All contraction conditions successfully resulted in active relative thickness change of LM and TrA. Relative thickness change of the LM when using the FRED was favourable in that it was lower than that observed in loaded contralateral arm raise and walking conditions, suggesting that one of the key features of specific motor control training (contraction intensity of 30-40 % MVC) has been met.

Thirdly, activity of the LM and TrA during this mode of exercise and other commonly used corrective/rehabilitative techniques based on relatively static challenges to stability was compared (gym ball, balance board). All stability challenges successfully

induced non-volitional concomitant activation of both the LM and TrA. Additionally, it was observed that the LM followed a pattern where all standing conditions elicited greater recruitment than seated conditions, with no additional effect of surface lability. Contrastingly, the TrA only demonstrated an effect of surface instability during FRED conditions. The preferential contraction ratio of the TrA in comparison to IO and EO was greatest during use of the exercise device in the standing position.

Fourthly, the intrinsic kinematic stability of the lumbopelvic region whilst using the exercise device was examined, revealing further evidence of the underlying mechanisms facilitating LM and TrA contraction. Key differences between FRED exercise and overground walking included reduced axial rotation of the trunk with respect to the pelvis (i.e. increased lumbopelvic stability) and a more anteriorly tilted pelvis. FRED exercise potentially moved the pelvis into a more advantageous position for the recruitment of TrA and LM. However, the unstable base of support afforded by FRED exercise would seem to add a challenge to movement control that could result in greater TrA and LM activity than overground walking.

Finally, the pattern of global muscle activation during this exercise was examined, and provided evidence as to the tonic nature of FRED mediated muscle activity of the lumbar paraspinal and anterolateral abdominal muscles. FRED exercise a) promoted more tonic activity of the lumbopelvic musculature compared to overground walking, b) resulted in greater spinal extensor activity than spinal flexor muscles compared with overground walking, and c) resulted in greater knee extensor activity compared with overground walking.

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LIST OF ABBREVIATIONS

ADIM	abdominal drawing-in manoeuvre
aLBP	acute low back pain
ANOVA	analysis of variance
ASLR	active straight leg raise
BF	biceps femoris
BMI	body mass index
CAL	contralateral arm lift
cLBP	chronic low back pain
CSA	cross-section area
CV	coefficient of variation
dLM	deep fibres of the lumbar multifidus
EMG	electromyography
EO	external oblique
FRED	Functional Re-adaptive Exercise Device
ICC	intraclass correlation coefficient
intEMG	intramuscular electromyography
IO	internal oblique
LBP	low back pain
LCAL	loaded contralateral arm lift
LES	lumbar erector spinae
LM	lumbar multifidus
MCE	motor control exercise
MCT	motor control training
MDC	minimum detectable change
MG	medial gastrocnemius
nsLBP	non-specific low back pain
OW	overground walking
PCSA	physiological cross-section area
RA	rectus abdominis
RCT	randomised controlled trial
RM ANOVA	analysis of variance with repeated measures
RMS	root mean square
RUSI	rehabilitative ultrasound imaging
SD	standard deviation
SEM	standard error of measurement
sEMG	surface electromyography
sLM	superficial fascicles of the lumbar multifidus
sLBP	subacute low back pain
SPSS	statistical package for the social sciences
ST	subcutaneous tissue
TA	tibialis anterior
TrA	transversus abdominis
ULM	upper limb movement
USI	ultrasound imaging
VM	vastus medialis

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AUTHOR DECLARATION

I declare that the work contained in this thesis has not been submitted for any other award and that it is all my own work.

Name.....

Signature.....

Date.....

CHAPTER I
-
Introduction

1.1 Topic Overview

With a lifetime prevalence estimate reported as high as 84 % (Walker, 2000), and associated economic burdens (including primary and secondary healthcare costs) of over £1.6 billion (Maniadakis & Gray, 2000; Savigny et al., 2009) and $28 billion (Dagenais, Caro, & Haldeman, 2008) per annum in the United Kingdom and United States, respectively, low back pain (LBP) is a common and costly global problem. With further consideration of the 20 % increase in the retail price index in the 17 year period from 1998 to 2015 (Amankwah et al., 2015) these costs are now likely to be considerably greater in the United Kingdom alone.

The natural history and prognosis following an acute episode of LBP is generally favourable with the majority of individuals recovering within six weeks (Koes et al., 2001). However, recurrence within the subsequent 12 months can be as high as 73 % (Pengel et al., 2003). In some cases, the symptom of pain can be attributed to a specific identifiable underlying pathology, however, these are in the minority and relatively uncommon (Adams, Burton, & Bogduk, 2013). For example, compression fractures and spondylolisthesis account for only 4 % and 3 % of cases, respectively, whilst ankylosing spondylitis or infection represent even fewer cases, with 0.3 % and 0.01 %, respectively (Deyo, Rainville, & Kent, 1992). As such, it is commonly estimated that a diagnosis based on exclusion of specific pathology, termed non-specific low back pain (nsLBP), accounts for 85-90 % of all patients with LBP (Airaksinen et al., 2006; Deyo, Rainville, & Kent, 1992; Koes, Van Tulder, & Thomas, 2006; Wand & O'Connell, 2008).

Whilst it is not representative of every individual case, acute, recurrent and chronic (persistence > 12 weeks) LBP are increasingly associated with changes in both the function and morphology of the local (deep) muscles of the trunk, including both the *lumbar multifidus* (LM) and the *transversus abdominis* (TrA). Such observations include atrophy of the LM at multiple vertebral levels (Danneels et al., 2000; Hides et al., 1994), attenuated activity of the LM (Kiesel et al., 2007b; MacDonald, Moseley, & Hodges, 2010; Sihvonen et al., 1997) and TrA (Ferreira, Ferreira, & Hodges, 2004), delayed activity of the LM (MacDonald, Moseley, & Hodges, 2009) and TrA (Hodges & Richardson, 1996), and a shift from tonic to phasic activation of the TrA (Saunders, Coppieters, & Hodges, 2004).

One of the key roles of these muscles is to provide stability at a segmental level to the lumbar spine (Bergmark, 1989; Cholewicki & McGill, 1996; Hodges, Cholewicki, & Van Dieën, 2013; Panjabi, 1992a; Vera-Garcia et al., 2007). In fact, due to their anatomical positioning, morphology and function, the deeper fibres of the LM and the TrA are considered crucial for local stability of the lumbar spine (Hodges, 1999; Hodges & Richardson, 1996; Kim et al., 2007). This is a fundamental requirement, given that the thoracolumbar and lumbar spine, devoid of any musculature, will experience structural failure under compressive loadings as small as 20 and 90 N in magnitude, respectively (Crisco et al., 1992). Considering spinal loadings experienced *in vivo* can range from 6 kN during selected everyday tasks (McGill & Norman, 1986) to in excess of 36 kN during competitive powerlifting (Cholewicki, McGill, & Norman, 1991), the human vertebral column is intrinsically incapable of meeting the physiological demands placed upon it without such stabilisation at a segmental level (Panjabi et al., 1989).

In light of such observations, one particular strategy, termed ‘specific stabilisation exercise’ or ‘specific motor control training’ has been developed with the aim of restoring the proper function of the LM and TrA (Richardson & Jull, 1995). Specific stabilisation exercise is founded upon five fundamental features, which include 1) sustained isometric contractions (around 10 seconds) at a low level of patient specific maximal voluntary contraction (MVC) (approximately 30-40 %), 2) co-contraction of the LM and TrA, 3) preferential recruitment of local rather than global muscles, 4) progressive increases in contraction volume and 5) progressive introduction of functional body postures and activities with increases in external load (Hodges, Cholewicki, & Van Dieën, 2013; Richardson & Jull, 1995). Whilst this mode of training is consistent with many of the proposed ‘needs’ for proper capacity, control and function of the stabilising elements of the lumbar spine, it has been found to be no more effective than general exercise strategies (Ferreira et al., 2006; Hauggaard & Persson, 2007; Macedo et al., 2009; May & Johnson, 2008; Smith, Littlewood, & May, 2014).

Despite propositions that dynamic functional activities should be incorporated into specific stabilisation programmes fully (Hodges & Cholewicki, 2007; Richardson & Hides, 2004) they often are not. Although this motor control training protocol does include incorporation of LM and TrA activation into “dynamic functional movements of the trunk” (Richardson & Jull, 1995, pp. 5) the implementation of this in published studies typically involves either relatively simple body positions such as standing on a wobble board and sitting on a gym ball (Costa et al., 2009). Additionally, this aspect is also typically attempted within a relatively short period of time such as the last three weeks of a 10 week treatment programme (Koumantakis, Watson, & Oldham, 2005; Richardson et al., 1999).

A novel mode of exercise has previously been described that purports to specifically target the local muscles during cyclical lower limb movement in weight-bearing (Korfmacher, Debuse, & Pinotti, 2006). The kinematics of the exercise is similar to that performed on an elliptical trainer where the feet follow an anti-phase quasi-elliptical path; however, it offers negligible external resistance to motion. As such, in order for the user to achieve a smooth motion, the frequency of motion must be kept low. Additionally, the absence of external resistance, results in the need for much greater motor control of the legs and pelvis for balance, than in conventional exercise devices. As one foot moves downwards through the front of the movement cycle, the muscles in the user's rear leg have to work eccentrically in order to maintain a smooth and controlled motion of the lower limbs. The resultant effect is hypothesised to be a specific, automatic, low-level isometric co-contraction of the LM and TrA during a functionally relevant movement.

The exercise modality was recently examined by Debuse and colleagues (2013) and it was shown that a single exposure to the Functional Re-adaptive Exercise Device (FRED), in either sitting or standing can induce non-volitional co-contraction of the LM and TrA in a group of healthy individuals. Since then the authors have also recommended this modality for use as a complementary exercise therapy following long-term bed rest (Evetts et al., 2014), which is an experimental protocol commonly used as an analogue to microgravity exposure (Hides et al., 2007a), and also known to induce many of the dysfunctions commonly observed in people with LBP (Belavý et al., 2011; Belavý et al., 2007).

1.2 Aims of Thesis and Research Questions

The aim of this thesis is to develop a greater understanding of the effects this newly developed exercise device, in the context of the potential relevance to LBP. Specifically, to evaluate the effectiveness of the exercise device in the recruitment of LM and TrA due to their dysfunction being identified to be either a risk factor for recurrence or a causal factor, in LBP. In order to do so this thesis aimed to address a number of fundamental research questions:

1) What are the intra- and interday reliability and precision of measurement of absolute and relative linear thickness changes of the LM and TrA muscles during functional re-adaptive exercise in a group of healthy participants? In addition, how does this compare to common alternative assessment techniques such as upper and lower limb perturbations (contralateral arm raises, active straight leg raises, walking), and volitional recruitment (the abdominal drawing-in manoeuvre)?

2) What is to be considered typical functioning of the LM and TrA during FRED exercise, as measured by absolute and relative linear thickness changes? Additionally, how does this compare to common alternative assessment techniques such as weight-bearing and non-weight bearing upper and lower limb perturbations (contralateral arm raises, active straight leg raises, walking), and volitional recruitment (the abdominal drawing-in manoeuvre)?

3) How does the typical functioning of the TrA, and internal and *external obliques* during FRED exercise compare with a range of commonly used static

challenges to upright lumbopelvic stability in both sitting and standing positions?

4) With a particular focus on lumbopelvic stability kinematics, how does FRED exercise compare to walking, arguably the most common dynamic challenge to postural stability experienced in daily living. In addition, can this comparison highlight further insights into the underlying mechanisms of action concerning LM and TrA recruitment?

5) What is the global distribution of muscle activity of the superficial trunk and lower limb musculature during FRED exercise in comparison to that seen during level walking?

These research questions were addressed using a combined total of 43 individuals participated in a total of five experimental studies designed to address each of the above research objectives. The design of these studies was such that the five experiments were achieved from only three independent sets of data collection exercises. As such, the 43 participants gave an independent total equivalent to that of 73 participants (Figure 1-1).

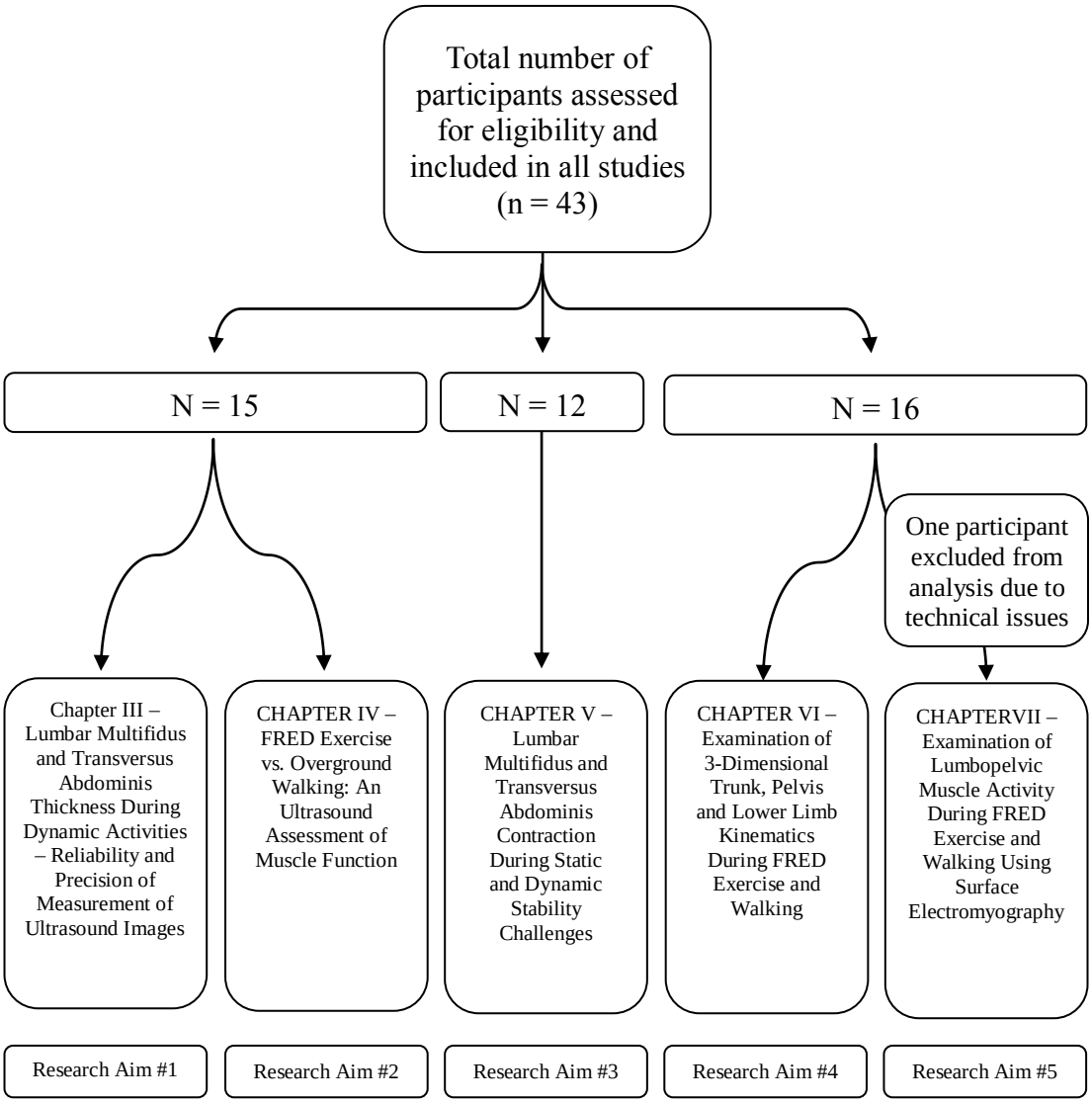

Figure 1-1 Overview of total participant utilisation. Highlighting the contribution of two single groups of participants to two separate experimental chapters.

CHAPTER II
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Literature Review

2.1 Literature Review Organisation & Search Strategy

This chapter presents a review of the literature that examines the complex structures and functions of the osteoligamentous and musculotendinous systems of the human spine and outlines the varying contributions of the sub-components within each group to the overall control of spinal motions. Following this, the scale of the burden of low back pain is highlighted along with a number of potential causal mechanisms of acute, recurrent and chronic low back pain. Specifically, mechanisms relating to the proposed optimum function of the deep muscular subsystem and observations of dysfunction and maladaptive motor control responses seen within low back pain populations are discussed. The final aspect of this review addresses the current state of knowledge concerning the effectiveness of specific motor control training, and highlights the need for a potential amendment to the current protocol by way of a standardised exercise modality to enhance the functional progression strategy contained within.

Although this was not a systematic literature review, articles of interest were identified using a general key word search strategy in several online searchable databases including PubMed, Google Scholar, and Medline. Major key words included low back pain, spinal stability, rehabilitation, lumbar multifidus, transversus abdominis, motor control training and exercise. Additional literature was subsequently identified from the reference lists of literature identified in the primary search.

2.2 Structural and Functional Anatomy of the Spine

The human spine contributes approximately 40 % of a mature individual's erect standing height and is a composite structure consisting of the functionally

interdependent passive osteoligamentous subsystem (bony elements of the vertebral column, intervertebral discs and interconnecting ligaments), the active musculotendinous subsystem (muscles and tendons surrounding the vertebral column) and the neural control subsystem (intrinsic force and motion transducers) (Panjabi, 1992a).

2.2.1 The Osteoligamentous Subsystem

As viewed in the frontal plane, a healthy vertebral column appears vertical, whereas when viewed in the sagittal plane a series of normal curves are observed distinguishing individual regions from one another. Each of these curves is characterised as being either lordotic or kyphotic, with lordosis defined as an anterior convex curve, and kyphosis as an anterior concave curve (Bernhardt & Bridwell, 1989). The individual vertebrae are also sequentially numbered craniocaudally with reference to their specific region such that the first cervical, thoracic and lumbar vertebrae are termed C1, T1 and L1 respectively (Carrino et al., 2011).

As a whole, the vertebral column functions as a closed kinetic chain with both the head and the ground, with all of the individual functional units acting together to maintain a somewhat stable position of the sensory organs (Levangie & Norkin, 2011). The smallest functional unit of the spine is generally held to be the mobile segment (Levangie & Norkin, 2011). That is to say, any two adjacent vertebral bodies, the intervertebral disc between them, the articulating zygapophyseal (facet) joints and the surrounding ligaments (Panjabi et al., 2001).

2.2.1.1 Vertebral Body

The vertebrae share a common structural design, although regional variations exist in size and organisation, reflective of the functional characteristics of any particular region (Adams, Burton, & Bogduk, 2013). The essential component of each individual vertebra is a vertebral body with a rectangular profile in sagittal view, and a semi-columnar profile when viewed in the transverse plane (Adams, Burton, & Bogduk, 2013). The craniocaudal diameter (height) of each vertebra contributes significantly (~75 %) to the overall column length (Warwick & Williams, 1973) and, as such, provides separation of the thoracic cage and skull from the pelvis facilitating the range of motion possible (Adams, Burton, & Bogduk, 2013). In terms of typical vertebral body height, each successive vertebral body is larger than the preceding one (Masharawi et al., 2008). The effective mass of these structures (augmented by contraction of the surrounding musculature) exerts a compression force on each vertebral body, thus they are designed to withstand this axial loading by way of their internal architecture (Eswaran et al., 2006). The outer shell consists of cortical bone that, although strong, requires internal bracing from vertically and horizontally oriented trabeculae that resist axial compression (Eswaran et al., 2006).

2.2.1.2 Intervertebral Discs

Situated between the inferior surface of one vertebral body and the posterior surface of the adjacent one, with the exception of the atlas (C1) and axis (C2), lies an intervertebral disc (IVD) that collectively contribute approximately 25 % of spine length (Warwick & Williams, 1973). These fibrocartilaginous pads provide separation between consecutive vertebrae, resist compression, permit limited intersegmental motions and distribute loads evenly on the vertebral bodies (Adams & Roughley, 2006). Each IVD (Figure 2-1) consists of a nucleus pulposus surrounded

circumferentially by an annulus fibrosus and sandwiched between two cartilaginous vertebral endplates (Adams & Roughley, 2006).

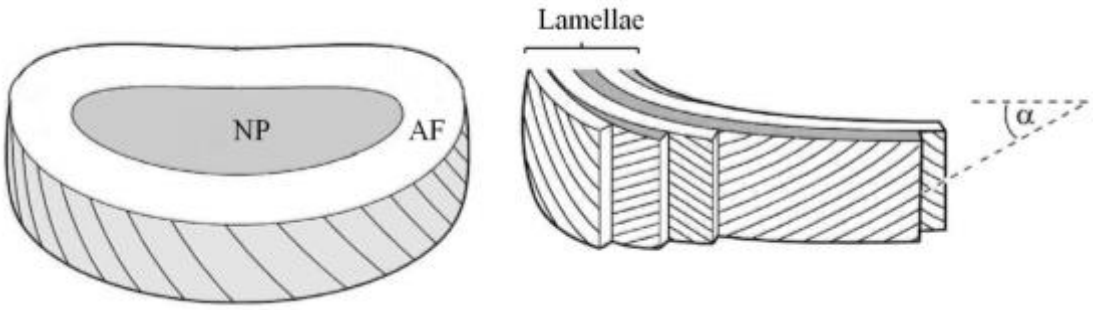

Figure 2-1. Intervertebral disc structure identifying the nucleus pulposus (NP), annulus fibrosus (AF) and highlighting the alternating oblique angle ($\alpha \approx 30°$) of the collagen fibres of the lamellae of the AF. Adapted from Adams and Roughley (2006), with permission (Appendix A).

The nucleus pulposus is a hydrated proteoglycan (predominantly aggrecan) gel bound by an irregular matrix of collagen fibres, of which approximately 80 % are type II collagen fibres (Humzah & Soames, 1988). When subjected to compressive loads the semifluid disperses radially and is simultaneously resisted by the surrounding annulus fibrosus, thus preventing it from collapsing inwards and maintaining the stiffness of the disc (Jensen, 1980).

The annulus fibrosus is the fibrous concentric ring of predominantly type I collagen fibres (~ 60 % of dry weight) organised in a 15-25 layer ply-laminate arrangement of lamellae (Coventry, Ghormley, & Kernohan, 1945). The collagen fibres of each individual lamella are aligned approximately 30° relative to the transverse plane of the disc, although in alternate directions in successive layers (Schollum, Robertson, & Broom, 2009). These alternating fibre alignments within the multi-layered annulus fibrosus are essential for conversion of the compressive forces to lateral forces in order to withstand the tensile stresses, ultimately allowing the intervertebral joints to flex,

extend and rotate in all directions whilst also resisting separation (Adams, Burton, & Bogduk, 2013). The annulus fibrosus is further subdivided into the ligamentous and capsular portions (Adams & Roughley, 2006). The outermost fibres of the annulus (ligamentous portion) attach directly to the bone at the ring apophysis whereas the innermost fibres (capsular portion) insert into the inferior and superior regions of the vertebral endplate and encapsulate the nucleus pulposus (Adams & Roughley, 2006).

The inferior and superior vertebral endplates are layers of hyaline and fibrocartilage approximately 0.5 - 1 mm, that decrease with age and proximity to the annulus fibrosus (Roberts et al., 1997) and cover the region within the ring apophysis (Ferguson & Steffen, 2003). The attachment of the vertebral endplate is stronger to the annulus than it is to the vertebral body and as such is considered an IVD component rather than a component of the vertebral body (Lotz, Fields, & Liebenberg, 2013). As with both the nucleus pulposus and annulus fibrosus, the vertebral endplates are a hydrated proteoglycan and collagen composite, although additional cartilage cells are aligned along the collagen (Roberts, Menage, & Urban, 1989).

2.2.1.3 Posterior Vertebral Elements

Projecting posteriorly from the lateral sides of the superior half of the vertebral body is a pair of sturdy bony pillars (pedicles) that support the posterior elements of the vertebra and transmit forces to the vertebra (Bogduk, 2005). Projecting medially from each pedicle is a lamina that seamlessly fuses in the centre to create the posterior element, the spinous process (Ebraheim et al., 1996). Together the pedicles and laminae form the neural arch that, in conjunction with the posterior surface of the vertebral body, encloses a channel known as the vertebral foramen through which the medulla spinalis (spinal cord) passes (Bogduk, 2005). Intermediately between each

pedicle and lamina, projecting laterally, are the transverse processes, and at the base of each transverse process is the accessory process (Adams, Burton, & Bogduk, 2013). Projecting superolaterally from the laminae are extensions of bone known as the superior articular processes (Adams, Burton, & Bogduk, 2013). On the medial surface of each superior articular process is an articular facet, covered by articular cartilage (Cavanaugh et al., 1996). On the inferolateral aspect of the laminae are further extensions that proceed inferiorly, also covered in articular cartilage, although in this instance it is the lateral surface that is covered (Cavanaugh et al., 1996).

When considered in series, the vertebral bodies and posterior elements form a three-joint complex (Kirkaldy-Willis & Farfan, 1982), where contiguous vertebral bodies articulate via the intervertebral discs and the interlocking surfaces of the superior and inferior facet joints. Each joint is numbered according to the vertebrae that are articulating at any one time, for example L4 articulates with L5 at the L4-5 intervertebral disc and the left and right L4-5 facet joints (Kirkaldy-Willis & Farfan, 1982).

The facet joints provide a bony mechanism for limiting anterior displacement of the superior relative to the inferior vertebra and axial torsion, both of which protect the intervertebral disc from excessive mechanical stresses (Figure **2-2**) (Varlotta et al., 2011). Anterolisthesis and rotation is prevented by the basic shape of the superior articular facet, be they either flat, C- or J-shaped (Varlotta et al., 2011). Notwithstanding their shape, the facets are obliquely oriented so that when rotation of the vertebra in one direction ensues, the associated lateral swing of the facet is impeded by the lateral edge of the inferior articular facet of the vertebra above (Varlotta et al., 2011). The greater medial position of the anterior edge of the superior articular facet

limits anterograde displacement in the same fashion (Bogduk, 2005). With respect to superoinferior motions, the flat surfaces of both articular processes allows for free gliding during flexion and extension, although excessive motion is limited by contact between the tips of the articular facets and the lamina below, and assisted by the surrounding and adjoining ligamentous constituents (Bogduk, 2005).

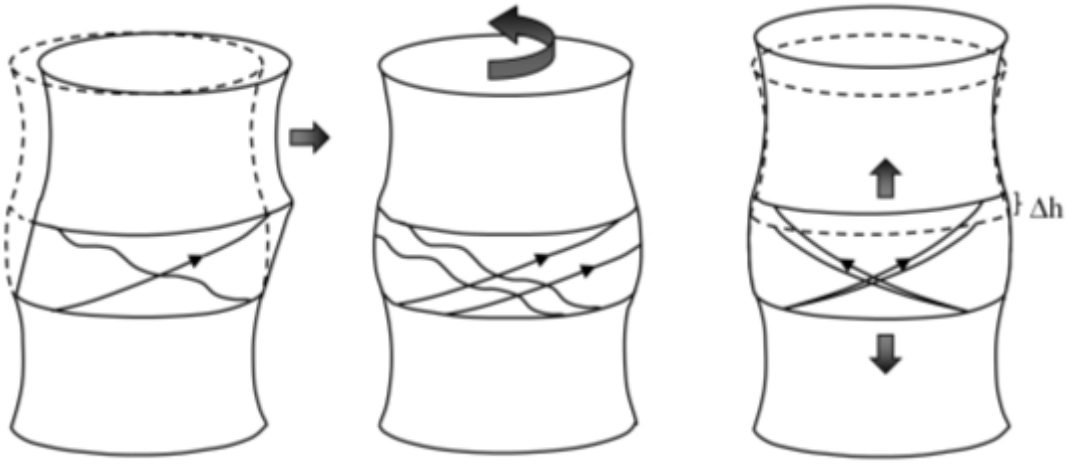

Figure 2-2. Arrangement of the collagen fibres of the annulus fibrosus resist excessive motion by tension (small arrows) in the fibres oriented in the direction of motion (large arrows) during sliding movements (left), twisting movements (middle) and separation (Δh) movements (right). Reproduced with permission (Appendix B) from Adams, Burton, Dolan and Bogduk (2006).

2.2.1.4 Paraspinal Ligaments

The paraspinal ligaments vary considerably by region, although generally six primary ligaments accompany the intervertebral and facet articulations (Adams, Burton, & Bogduk, 2013; Bogduk, 2005; Pintar et al., 1992). These are the anterior, posterior and supraspinous longitudinal ligaments, and the intersegmental ligaments consisting of the interspinous and intertransverse ligaments, and the ligamentum flavum. The three longitudinal ligaments span the entire vertebral column and connect adjacent vertebrae at either the anterior (anterior longitudinal ligament) or posterior (posterior longitudinal ligament) surfaces of the vertebral bodies and the adjacent spinous

processes (supraspinous ligament) (Pintar et al., 1992). The intersegmental ligaments connect either bilaterally to the superior and inferior laminae (ligamentum flavum), and the superior and inferior transverse processes (intertransverse ligament), or unilaterally to the spinous process of adjacent vertebrae (interspinous ligament) (Adams, Burton, & Bogduk, 2013).

The anterior longitudinal ligament spans the anterior and lateral surfaces of vertebral bodies and intervening discs from C2 through S5 (Bogduk, 2005). It consists of at least two distinct layers of thick crossed collagen fibre bundles with differing arrangements (Pintar et al., 1992). The superficial fibres are long and span several vertebral bodies, with the deep fibres anchoring segmentally whilst also blending with the ligamentous portion of the annulus fibrosus. This ligament is also particularly well developed in the lordotic regions of the vertebral column, whereas in regions of kyphosis it is much less so (Levangie & Norkin, 2011). From the lower thoracic region to L5/S1, the anterior longitudinal ligament increases in thickness and width as well as being reported to be twice as strong as the posterior longitudinal ligament (Myklebust et al., 1988). The superficial layer can therefore be thought of as a bracing system for passive maintenance of the lumbar lordosis with the deeper layer acting to limit segmental hyperextension by resisting anterior intervertebral distraction/separation (Putz, 1992).

The posterior longitudinal ligament forms the floor of the vertebral canal and extends from the body of the axis, as a continuation of the membrana tectoria to the sacrum (Levangie & Norkin, 2011). It is also a non-uniform structure, similar to the anterior longitudinal ligament where the superficial layer traverses multiple segments and the deeper fibres attach only to adjacent vertebrae and again interlace with the ligamentous portion of the annulus fibrosus and attaching to the boundary region of the vertebral

end plates (Behrsin & Briggs, 1988). The structure is regular as far as T2/T3 where the superficial layer reduces in width ultimately forming a narrow ribbon that continues down to the sacral canal (Levangie & Norkin, 2011). On account of its fibre arrangement the posterior longitudinal ligament is considered to be a simple passive brace in the cervical and thoracic regions, whereas in the lumbar region the monosegmental divergent arrangement produces tension in the final stages of all principal movement directions (Putz, 1992).

The supraspinous ligament connects the tips of the spinous processes and is developed as a separate structure from C7 to L3 or L4 (Putz, 1992). Consisting of both elastin and collagen fibres, it runs parallel to the superficial fibres of the interspinous ligament (Putz, 1992). Similarly to the interspinous ligament, the supraspinous ligament resists separation of the spinous processes during flexion and is the first to fail during hyperflexion (Adams & Hutton, 1983). It is richly endowed with mechanoreceptors and appears to be involved in the recruitment of stabilising musculature such as the multifidi (Solomonow et al., 1998).

The interspinous ligament connects the opposing superior and inferior edges of contiguous vertebrae and is a fibrous sheet of type I collagen fibres, and profuse elastin fibres (Yahia et al., 1990). The specific orientation of fibre direction is not equivocal, although a number of authors have described an oblique direction from one vertebra to the next (Fujiwara et al., 2000; Yahia et al., 1990). In addition to resisting separation during flexion it is also suggested that the interspinous ligament produces an anterior shear component during flexion (McGill, 2007).

The intertransverse ligaments pass between adjacent transverse processes bilaterally and serve to offer resistance during lateral bending (Putz, 1992). In the cervical and thoracic regions, relatively few discernible fibres are found, whereas in the lumbar region these are more membranous and form part of the thoracolumbar fascia as well as serving to separate dorsal and ventral musculature (Bogduk, 2005).

The ligamentum flavum is a short, thick bilateral ligament adjoining the adjacent vertebrae from lamina to lamina from C2 to the sacrum (Yong-Hing, Reilly, & Kirkaldy-Willis, 1976). Histologically, it is predominantly elastin (approximately 80 %) with the remainder consisting of collagen fibres (Yahia et al., 1990). As such, the ligament differs from other lumbar spinal ligaments and is suggested to aid in restoration of the lumbar spine from flexed positions and pre-stress the intervertebral disc (Nachemson & Evans, 1968). Additionally, the highly elastic nature of the ligament reduces the risk of nerve root impingement by resisting buckling during extension, a property not endowed in ligaments that are more collagenous (Panjabi & White III, 1980).

2.2.2 The Musculotendinous Subsystem

Where the osteoligamentous components endow the vertebral column with a degree of intrinsic stability, their function is predominantly to limit excessive motions (Bergmark, 1989; Panjabi & White III, 1980). The surrounding muscles provide the column with a greater degree of motion within the normal physiological ranges by way of tension development between the origin and insertion (Panjabi & White III, 1980). Ultimately, this serves as a torque generator system about any segment's respective axis of rotation (Bergmark, 1989).

Topographically, the intrinsic muscles (those that act directly on of the vertebral column rather than the upper extremities) can be subdivided by region (cervical, thoracic and lumbar) as well as by whether the specific muscle in question is superficial, intermediate or deep (Adams, Burton, & Bogduk, 2013). Additionally, a common functional classification scheme exists whereby muscles are can be categorised as either local or global depending on their primary mechanical functions (Bergmark, 1989).

Local muscles principally act to instil stability at an intersegmental level, whereas global muscles function to produce large moments for gross movement, increase intra-abdominal pressure (IAP) and distribute/balance outer loads between the thoracic cage and pelvis (Bergmark, 1989).

2.2.2.1 Local and Global Musculature

All muscles originating or inserting onto elements of the vertebrae are considered as part of the local system and include the *multifidi, interspinales, intertransversarii, rotatores,* and the medial parts of the *quadratus lumborum* (Bergmark, 1989). The notable exceptions from this classification are the *psoas major* and *latissimus dorsi* (Bergmark, 1989). Both muscles do arise from elements of the vertebrae though their functions are predominantly to produce movement of the extremities and simply use the vertebral column as a solid base of attachment from which to act (Bergmark, 1989). The *psoas,* however, can exert large compressive forces on the lumbar IVDs in the execution of hip flexion (Bogduk, Pearcy, & Hadfield, 1992). Global musculature includes the *erector spinae, obliquus internus* and *externus, rectus abdominis* and the lateral parts of the *quadratus lumborum* (Bergmark, 1989).

2.2.2.1.1 Intersegmental Local Musculature

The truly intersegmental muscles of the spine are the bilaterally paired *intertransversarii*, *interspinales* and *rotatores* (Donatelli, 2007) . Within the *intertransversarii* muscle group are three distinct muscles, *laterales ventrales*, *laterales dorsales* and *mediales* (Bogduk, 2005).

Collectively these intersegmental muscles should contribute to lateral flexion, extension and rotation of individual vertebrae. However, due to their small physiological cross-sectional areas (79 ± 6 and 126 ± 53 mm^2) and short moment arms (approximately 5 and 3 cm), the lumbar fascicles of these intersegmental muscles can generate a maximum torque contribution of less than 3 % of the total maximum torque at any lumbar level (Daggfeldt & Thorstensson, 2003). However, the concentration of muscle spindles within these muscles can be between 4.5 and 7.3 times greater than that of other muscles such as the *multifidus* (Nitz & Peck, 1986). As such, these unisegmental muscles are thought to function more as a series of position sensitive force transducers enhancing proprioceptive awareness (Adams, Burton, & Bogduk, 2013; Fritz, Erhard, & Hagen, 1998; Gilchrist, Frey, & Nadler, 2003; Hodges, 2003; McGill, 2007).

2.2.2.1.2 Multisegmental Local Musculature

The *multifidus* is the most medial of the major paraspinal muscles and the largest that spans the lumbosacral junction (Kay, 2000; Macintosh et al., 1986). Fibres of the *multifidus* are present in cervical, thoracic and lumbar regions of the spine, although the muscle is much more developed in the lumbar region (Macintosh et al., 1986). In the lumbar region, the fibres of the *multifidus* are divided into five separate bands that are centred on each of the lumbar spinous processes and radiate inferiorly to assume a

variety of insertions (Figure **2-3**). Each individual band is distinguished principally by the inferior attachment with the deepest fascicles arising from the vertebral lamina and all others from the spinous process (Macintosh et al., 1986).

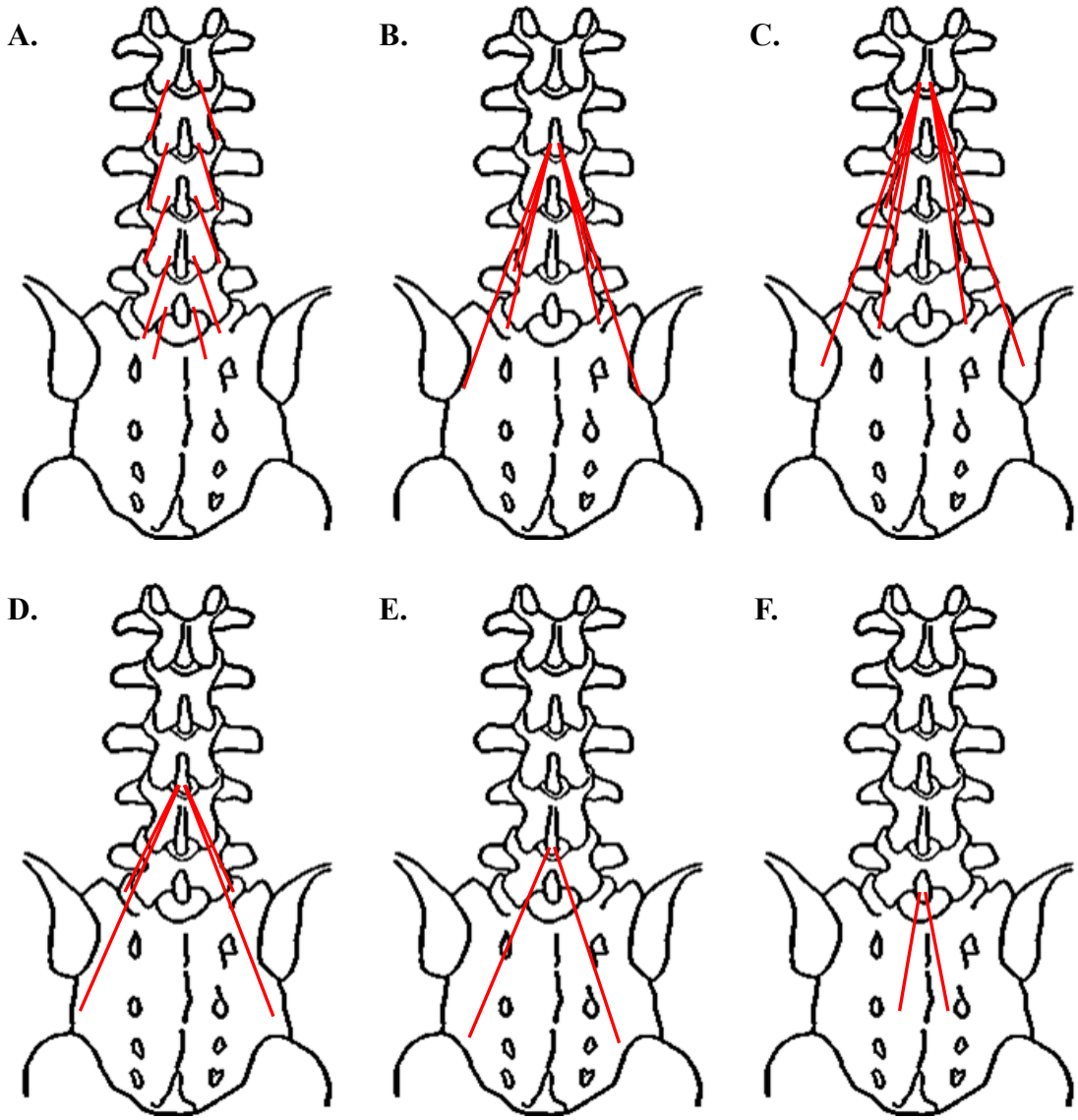

Figure 2-3. Fascicular arrangement of the lumbar segment of the *multifidus* viewed in the posterior frontal plane. A depicts the schematic illustration of the deep fibres at every vertebral level. B-F depicts the longer more superficial fibres projecting from the caudal edges of the spinous processes Redrawn from Macintosh and Bogduk (1986), with permission (Appendix C).

Although the general pattern of distal attachments varies from segment to segment, the archetypal arrangement is that demonstrated by those fibres arising from the L1

vertebra (Kay, 2000) (Figure **2-3**A). The fibres arising from the lamina of L1 insert at the mammillary process of the vertebra two levels inferiorly (L3) (Macintosh et al., 1986). Fibres arising from the lateral surfaces of the spinous process attach at the mammillary process three levels inferiorly (L4) (Macintosh et al., 1986). The fascicles from the tubercle of the spinous process insert at the mammillary process four (L5) and five (S1) vertebrae inferiorly (Macintosh et al., 1986).

Inferior to L1, this pattern is modified, as mammillary processes below S1 are absent. The fibres arising from the lamina of L2 attach to the mammillary process of L4 (Macintosh et al., 1986). The fibres of the lateral surface of the spinous process attach to the mammillary process of L5 (Macintosh et al., 1986). The three fibres of the spinous process attach to the mammillary processes of L5 and S1 plus an area surrounding the medial and inferior regions of the posterior superior iliac spine (Kay, 2000). Vertebral levels L3 and L4 share a similar arrangement, although the most superficial fascicles attach at the deep surface of the *erector spinae* aponeurosis (Figure **2-3**A, D and E). Those from the L5 level insert at the superior medial region of the sacrum projecting as far as S3 (Macintosh et al., 1986) (Figure **2-3**A and F).

By way of dissection of 12 adult cadaveric spines, Macintosh et al. (1986) also found that the LM is innervated unisegmentally. All fascicles attaching to the spinous process or lamina are innervated by the medial branch of the dorsal ramus originating inferior to the respective vertebrae. Additionally, the same innervation pattern exists for the intersegmental muscles. Therefore, the muscle responsible for torque generation preventing or decelerating unisegmental motion is innervated by the nerve of that given segment, highlighting the important role they have in stabilising the spine during dynamic movements/activities.

When the lines of action, viewed in the frontal plane, of the LM are resolved (Figure **2-4**A) the horizontal component is minor (22.45 ± 5.59 %) in comparison to the vertical (Macintosh & Bogduk, 1986). When viewed in the sagittal plane (Figure **2-4**B) the inferior attachments of the fascicles of each band tend to lie almost perpendicular to the spinous processes and posterior to the axis of sagittal rotation of the vertebra with considerable mechanical advantage (Macintosh & Bogduk, 1986). Together this suggests that the fascicles of the LM are suited to act as posterior sagittal rotators of the vertebra of origin, and with little to no posterior shear action the multifidus is unlikely to contribute to retrolisthesis (Macintosh & Bogduk, 1986).

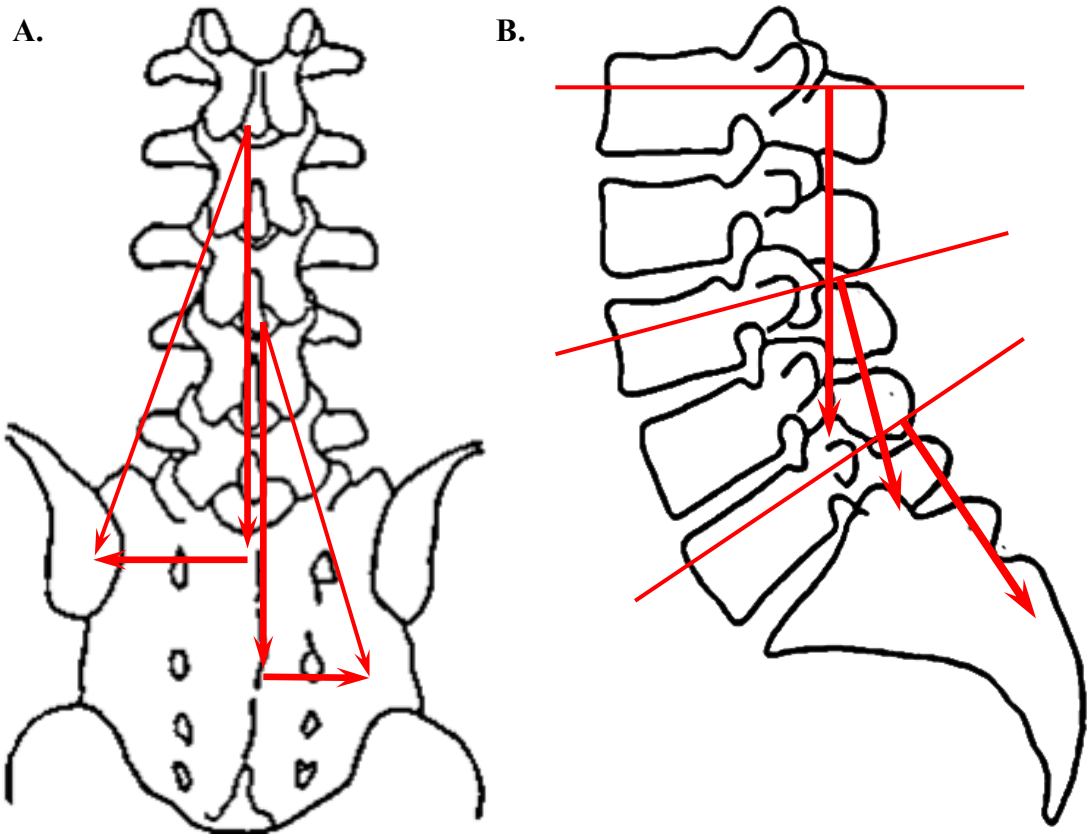

Figure 2-4. Frontal plane schematic of the resolution of the oblique fascicular orientation of the *lumbar multifidus* into both vertical and horizontal components (A). Sagittal plane schematic of the mean fascicular orientation of the *lumbar multifidus* being almost entirely perpendicular to the spinous process (B). Redrawn from Macintosh and Bogduk (1986), with permission (Appendix C).

Additionally, skeletal muscle architecture properties including the number and orientation of muscle fibres, sarcomere length, and physiological cross-sectional area also support the hypothesis that the LM is uniquely designed as a deep lumbopelvic (Ward et al., 2009). In their study of eight cadaveric lumbar spines Ward and colleagues (2009) found the *multifidus* muscle to combine both a large cross-sectional area of 23.9 ± 3.0 cm^2 with short muscle fibre lengths of 5.66 ± 0.65 cm from T12 to L5 (Figure **2-5**). Although force production was not explicitly measured within this study, the force producing capacity can be approximated to 60 N according to the assumption of 250 N/m2 as demonstrated by Powell et al. (1984). Twice that of any other lumbar extensor (Delp et al., 2001).

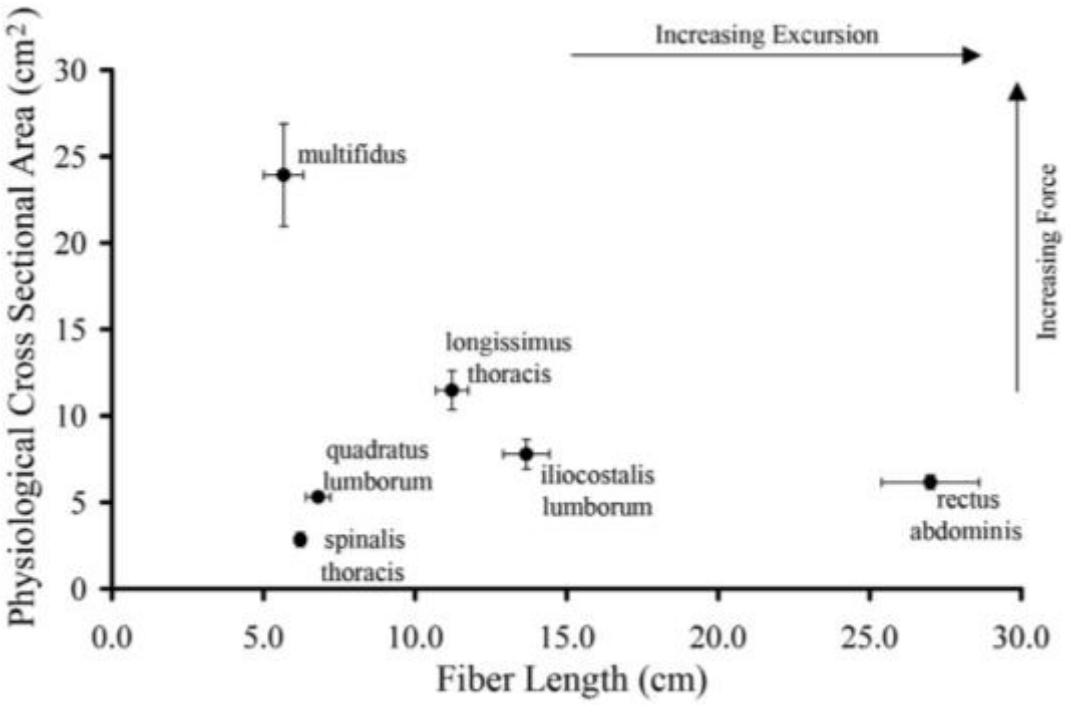

Figure 2-5. Relationship between physiological cross-sectional area and fibre length of the *lumbar multifidus* in comparison to other muscles (Ward et al., 2009), with permission (Appendix D).

The *quadratus lumborum* is lateral to the lumbar spine with attachments between the twelfth rib, ilium and the lumbar spine. In a recent detailed dissection study (Phillips,

Mercer, & Bogduk, 2008) of four embalmed cadavers the *quadratus lumborum* was found to consist of an arrangement of three multifascicular layers (Figure **2-6**) in anterior, intermediate and posterior compartments (Phillips, Mercer, & Bogduk, 2008). As defined by osseous attachment sites, each layer contains principal fascicles ascribed as iliocostal, iliothoracic, iliolumbar or lumbocostal. Those arising from the ilium do so from an attachment area starting almost opposite the L5 transverse process and extends 5-7 cm laterally along the posterior edge of the iliac crest (Phillips, Mercer, & Bogduk, 2008).

In the anterior layer, the iliocostal fibres arise from the anterior margin of the iliac attachment site and insert along an area on the lower anterior surface of the twelfth rib that extends 4.5-7 cm from the head of the rib (Phillips, Mercer, & Bogduk, 2008). Iliothoracic fibres again arise from the iliac attachment area though converge and insert onto the lateral surface of the twelfth thoracic vertebral body (Phillips, Mercer, & Bogduk, 2008). In some instances, these were accompanied by fibres arising from the L4 and L5 transverse processes. The intermediate layer consists of a number of lumbocostal fascicles distinguished by its radiate arrangement, arising from the tips of the L4 and L3 transverse processes and inserting on the costal attachment area, behind the anterior layer (Phillips, Mercer, & Bogduk, 2008). The posterior layer consisted of both iliocostal and iliolumbar fascicles, with iliocostal fibres arising from the lateral third of the iliac attachment area and attaching to the full costal attachment area. Iliolumbar fibres arise from all thirds of the iliac attachment area and inserted onto the tips of the upper lumbar vertebrae L2 and L3 most often, and on some occasions L1 and L4 (Phillips, Mercer, & Bogduk, 2008).

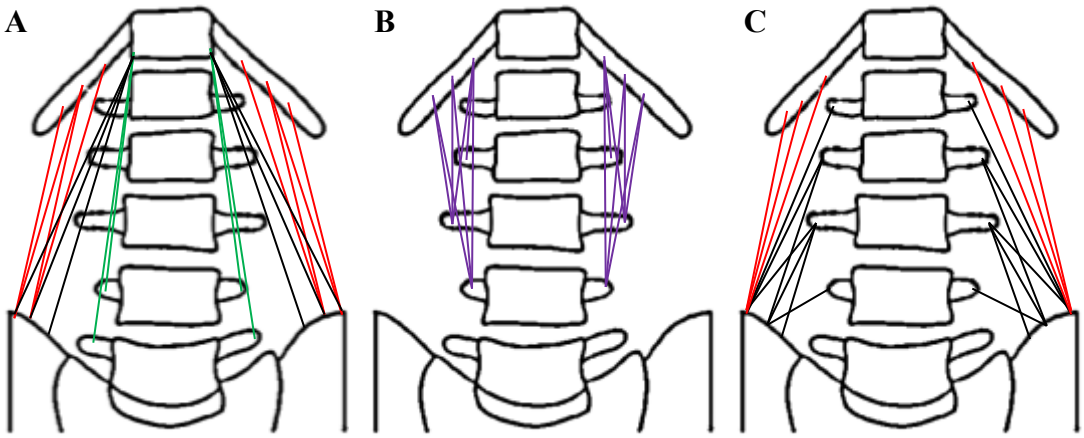

Figure 2-6. Multifascicular arrangement of the anterior (A), intermediate (B) and posterior (C) compartments of the *quadratus lumborum* viewed in the anterior frontal plane anterior intermediate posterior. Red, black purple and green lines indicate the general grouping of fibres as being iliocostal, iliothoracic, lumbocostal or iliolumbar respectively. Redrawn from Phillips et al. (2008), with permission (Appendix E).

As a result of the costal attachments, many accord it a functional role in respiration, and as a result of its vertebral attachments, also accord it a function in lateral flexion of the lumbar vertebrae (Norris, 2008). Additionally the *quadratus lumborum* also possesses a rather small extensor moment arm of approximately 20 mm (McGill, Santaguida, & Stevens, 1993) suggesting a contribution also to extension. Using the measured PCSA of the individual fascicles and a force equivalent of 46 N/cm^2, Phillips and colleagues (2008) determined the relative segmental torque contributions of the *quadratus lumborum* in extension and lateral flexion. In both instances, combined magnitudes of less than 10 Nm per segment were calculated. Given the extensor moments of the *erector spinae* and *multifidus* contribute between 100 and 150 Nm per segment (Bogduk, Macintosh, & Pearcy, 1992), respectively, the *quadratus lumborum* must be regarded as a limited contributor to extension. Although no studies to date have modelled the contribution of all muscles to lateral flexion, it has been established that mean lateral flexion torque can be approximately 103 Nm (McNeill et al., 1980), with the contribution of the *quadratus lumborum* to this again accounting for around

10 % (Phillips, Mercer, & Bogduk, 2008). Thus, the *quadratus lumborum* appears to have no more than a limited contributory action on the lumbar spine.

2.2.2.1.3 Multisegmental Global Musculature

The *erector spinae* is a long, substantial muscle group lying lateral to the *multifidus* spanning from the sacrum and ilium to the posterior thoracic region (Macintosh & Bogduk, 1987) and consists of the *iliocostalis lumborum* and *longissimus thoracis*, each with distinct lumbar and thoracic elements.

The *iliocostalis lumborum* is composed of eight or nine fascicles arising from the angles of the lower eight or nine costals (*iliocostalis lumborum pars thoracis*) that insert onto the ilium above and medial to the posterior superior iliac spine (Macintosh & Bogduk, 1987). Each fascicle is of uniform length measuring between 10-13 cm with the rostral tendons converging and forming the lateral part of the *erector spinae* aponeurosis (Delp et al., 2001; Macintosh & Bogduk, 1987). The *iliocostalis lumborum pars thoracis* are arranged such that they extend the thorax on the pelvis or resist forward flexion (Macintosh & Bogduk, 1987). They have no direct action on individual lumber vertebrae, although effectively act to exert a bowstring effect on the lumbar spine helping to maintain the lumbar lordosis (Adams, Burton, & Bogduk, 2013; Delp et al., 2001). Deep to the *iliocostalis lumborum pars thoracis* is the *iliocostalis lumborum pars lumborum* whose fibres arise from the lateral quarters of the L1-L4 transverse processes and the adjacent middle layer of the thoracolumbar fascia in systematic order (Macintosh & Bogduk, 1987). Inferiorly each lumbar fascicle has a direct insertion onto the iliac crest in a laminated structure – those fascicles from L1 cover those from L2, and so on. Similar to *iliocostalis lumborum pars thoracis*, the fibres of *iliocostalis lumborum pars lumborum* also serve to bring

about extension and resist flexion, although they do so to each individual vertebra to which they attach (Macintosh & Bogduk, 1987).

The *longissimus thoracis* consists of two sets of fascicles arising from the transverse processes of all thoracic vertebrae and the ribs only below T3 (Aspden, 1992). Irrespective of fascicular origin they span the lumbar region side by side and form the medial part of the *erector spinae* aponeurosis, covering the *longissimus thoracis pars lumborum* and the *multifidus* (Aspden, 1992). Caudally, the individual tendons insert systematically on to the sacral spinous processes, across the lower end of the sacrum and the posterior segment of the iliac crest (Adams, Burton, & Bogduk, 2013; Macintosh & Bogduk, 1987). As with the *iliocostalis lumborum pars thoracis*, the *longissimus thoracis pars thoracis* has no direct action on individual lumber vertebrae, although effectively act to exert a bowstring effect on the lumbar spine helping maintain the lumbar lordosis (Adams, Burton, & Bogduk, 2013). The *longissimus thoracis pars lumborum* arises from the tips of the accessory processes of L1 through L4 and insert onto the ilium superior and medial to the posterior superior iliac spine (Macintosh & Bogduk, 1987). Together these fibres act to extend the individual vertebrae to which they attach and are, therefore, able to extend and resist flexion (in bilateral contraction) as well as contribute to lateral flexion (in unilateral contraction) of the lumbar spine (Delp et al., 2001; Macintosh & Bogduk, 1987). However this action is less efficient than that of the *multifidus*, which acts on the longer lever arms afforded by the spinous processes (Bogduk, 2005; Macintosh & Bogduk, 1987; Macintosh et al., 1986)

2.2.2.1.4 Abdominal Wall Musculature

With fibres arising from the anterior surfaces of the costal cartilages of costals 5-7 and inserting onto the pubic crest and symphysis, the *rectus abdominis* forms a broad bilateral strap either side of the midline of the abdomen. Oriented almost entirely vertically (85.3 ± 2.7 °) with a relatively large PCSA (7.41 ± 1.91 cm^2) and the longest moment arm of all trunk muscles (Dumas et al., 1991; Guzik et al., 1996) it is a prime contributor to flexion of the torso.

Lateral to the rectus abdominis lies the EO that arises from the external surfaces of the lower eight costals and crosses the abdomen inferomedially forming an aponeurosis near to the border of the rectus abdominis and adjoining that of the contralateral muscle at the linea alba (Adams, Burton, & Bogduk, 2013). Fibres from the lowest ribs descend almost vertically downward and insert onto the anterior half of the outer lip of the iliac crest (Arslan, 2005). As a result of the inwardly oblique bilateral structure, the mechanical role is a global one of either thoracic flexion with bilateral contraction or combined thoracic flexion-rotation with unilateral contraction (Bergmark, 1989).

Deep to the EO the fibres of the IO arise from the lateral raphe of the thoracolumbar fascia, iliac crest and lateral two thirds of the inguinal ligament (Bogduk & Macintosh, 1984) and pass superomedially, perpendicular to that of EO. The posterior fibres arising from the iliac crest insert onto the inferior surface of the lower three or four costals. Middle fibres project towards the midline where they become aponeurotic and divide into two layers covering the *rectus abdominis* anteriorly and posteriorly before inserting onto the linea alba (Arslan, 2005). The lowest fibres arch inferiorly and insert onto the pubic bone (Adams, Burton, & Bogduk, 2013). Acting on the thorax in a

similar fashion to EO the IO induces flexion with bilateral contraction and flexion-lateral bending with unilateral contraction (Bergmark, 1989).

Deepest of all anterolateral abdominal wall muscles is the *transversus abdominis* that has a variety of regional origins from the inner surfaces of the lower six costals, the lateral raphe of the thoracolumbar fascia and the inner lip of the iliac crest (Askar, 1977; Bogduk & Macintosh, 1984; Warwick & Williams, 1973). The muscle crosses the abdomen with an almost horizontal fibre orientation, fuses medially with the aponeuroses of *obliquus externus* and *internus* before terminating at either the linea alba, xiphoid process of the sternum or the pubic crest. Those fibres arising from the thoracolumbar fascia do so most commonly from a band approximately 1-4 cm superior to the iliac crest, an attachment site also corresponding to the adjoining of fibres from the posterior layer of thoracolumbar fascia that attaches to the L3 spinous process (Bogduk & Macintosh, 1984). Due to the horizontal fibre orientation of the *transversus abdominis*, contraction results in a reduction of the circumference of the abdominal canister with a concomitant increase in intra-abdominal pressure (Hodges, 1999).

Whilst it is evident that each of these muscles can contribute to spinal control (repositioning of individual vertebrae to control the displacement and velocity displacement of intervertebral joints including IVDs), the relative contributions of each to achieving this is dependent upon activity, posture and load (Hodges, McGill, & Hides, 2013). Compromised function in any of the muscles contributing to control of spinal motion or spinal stiffness results in increased load and increased risk of back injury (Hodges, McGill, & Hides, 2013).

2.3 Musculoskeletal Conditions and Back Pain

Musculoskeletal disorders and diseases are a diverse heterogeneous grouping in respect of the underlying pathophysiology (Woolf & Pfleger, 2003) and collectively form the most common causes of physical disability, affecting hundreds of millions of people globally (Lidgren, 2003; Weinstein, 2000). In the UK for example, the 1995 General Household Survey sampled 18,087 individuals aged 16 and over from 11,914 households by way of interview between April 1995 to March 1996 (Rowlands et al., 1997). It was found that musculoskeletal problems were the most common at all ages, with a rate of 143 and 159 per 1000 adult men and women, respectively, and a pooled gender rate of 152 per 1000. Similarly, in the corresponding survey of Australia from February 1995 to January 1996 of the 54,000 respondents, the reported pooled prevalence of musculoskeletal conditions was 130 per 1000 (Skinner, 1997). Additionally, in the most recent Global Burden of Diseases, Injuries, and Risk Factors Study, the largest contributor to the burden of years lived with disability (YLDs) in the United Kingdom was that of musculoskeletal disorders accounting for almost one-third (30·5 % [25·5 – 35·7 %]) of all YLDs (Murray et al., 2013).

Given the many functional demands and the innate complexity of the human spine it may not be surprising that back pain features heavily within these burden of disease statistics for musculoskeletal conditions. When these conditions are further categorised by anatomical site, pain in the lower back is often cited as the most common (Urwin et al., 1998). Urwin and colleagues (1998) for example, surveyed 5752 adult men and women in the Greater Manchester area of the UK and found that of those reporting musculoskeletal pain lasting longer than one week in the previous

month, 23 % identified the lower back as the source followed by the knee at 19 % and shoulder at 16 %.

2.3.1 Low Back Pain Prevalence

Whilst the primary method of data collection in epidemiological enquiry is restricted to questionnaires by necessity, this can lead to difficulty in comparing separate studies due to the precise wording of questions influencing the responses (Choi & Pak, 2005). Additionally estimating the true prevalence of back pain can also be problematic due to variances between studies in definitions of the severity of the problem, to what exactly represents an 'episode' and even the descriptions of the low back itself (Walker, 2000). Because of such differences, it is not surprising that the available literature offers many differing findings regarding the prevalence of LBP in the general population.

Some comprehensive reviews, however, have attempted to find some degree of consensus. Since 2000, two large scale reviews have been published (Hoy et al., 2012a; Walker, 2000). Walker (2000) reviewed all English language research papers published in the previous 32 years (between 1966 and 1998). Of the 56 studies identified, representing approximately 330,000 individuals from 23 countries, 30 were identified as being methodologically acceptable according to the criteria proposed by Leboeuf-Yde and Lauritsen (1995). Of the remaining studies, only those including both genders and specifying the anatomical area as being the low back were included. Thus, six studies of point prevalence (proportion of individuals affected at any given time), eight studies of yearly prevalence (proportion of individuals affected over a 12-month period) and twelve studies of lifetime prevalence (proportion of individuals affected during their lifetime) were included in the analysis (Figure **2-7**). Point

prevalence ranged from 12 % to 33 % with a weighted mean of 16.8 % from a total sample size 12,299. One-year prevalence ranged from 22 % to 65 % with a weighted mean of 40.11 % from a total sample size of 15,909. In addition, lifetime prevalence ranged from 11 % to 84 % with a weighted mean of 59.18 % from a total sample size of 27,386. Due the methodological heterogeneity and population variances across the studies included, however, pooled data should be treated with some degree of caution.

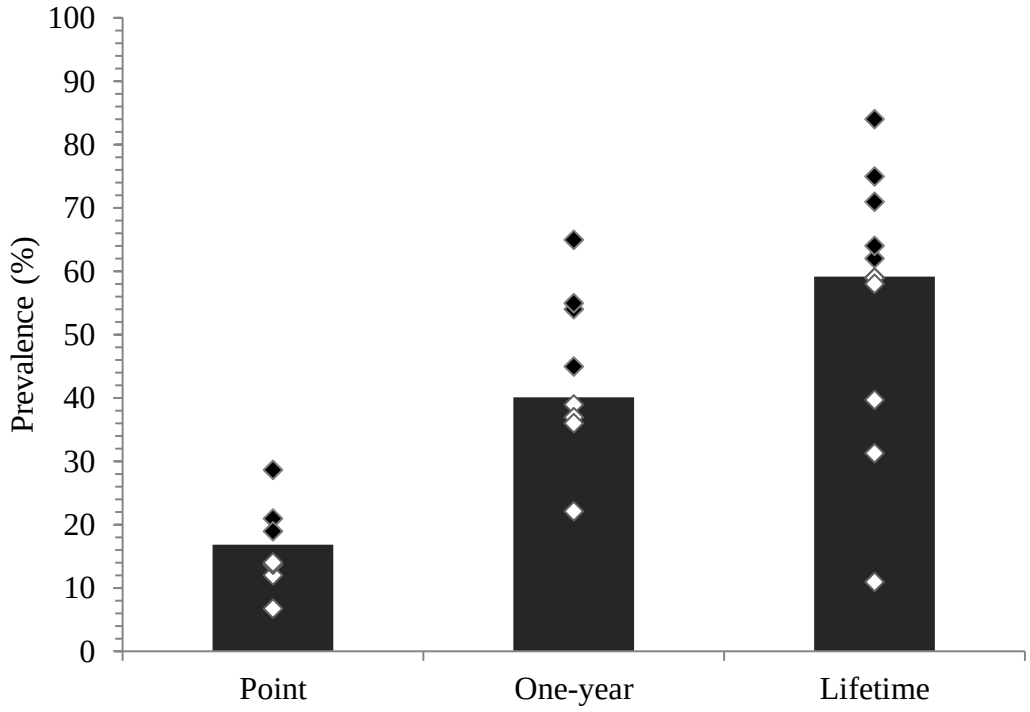

Figure 2-7. Point, one-year year and lifetime prevalence of individuals reporting low back pain. Bars represent pooled mean average prevalence (weighted by study sample size) with diamonds denoting the reported prevalence of each individual study. Data from Walker (2000).

In the most recent of these systematic reviews (Hoy et al., 2012a) it was found that a substantial number of studies had been published in the interim period that were subsequently included in the new analysis. Additionally, the authors also stratified the pooling of data according to an assessment of the risk of bias (Hoy et al., 2012b). It was found that with the additional available data and the exclusion of studies with a

high risk of bias that point and one-year prevalence was comparable at 18.3 ± 11.7 % and 38.0 ± 19.4 %, respectively. However, lifetime prevalence estimates decreased significantly (38.9 ± 24.3 %).

Additionally, with associated economic burdens (including primary and secondary healthcare costs) of over £1.6 billion (Maniadakis & Gray, 2000) and $28 billion (Dagenais, Caro, & Haldeman, 2008) per annum in the United Kingdom and United States, respectively, low back pain (LBP) is a common and costly global problem. And with further consideration of the 20 % increase in the retail price index in the 17 year period from 1998 to 2015 (Amankwah et al., 2015) these costs are now likely to be considerably greater in the United Kingdom alone.

2.3.2 The Natural History and Time Course of Low Back Pain

Based on their review of the epidemiology of LBP, Wood and Badley (1980) proposed a simple taxonomy classifying LBP as either the transient twinges experienced by the majority, acute episode of pain experienced by many or persistent back pain and disability affecting the minority. This basic taxonomy continues today and LBP is often described as acute, subacute or chronic based on the recent temporal presentation of pain. Acute LBP (aLBP) is typically defined as ≤ 6 weeks in duration, subacute LBP (sLBP) as 6 - 12 weeks and chronic LBP (cLBP) as a current duration of symptoms greater than 12 weeks (Koes et al., 2010).

The natural history (development without clinical intervention) and prognosis of an initial episode of LBP is commonly purported to be excellent with 90 % of people with LBP recovering within 4-6 weeks (Dixon, 1973; Koes et al., 2001; Pengel et al., 2003). However, as with estimates of prevalence, the data available are limited by a large

degree of variability between study designs, and particularly in the choices of outcome measures used such as pain, disability, return to work and/or GP consultations (Hestbaek, Leboeuf-Yde, & Manniche, 2003).

Following a retrospective audit of 940 consecutive cases of LBP related work absence registered with the Public Health Insurance Office in Göteborg, Anderson et al. (1983) reported a negative exponential decrease in the rate of return to work as a function of the duration of current absence (Figure **2-8**). Thus, a return to work was predicted within 12-days in 50 % of cases, 75 % within 30-days and 90 % within 55-days. Furthermore, Waddell (1987) is often cited as purporting that 80-90 % of LBP episodes 'recover' within six weeks, irrespective of the administration of treatment (Bakker et al., 2007; Elfering & Mannion, 2008; Pengel, Maher, & Refshauge, 2002). Although, as in Anderson et al. (1983), this estimate refers not to the cessation of pain or disability but to return to work.

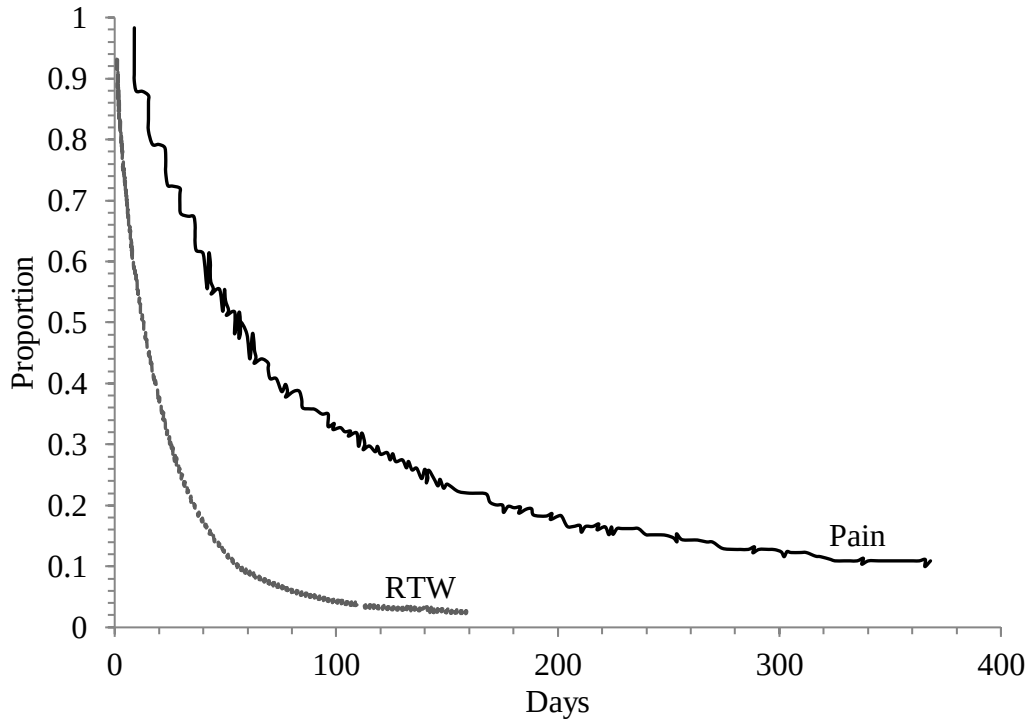

Figure 2-8. Kaplan-Meier curve of the proportion of individuals either returning to work (RTW) or reporting cessation of pain (Pain) as a function of low back pain episode duration. Data from Andersson et al. (1983) and van den Hoogen et al. (1998).

As part of a prospective study, 443 consecutive individuals consulting their general practitioner were followed-up with monthly questionnaires that included a visual analogue scale for ratings of pain severity for 12 months. Participants were considered to have recovered following four consecutive pain free weeks. Similar to the data presented by Anderson et al. (1983), a negative exponential decrease in the rate of pain cessation was noted (Figure **2-8**), however the time course to recovery was significantly longer. After four weeks, only 30 % of patients were free from pain, after eight weeks this was 52 % after 12 weeks 65 % were free from pain. At the end of the follow-up period (12 months), 10 % of patients had still not recovered. Although return to work and cessation of pain offer useful insights into the natural history of LBP, clearly the two should not be considered interchangeable (Hestbaek, Leboeuf-Yde, & Manniche, 2003).

More recently, a systematic review by Itz et al. (2013) examined the clinical course of pain in patients with non-specific acute low back pain in primary care settings using only articles published from 1990-2010. In this review, that includes data from 11 different studies across six countries (United Kingdom, Germany, The Netherlands, Denmark, Australia and the United States) with a total sample size of 3118, the prognosis is not so promising. The findings from this study were that spontaneous recovery from non-specific LBP occurs in the first three months in 67 % (95% CI: 50–83%) of individuals, however the majority of patients (65 % [95% CI: 54–75 %]) still experience pain one-year after onset of LBP. Thus indicating that the assumption of spontaneous recovery in a large majority of individuals quite often purported in clinical management guidelines (Dagenais, Tricco, & Haldeman, 2010; Van Tulder et al., 2006) is not justified.

2.3.3 Recurrence and Chronicity of Low Back Pain

Following any initial resolution of an episode of LBP, there is a tendency for many patients to experience a recurrent episode within the following 12 months (Faas et al., 1993; Hides, Jull, & Richardson, 2001). Of 162 patients receiving no active treatment for LBP, Faas and colleagues (1993) reported that 66 % of individuals would experience at least one recurrent episode within 12 months of initial consultation. In a similar study, though with a much smaller sample size (n=19), Hides et al. (2001) reported that 84 % of control group patients receiving no active treatment would also experience at least one recurrence. In both of these studies, however, initial episode recovery time was not considered and has subsequently been found to be a potential risk factor for recurrence (Stanton et al., 2008). In their cohort of 353 patients who had recovered within six weeks of initial consultation, Stanton et al. (2008) reported the total recurrence rate to be substantially lower at 33 %, suggesting that high recurrence

rates reported in the literature may be somewhat of an overestimate if initial episode recovery time is not considered. Additionally, however, the authors also reported the largest predictor of recurrence within their model was prior episodes of LBP in the 12 months pre-study enrolment.

In a study including a seven-year follow-up of 444 patients initially seeking GP consultation for aLBP, it was found that chronicity of pain developed in 28 % of the people with the majority (61 %) of these patients also reporting that pain was at least as severe (47 %) or worse (14 %) than at the time of initial consultation (Miedema et al., 1998). Recently, in a large scale comparative study of cLBP prevalence between 1992 and 2006 in the United States, an alarming increase of prevalence from 3.9 % to 10.2 % was found over the 14 year period (Freburger et al., 2009). In the same study, the authors also noted an increase in instances of aLBP from 7.3 % to 10.5 %, with this smaller increase in prevalence of aLBP vs cLBP being consistent with a greater percentage of acute cases transitioning to chronicity.

Whilst in some cases the symptom of pain can be attributable to a specific pathology, these are in the minority (Balagué et al., 2012). As such it is commonly estimated that a diagnosis based on exclusion of specific pathology, termed non-specific low back pain (nsLBP), accounts for 85-90 % of all people with LBP (Airaksinen et al., 2006; Deyo, Rainville, & Kent, 1992; Koes, Van Tulder, & Thomas, 2006; Wand & O'Connell, 2008).

Thus, LBP should not be regarded as a benign, self-limiting disease that will tend towards spontaneous improvement (van Tulder et al., 2000; Waddell, 1987) but rather one of a multi-episodic, recurrent course characterised by variability, a cumulative risk

of chronicity, and a high likelihood that no specific cause can be identified (Burton, 2005; Hestbaek et al., 2003; Von Korff & Saunders, 1996).

2.4 The Muscle Capacity and Control Model

The muscle capacity (strength and endurance) and control (neural) model centres on the premise that spinal stability is dependent on the proper and optimal contributions of the surrounding musculature (Hodges, 2003). Devoid of this musculature, the osteoligamentous cervical, thoracolumbar and lumbar spines will experience structural failure under compressive loadings as small as 10, 20 and 90 N in magnitude, respectively (Crisco et al., 1992; Panjabi et al., 1998). Considering spinal loadings experienced *in vivo* can range from 6 kN during selected everyday tasks (McGill & Norman, 1986) to in excess of 36 kN during competitive powerlifting (Cholewicki, McGill, & Norman, 1991) the human vertebral column is intrinsically incapable of meeting the physiological demands placed upon it without additional stabilisation.

A classical view of spinal stability refers to the ability of the spine under physiological loads to limit displacement in each motion segment (White III et al., 1975). This concept has since been further refined such that total range of motion includes the non-linear load/displacement curves (Figure **2-9**) of a neutral zone and an elastic zone (Panjabi, 1992b). Panjabi also used the analogy of a ball in a wine glass to describe spinal stability, where the ball could move freely at the base of the glass (within the neutral zone) but the steeper sides (elastic zone) provide increasing resistance. Compatible with the muscle capacity and control model, it is the role of the surrounding musculature to control the size of the neutral zone and failure to do so can result in deformity, neurologic deficit and pain (Panjabi, 1992b).

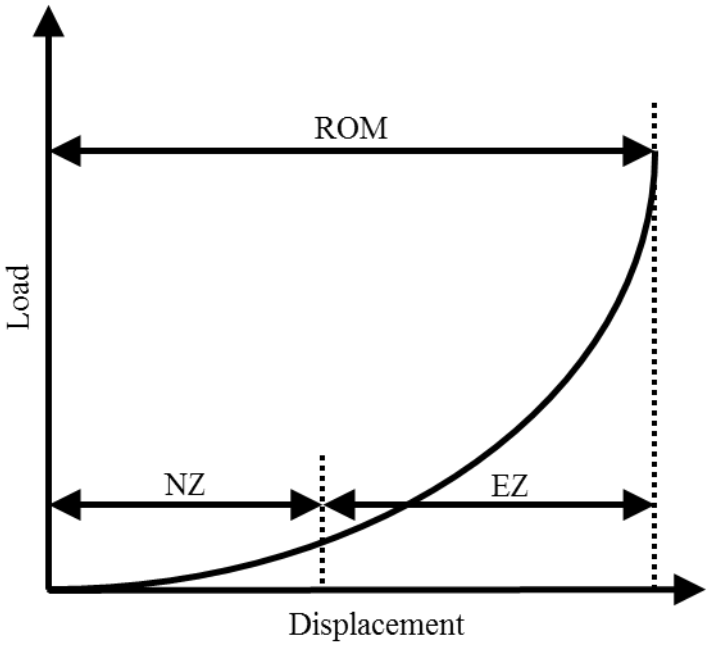

Figure 2-9. Load/displacement curve of hypothesised spinal motion segment. The range of motion of the spinal joints includes an initial neutral zone (NZ) with relatively low loads incurring large displacements and an elastic zone (EZ) requiring greater load per unit of displacement. Redrawn from Panjabi (1992b), with permission (Appendix F).

2.4.1 Roles of Lumbar Multifidus and Transversus Abdominis in Spinal Stability

Using the classification scheme of Bergmark (1989), Panjabi (1989) suggested that the role of the global muscles in stabilisation was to augment the stiffness of the entire vertebral column, and the local muscles act more to control intersegmental motion. Debate continues, however, as to which muscles are important intersegmental spinal stabilisers and how best to ensure sufficient stability (Kavcic, Grenier, & McGill, 2004; McGill, 2015) with some authors suggesting that no single muscle (local or global) holds a dominant responsibility for lumbar spinal stability (Cholewicki & McGill, 1996; Cholewicki & Van Vliet 2002; McGill, 2015).

In recent years, a growing body of evidence has highlighted the significant roles of two local muscles, namely the *lumbar multifidus* (LM) and the *transversus abdominis* (TrA), in the correct functioning of the lumbar spine, and in low back pain.

Paradoxically, people with low back pain have been shown to demonstrate both a stiffening strategy via hyperactivity of the global trunk muscles and a passive postural strategy via hypoactivity (Brumagne et al., 2008; Henry et al., 2006; Hodges, 2003; van Dieën, Selen, & Cholewicki, 2003). Thus, some individuals may increase global muscle activity to stiffen the spine, whereas others may not and will consequently be reliant on the passive structures for stability. In the short term, either strategy may serve a functional benefit (buckling prevention) in people with LBP (Lund et al., 1991), although this may subsequently come with a loss of ability to finely tune intersegmental motion (Hodges, 2003).

2.4.1.1 Contribution of Lumbar Multifidus to Spinal Stability and Low Back Pain

Whereas previously the muscle capacity and control model focussed on the superficial torque generators (Manniche et al., 1988; Nelson et al., 1995), it has since been argued that the deeper fibres of the paraspinal muscles are most suited to controlling intersegmental motion around the neutral zone (Demoulin et al., 2007; Freeman, Woodham, & Woodham, 2010; MacDonald, Moseley, & Hodges, 2009; Richardson & Jull, 1995; Ward et al., 2009).

As previously noted, the anatomical arrangement of the deeper fibres of the LM (dLM) is markedly different from other spinal muscles such as the lumbar *erector spinae* (LES) and the more superficial fibres of the LM (sLM). Mechanical models based on these arrangements have highlighted some interesting differences (Bogduk,

Macintosh, & Pearcy, 1992; Kim et al., 2007; Macintosh & Bogduk, 1986), specifically that LES and sLM contribute approximately 80 % of the extensor torque at L4/5 whereas dLM generates primarily compressive forces. Such compressive forces generated by the dLM have also been shown *in vitro* to contribute greatly (approximately two thirds) to intersegmental stiffness (Wilke et al., 1995). It should be noted, however, that all intrinsic spinal muscles contribute to spinal stiffening and this effect can be much greater should a co-contraction strategy between superficial flexors and extensors be utilised (Gardner-Morse & Stokes, 1998; Granata & Marras, 2000). This strategy, however, is at the cost of increased effective spinal load, which if sustained longitudinally may have detrimental effects (Granata & Marras, 2000).

Consistent with this proposed primary action of stabilisation is that dLM and sLM fascicles are differentially activated in healthy individuals in a range of tasks. Using intramuscular electromyography (intEMG), Moseley et al. (2002) found that, whereas the dLM, sLM and LES were all active during an upper limb movement perturbation, only dLM was not temporally sensitive to changes in movement direction. That is to say, dLM acted in a consistent non-direction-specific manner and sLM and LES were direction-dependant. The authors' suggested explanation was that increasing intervertebral stiffness via the dLM is likely the optimal strategy as opposed to the alternative agonist-antagonist co-contraction strategy which would result in an excessive energy and compressive cost (Granata & Marras, 2000) that may lead to subsequent nociceptor stimulation and pain (Panjabi, 1992a).

Subsequently, this same group further refined this differential activation model using unpredictable perturbations (Moseley, Hodges, & Gandevia, 2003). In a manner similar to the first study, intEMG was recorded from the dLM and sLM, although in

this study the participants could either predict a perturbation (self-initiated release of a hand-held weight) or not (investigator initiated release). When the perturbation was predictable, a response concurrent to the first study was found in that activation of the dLM occurred differentially (prior) to that of the sLM and LES. When the perturbation was not predictable, this temporal distinction was not evident. Together, these studies show that the dLM and sLM are differentially active in an anticipatory non-direction-specific manner when a perturbation is expected or predictable. As such, provided the requisite information is available the neural control system would rather plan muscle activity such that dLM is acting in advance of, not in response to, perturbation.

In the presence of pain, however, a number of associated changes in this typical behaviour of *lumbar multifidus* have previously been observed. When exposed to acute pain induced experimentally with hypertonic saline (5 %) injections into the longissimus muscle at the L4 level, intEMG activity of the dLM is increased, whilst that of the sLM is decreased (Moseley, Hodges, & Gandevia, 2003) suggesting immediate changes in postural control strategy in the presence of painful stimuli. Changes to trunk activity mediated stability is also evident in clinical LBP, and interestingly during a remission episode of recurrent LBP (MacDonald, Moseley, & Hodges, 2009). Where in a healthy control group MacDonald et al. (2009) found dLM and sLM to be differentially active in an anticipatory non-direction-specific manner consistent with Moseley et al. (2002; 2003), this was not the case in the recurrent LBP comparison group. In this group, dLM and sLM of people with LBP were found to be recruited simultaneously during both flexion and extension of the upper limb. This suggested a potential mechanism for recurrence and subsequent chronicity. Interestingly, this en-masse recruitment strategy has also been evidenced by comparative differences in representation at the level of the motor cortex in people

with recurrent LBP compared with healthy controls (Tsao, Danneels, & Hodges, 2011). Using transcranial magnetic stimulation Tsao and colleagues (2011) found a discrete reorganisation of the motor cortex such that in a healthy population the LES and dLM were distinctly separated, whereas for patient with recurrent LBP they were not. In the healthy control participants, dLM was also represented by a greater area of activation to that of people with recurrent LBP. Together these findings support the hypothesis that dLM is functionally disparate to the more superficial musculature with a more complex neural control strategy and that this is altered to a simpler en-masse strategy in the presence of LBP. It should be noted, however, that the control and LBP groups were relatively young (control, 24 ± 5 years; LBP, 25 ± 3.4 years) and this finding is yet to be confirmed in older adults.

A number of authors have also highlighted several inconsistencies in LM form and function in the presence of low back pain. Hides et al. (1994) found asymmetries in LM cross-section area (CSA) measured with ultrasound imaging (USI) ipsilateral to the clinically determined vertebral level of symptom provocation in people with unilateral aLBP. This between-side asymmetry was 31 ± 8 % in the patient population compared with asymmetry of < 6 % in the healthy control group. It was further found that, despite resolution of symptoms in people with such asymmetry, recovery of muscle size at the affected level does not occur if only medical management was prescribed (Hides, Richardson, & Jull, 1996). Similar morphological deficits have also been shown in cLBP, where CSA measured with computed tomography (CT) scanning of the LM was also significantly smaller at the L4/5 level than that seen in healthy matched controls (Danneels et al., 2000). Moreover, intramuscular fatty infiltrates in LM have also been found to be greater in people with cLBP (Kader, Wardlaw, & Smith, 2000).

Considering this evidence of change in morphology of the LM comes from retrospective studies of existing pain populations it would seem pertinent to question whether such changes are causal in nature or a resultant adaptive mechanism. For example in the study by Hides et al. (1994) it is difficult to ascertain whether the approximate 30 % reduction in CSA of the *lumbar multifidus* which was observed within days of the onset of symptoms can be explained as a response to injury/pain, or that such changes were pre-existing. Using an animal model, Hodges et al. (2006) found that following an experimentally induced injury (disc lesion at L3-4) in nine pigs, a rapid reduction of CSA of approximately 17 % at the level of the injury occurs within three days. In the same study, Hodges et al. (2006) also found that following an induced nerve lesion to the medial branch of the dorsal ramus at L3-L4, a similar reduction in CSA was observed at L4, although it affected a greater number of levels distally (L4, L5 and L6). These findings suggest that following a localised injury to a particular structure of the lumbar spine, rapid segmental atrophy is likely to ensue.

Although USI techniques were initially used as a tool to investigate the morphometric attributes of LM, it has since been shown that USI can also provide simple, non-invasive insights into recruitment and activation strategies (Hodges et al., 2003; Kiesel et al., 2007a; McMeeken et al., 2004). Hodges et al. (2003) and McMeeken et al. (2004) initially validated this technique using isometric contractions whilst simultaneously recording intEMG and measuring changes in linear thickness of several muscles of the lateral abdominal wall including obliquus internus and externus, and TrA. In both instances changes in intEMG correlated strongly ($p < 0.001$; $r = 0.93$; $r^2 = 0.87$) with observable changes in muscle thickness, at least at intensities below ~30 % of maximum voluntary contraction. In a manner similar to Hodges et al. (2003) and McMeeken et al. (2004), Kiesel and colleagues (2007a) also demonstrated that

this technique was valid for LM. Using simultaneous USI and intEMG it was found that LM thickness change also correlated well with increases in graded activity ($p < 0.001$; $r = 0.79$ $r^2 = 0.62$). As LM thickness is taken as a linear measurement between the posteriormost portion of L4/5 facet joint and the thoracolumbar fascia, however, it is important to note that any change in thickness could be as a result of thickness changes of the sLM or dLM alone, or in combination.

Using USI derived measurements of thickness change has further highlighted some interesting disparities between functioning of the LM in healthy individuals and those with LBP (Kiesel et al., 2007b). Based on the treatment based classification system of Delitto et al. (1995), people with LBP categorised as either requiring stabilisation or direction-specific exercise were found to demonstrate attenuated thickness change of the LM at L4/5 by Kiesel et al. (2007b) when performing a weighted arm raise. A similar finding was also reported by Wallwork and colleagues (2009) in that when compared to a healthy control group, people with LBP demonstrated a significantly lower change in LM thickness at the level of L5 when asked to voluntarily contract isometrically. These authors also reported findings consistent with those reported previously (Hides et al., 1994) in that CSA of the LM was also significantly reduced at the L5 level in the LBP group compared to that of the controls. A similar finding was also reported when pain was experimentally induced in healthy participants with hypertonic (5 %) saline solution injected into the longissimus muscle adjacent at the L4 level (Kiesel et al., 2008). Percentage increase in LM muscle thickness was significantly lower during graded weighted upper extremity raises in the induced pain trials compared to that observed during the control condition.

2.4.1.2 Contribution of Transversus Abdominis to Spinal Stability and Low Back Pain

Although TrA was not originally designated to the local subsystem of stabilising muscles by Bergmark (1989), a growing body of evidence has supported its role in spinal stability alongside its role in respiratory function (Stokes, Gardner-Morse, & Henry, 2010; Stokes, Gardner-Morse, & Henry, 2011; Wang & McGill, 2008). An initial investigation by Cresswell, Grundström and Thorstensson (1992) found that intra-abdominal pressure (IAP) was consistently related to TrA activity during maximum voluntary isometric flexion and extension tasks, whereas this was not the case for EO and IO. Additionally, it was observed that during trunk rotation reciprocal activity was present indicating a role in torque development. Furthermore, Cresswell (1993) also reported that a similar mechanism was active during dynamic flexion and extension with TrA again related to IAP development, although more so in flexion. The authors surmised that this ability to generate intra-abdominal pressure independently of the trunk extensor moments generated by EO and IO could be related to a role within spinal stabilisation.

The response of the TrA during predictable and unpredictable perturbations (anterior and posterior trunk loadings) has also been studied (Cresswell, Oddsson, & Thorstensson, 1994) using a method latterly adopted by Moseley et al. (2002, 2003) in their studies of LM function. Here the authors, again using intEMG, determined the activity onset latencies of several muscles of the lateral abdominal wall including the TrA, IO and EO, as well as the LES. With self-initiated anterior perturbations, onset of TrA activity was observed to occur first in advance of (-175 ± 24 ms) and then concurrently with initiation of IAP increase (-170 ± 16 ms) followed by onset of the OI (-110 ± 13 ms), and OE (-85 ± 18 ms). Activity onset of the RA (-55 ± 26 ms) and

LES (-58 ± 19 ms) were simultaneous with one another although again occurring prior to perturbation. As expected when an anterior load was applied unpredictably, activity onset of the TrA, IO, EO and RA was temporally deferred. What is also interesting to note is that, in this instance, abdominal muscles were all recruited en-masse, followed by a later activation of the LES. A similar pattern was also seen during unpredictable posterior loading. Together, this indicates a feed-forward anticipatory response similar to that of the LM, increasing IAP as a mechanism to increase trunk stability in readiness for an ensuing perturbation.

In a following series of studies, Hodges and Richardson further validated this movement model with both lower limb (1997) and upper limb (1997) movements. Again, using intEMG assessment of the TrA, EO, IO, and RA, it was found that when performing either standing hip extension or flexion, a consistent pattern of activity onset latency was evident. Firstly, TrA, then IO and EO, and finally RA. When participants performed an arm-raising task, TrA was again consistently the first muscle active, supporting the hypothesis that TrA acts in anticipation of a perturbation to the spine to increase trunk stiffness and minimise disturbance.

There are also data to suggest that although the TrA acts similarly to LM in its feed-forward anticipatory response, activation may be directionally sensitive when examined bilaterally (Allison, Morris, & Lay, 2008). Here, the authors implemented the existing arm raising perturbation model as described previously, although, in this instance, TrA, EO, IO were studied bilaterally. When a unilateral arm raise was performed, there was a finding of anticipatory responses of contralateral TrA consistent with the studies mentioned above. Although, TrA activation ipsilateral to the side of arm raising demonstrated a temporally deferred onset. Thus, muscles of the

abdominal wall are differentially active, with TrA activation being the predominant response, in an anticipatory, non-direction-specific manner (contralaterally), but sensitivities exist between sides.

Similarly to LM, inconsistencies have been highlighted between these 'normal' pre-programmed functional responses and those observed in people with LBP during arm raising. When compared with a group of matched controls, the individuals with LBP demonstrated significantly delayed TrA activation that was outside of the feed-forward window during all movement directions (Hodges & Richardson, 1996). Onset of EO, IO and RA activation was also delayed, although this only occurred in arm flexion (Hodges & Richardson, 1996). Variation between healthy individuals and people with LBP has also been shown when movement velocity of the upper limb (disturbance magnitude) is manipulated (Hodges & Richardson, 1999). Here, the typical pattern of recruitment for healthy individuals is that of a decreasing latency (greater anticipatory response) with increases in movement velocity. In people with LBP, however, the response is atypical and, generally, TrA, IO and EO are activated in response to perturbation, with no adaptation apparent to the increasing magnitude of perturbation (Hodges & Richardson, 1999). In light of the theory that the passive elements of the spine are sufficiently capable of maintaining stability if the disturbance is small (Reeves, Narendra, & Cholewicki, 2007), this further supports the idea that people with LBP may adopt a simple bracing strategy to maintain spinal stiffness. In 'normal' functioning, however, the strategies appear to be more adaptive to the individual demands of any given situation.

Using USI, differences in function between healthy control participants and people with a history of LBP have also been evidenced (Ferreira, Ferreira, & Hodges, 2004).

Using an isometric knee extension/flexion task, Ferreira and colleagues (2004) demonstrated that for both small (7.5 % body weight) and large (15 % body weight) external loads in flexion and extension, TrA thickness change was significantly lower in people with LBP than that observed in control participants. Although the mechanism for dysfunction here cannot be truly known, induced experimental pain using hypertonic (5 %) saline injected into the longissimus muscle at the level of L4 also reduces activity of the TrA (Moseley, Hodges, & Gandevia, 2003). This suggests that pain can be an initial stimulus for alteration of motor control strategies that may not resolve during pain free periods.

These studies primarily investigate automatic recruitment strategies in response to some form of external disturbance to trunk stability. In an instance where a volitional contraction of the TrA - as in the abdominal drawing in manoeuvre (ADIM) – has been investigated, it was found that no differences existed between the healthy control population and that of people with lumbopelvic pain (Teyhen et al., 2009a). Although in people with LBP (including experimentally induced LBP) this voluntary facilitation of TrA contraction has been reported to be significantly lower (Critchley & Coutts, 2002; Kiesel et al., 2008; Kiesel et al., 2007b). This, therefore, suggests the role of TrA within LBP may be primarily a motor control issue rather than one of absolute capacity.

Given the apparent loss of function of the LM and TrA (either as a result of LBP or a precursor to it) it is crucial to establish strategies to restore normal TrA and LM function and to find out whether these strategies can also reduce or eliminate pain.

2.5 Treatment Strategies for Low Back Pain

Until relatively recently, the most common treatment strategy for aLBP was that of short term bed rest (Malmivaara et al., 1995). However, a series of systematic reviews of randomised controlled trials (Hagen et al., 2004; Koes & Van Den Hoogen, 1994; Van Tulder, Koes, & Bouter, 1997) have since found either no positive effect - and in some instances negative effects - of bed rest using outcomes of pain, recovery duration and return to daily activities compared to alternative treatment strategies such as exercise, physiotherapy, manipulation or non-steroidal anti-inflammatory drugs (NSAIDs). This led to the recommendation of the COST B13 Working Group that passive treatment modalities such as bed rest should not be prescribed and patients should instead stay active or become active (Van Tulder et al., 2006).

2.5.1 Exercise Therapy

According to the fear-avoidance-deconditioning paradigm of LBP (Figure **2-10**), an individual experiencing pain may firstly interpret their pain as threatening, which can then lead to fear of movement, limiting future activity leading to secondary muscle atrophy that further limits activity and compounds the problem further still (Vlaeyen & Linton, 2000). Such a deconditioning correction strategy forms the basis of many exercise therapy interventions, although these strategies encompass a rather heterogeneous grouping including general strength and endurance, flexibility and aerobic exercises, as well as more specific approaches such as segmental stabilisation (Nordin & Campello, 1999).

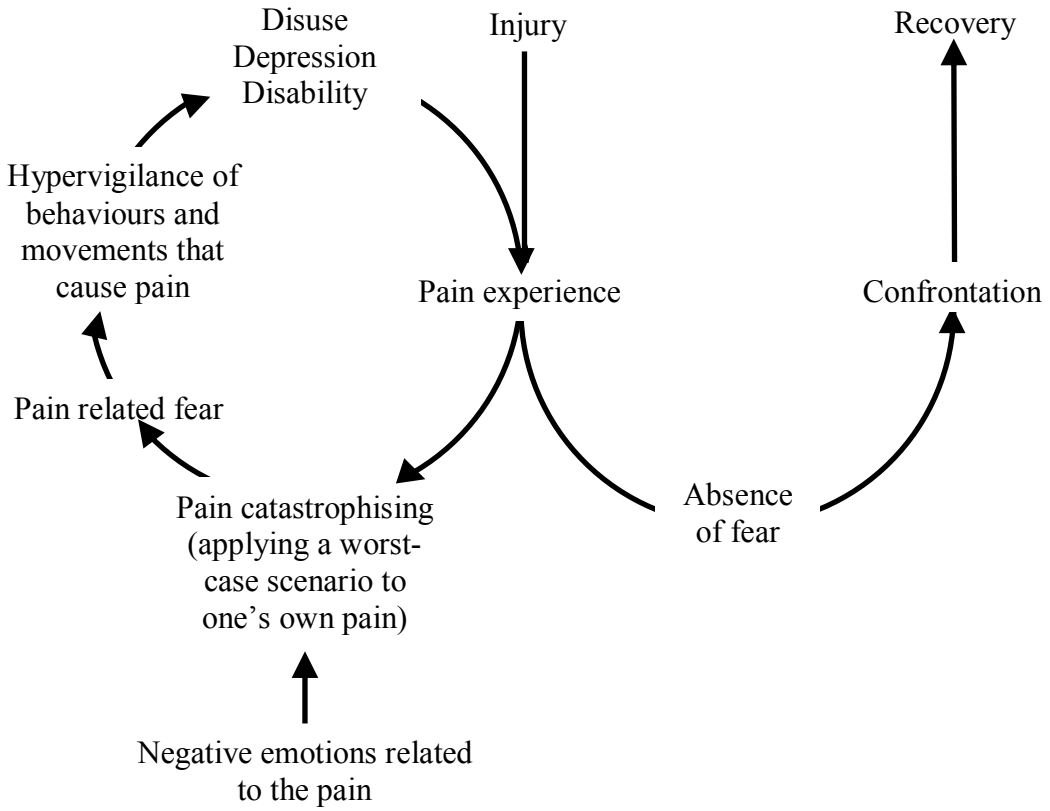

Figure 2-10. Fear-avoidance model of chronic pain development informed by Vlaeyen and Linton (2000) highlighting the compounding negative effects of fear of pain and subsequent avoidance strategies on future pain experience by way of a disuse and deconditioning mechanism. Reproduced with permission (Appendix G)

Indeed, where individuals demonstrate fear-avoidance beliefs treatment strategies based on education in addition to graded exercise have been found to be successful in reducing disability (George et al., 2003), whereas in individuals low in these beliefs, the cognitive based therapy appears counterproductive. Pain related fear therefore, should not be overlooked when establishing the effectiveness of exercise based interventions.

2.5.1.1 Aerobic Exercise Training

Consistent with the deconditioning theory is the assumption that a decrease in physical activity and aerobic fitness exists in people with cLBP, however this remains unclear within the literature (Bousema et al., 2007; Smeets et al., 2006). In a longitudinal

inception cohort study, Bousema et al. (2007) used seven-day actigraphy on 89 people (not including those with incomplete datasets or withdrawals) with subacute (4-7 weeks) LBP at baseline and at one-year follow-up. They found that 62 patients developed cLBP and 27 patients recovered, and a decrease in physical activity was only seen in less than half of the patients. Of those who developed cLBP, there was even a significantly increased mean score for physical activity. The authors hypothesised that increased physical activity may have caused chronicity to develop by way of overuse and muscular hyperactivity, although the data presented could not substantiate this claim. It should also be noted that habitual physical activity was not monitored at any other points other than baseline and follow-up. As such, these findings may not truly reflect the general course of physical activity during the intervening period. In contrast to the study by Bousema et al. (2007), significant differences in aerobic fitness ($\dot{V}O_2max$) were observed by Smeets et al. (2006) in 84 people with cLBP versus healthy controls. Males had a greater absolute difference (10 $ml \cdot kg^{-1} \cdot min^{-1}$) than females (5.6 $ml \cdot kg^{-1} \cdot min^{-1}$) with an overall mean difference of 8 $ml \cdot kg^{-1} \cdot min^{-1}$, equivalent to 2.3 METS (metabolic equivalents).

Whilst a direct causal mechanism is unlikely, it is possible that such general physical activity deconditioning is a cumulative effect of pain related fear-avoidance in cLBP. Evidence for this comes from a recent systematic review and meta-analysis (Lin et al., 2011) that found only a weak and non-significant relationship (r = -0.08) between physical activity in acute/sub-acute LBP and a stronger significant negative relationship (r = -0.33) in cLBP. As such, it is most likely that any positive effect of aerobic training is a result of reduced psychosocial stress response that subsequently attenuates pain catastrophizing in people with cLBP and not of any functional restoration (Verbunt et al., 2003).

2.5.1.2 Strength and Endurance Training

On the basis of several observations of reduced strength (Lee, Ooi, & Nakamura, 1995; Mayer et al., 1985; Taimela & Härkäpää, 1996) and endurance (Biedermann et al., 1991; Luoto et al., 1995; Mayer et al., 1989; Nicolaisen & Jørgensen, 1984; Roy, De Luca, & Csavant, 1989) of the trunk extensor muscles in people with LBP, it was hypothesised that this could limit their capability to prevent spinal instability in the presence of perturbation (van Dieën, Selen, & Cholewicki, 2003).

In an early clinical trial, Manniche (1988) randomised 105 people with cLBP into one of three groups with each group being either a high or low intensity progressive resistance exercise or a usual care control group. The high intensity exercise group performed 50 sets of hip and back extensions and pull downs three days per week for 4 weeks, and then twice per week for a further eight weeks. The low intensity group was similar to the high intensity group with the exception that only 20 sets of each exercise were performed. The usual care (control) group also received passive therapy including thermotherapy (heat packs) and massage plus mild flexion exercises for four weeks. The high intensity group demonstrated significant improvements in isometric extensor endurance, pain, disability and physical impairment, whereas little to no change was observed in the low intensity and usual care groups. If high intensity exercise was continued at least once per week during a one-year follow-up period, improved symptomology was maintained (Manniche et al., 1991).

In a similar study, Nelson et al. (1995) enrolled 734 people with cLBP with an average symptom duration of 26 months for participation in an intensive trunk strengthening programme. Of these 734 patients, 107 opted not to participate in the exercise programme following initial assessment and served as the control group, leaving the

remaining 627 as active participants. The intervention consisted of both lumbar extension and rotation exercises to fatigue and ended when patients were either pain-free, improvements in spinal function plateaued or patient effort was no longer considered sufficient. On average, patients required 18 one-hour sessions with two sessions per week. Significant improvements were seen in isometric and dynamic strength assessments, and for 64 % of patients a substantial decrease in pain was observed. Those people with the largest improvements in lumbar extension strength also demonstrated greater improvements in symptoms. Furthermore, at one-year follow-up patients' improvements in symptoms were maintained. However, this research design involved non-random allocation of participants to the control group that could be seen as a potential source of bias, particularly if fear-avoidance beliefs were present in those individuals opting not to participate in the exercise programme.

Although the outcomes of this kind of strategy appears favourable, it is noticeable that any restoration of function is not targeted specifically at the local stabilising musculature such as the TrA and LM but rather the more superficial global torque producing muscles, such as the ES and RA. As a response to this mismatch, in a seminal paper, Richardson and Jull (1995) described the basis of a hybrid therapeutic exercise model focussed primarily on the restoration of capacity, control and function of the LM and TrA termed "specific stabilisation exercise".

2.5.1.3 Specific Stabilisation Exercise (Motor Control Training)

Specific stabilisation exercise, also known as motor control training (MCT) is founded upon five fundamental features including 1) sustained isometric contractions (around 10 seconds) at a low level of patient specific MVC (approximately 30-40 %), 2) co-contraction of the LM and TrA, 3) preferential recruitment of local rather than global

muscles, 4) progressive increases in contraction volume, and 5) progressive introduction of functional body positions with increases in external load (Richardson & Jull, 1995).

This model of therapeutic exercise is consistent with many of the proposed 'needs' for a strategy aiming to restore proper capacity, control and function of the stabilising elements of the lumbar spine. A number of trials have reported on its effectiveness in LBP patient populations (Costa et al., 2009a; Ferreira et al., 2007; Hides, Jull, & Richardson, 2001; Hides, Richardson, & Jull, 1996; Hides & Stanton, 2014; Hides et al., 2012; Koumantakis, Watson, & Oldham, 2005; O'Sullivan, Twomey, & Allison, 1997; Streicher et al., 2014; Unsgaard-Tøndel et al., 2010).

In one of the first RCTs exploring MCT effectiveness, O'Sullivan et al. (1997) allocated a group of people with clinical spinal instability (a radiologic diagnosis of spondylolysis or spondylolisthesis) to receive either MCT (n = 21) or general medical management (typically passive thermotherapy, massage, swimming, walking). Individuals in the experimental treatment group underwent a ten-week long program of MCT with one session per week lasting between 10-15 minutes. Immediately post intervention, participants were reassessed and then again at 3-, 6- and 30-months follow-up. Results showed that pain intensity and functional disability were significantly improved in the MCT group who maintained this improvement for the duration of follow-up. This was not the case in the control group, as pain intensity and functional disability remained unchanged throughout.

In a similar study, Hides and colleagues (1996) allocated a group of people with acute (<3 weeks), first episode unilateral LBP to receive either medical management (MM)

alone or MM supplemented with MCT (MM+MCT). Medical management (n = 20) consisted of advice on bed rest (1-3 days maximum), absence from work and prescription of analgesic medication including aspirin, paracetamol, codeine, NSAIDs and Valium for four weeks. The MM+MCT group received the general MM, but this was supplemented with MCT including home exercises. No differences were found between groups and decreases in both pain and disability paralleled one another. However, CSA asymmetries of the LM were restored in the MM+MCT group but not in the MM alone group. Subsequently, when these patients were followed up long-term (Hides, Jull, & Richardson, 2001) it was found that of those in the MM+MCT group, participants were 12.4 times less likely to experience symptom recurrence than those who had received MM alone in the preceding 12 months. After three years, this had reduced to 9 times less likely; however, the LBP episodes of those who did experience recurrence were predominantly linked to a traumatic incident. Thus, MCT may be effective in instances where specific clinical instability (spondylolysis or spondylolisthesis) is present but not in first episode nonspecific aLBP. Evidence is available, however, to support the notion that MCT may protect individuals from future recurrence.

In a controlled trial of MCT in people with recurrent LBP (Koumantakis, Watson, & Oldham, 2005), patients were allocated into one of two groups, either a general exercise (GE) group or a GE group with supplementary MCT (GE+MCT). Both GE and GE+MCT programmes consisted of 8-weeks (twice per week) of sessions (45-60 minutes each). General exercises focussed on classical trunk flexor and extensor training while GE+MCT integrated the prescriptions of Richardson and Jull (1995). It was found that whilst following this programme GE and GE+MCT both significantly improved pain and disability both immediately post and at 20 weeks follow-up and

there were no differences between groups. The authors concluded that MCT was of no additional benefit. However, these results should be interpreted with a great degree of caution. Classic flexor/extensor training and MCT are fundamentally different and as previously mentioned, the classic model may not be the most suitable method. As such, it is not surprising, that supplementation with MCT was not successful in this instance.

When MCT and GE are directly compared in a cLBP population the findings are more favourable (Akbari, Khorashadizadeh, & Abdi, 2008). Here, the authors found that, following two 30-minute sessions per week for eight weeks, pain and functional ability improved in both groups, though improvements in the MCT group were significantly greater than the GE group.

In a further study, Rasmussen-Barr et al. (2003) allocated a group of people with either aLBP or cLBP to receive either MCT (n = 22) or manual therapy (MT) (n = 20). Each group received one session per week (45-minutes) for a period of six weeks and was assessed for pain and functional disability at baseline, immediately after treatment and after 3 and 12 months. It was found that the MT group showed no improvements in pain or function in either the short- or longer-terms. The MCT group, however, did experience significant improvements in pain and functional disability in both short- and longer-terms.

In a series of systematic reviews and meta-analyses (Ferreira et al., 2006; Hauggaard & Persson, 2007; Macedo et al., 2009; May & Johnson, 2008; Smith, Littlewood, & May, 2014) the overall effectiveness of MCT has also been examined. The consensus of these reviews is that MCT is not an effective strategy in reducing pain and functioning in aLBP but can be for cLBP. Additionally, there is limited further benefit

from combining MCT with other active therapy modalities and, in comparison to general exercise alone, there is no clinically meaningful difference. It is increasingly effective, however, when used in sub-groups of people with demonstrable dysfunction of the LM and TrA, and when used as a single treatment strategy.

Additionally, there is an increasing recognition of the limitations of traditional RCTs in studies of the effectiveness of therapeutic interventions (Dreyer et al., 2010; Horn & Gassaway, 2010; Howard, Best, & Nickels, 2015). Particularly, in relation to people with low back pain, the use of such RCTs has been heavily criticised (McGill, 2013), due in part to the varied aetiology of low back pain, and a range of psychosocial aspects influencing individuals' responses to both the pain and attempted interventions. For this reason the use of relatively small sample sizes consisting of carefully screened participants, ensures they are comparable in the type of low back pain they experience and likely to benefit from the intervention (McGill, 2013).

The feed-forward dysfunction of TrA has previously been shown to be responsive to a single session of isolated voluntary isometric contractions in individuals with recurrent LBP (Tsao & Hodges, 2007). However, one session was not enough to achieve full recovery of function. Even following a number of further training sessions, the recovery of function was found to be incomplete, and improvements were found to plateau. However, the changes persisted for up to six-months (Tsao & Hodges, 2008). Such changes have not yet been described in LM, although positive responses in terms of gross morphology of the LM have.

In their study of young elite cricketers, Hides et al. (2008) allocated individuals with LBP (n=7) to a rehabilitation group that consisted of a six-week MCT training

programme and individuals without LBP (n=14) to a usual training group. Following the rehabilitation intervention period both groups were given a one-week break and continued regular training for a further six-weeks. Bilateral CSA of the LM was assessed using ultrasound imaging in both groups, prior to and following the 13-week training period. At baseline, it was found that in the LBP group, a specific asymmetry in CSA at the level of L5 existed that did not exist in the group without LBP. However, post MCT training this asymmetry was no longer present, and additionally CSA was now significantly greater in the MCT group than that in the usual training group.

Motor control training may be effective in augmenting positive changes LM and TrA motor control and morphology; however, with the MCT method of training, this tends towards a plateau and appears only partial. Although the fundamental principles of MCT appear, therefore, to be reasonable, it may not yet be a fully developed strategy and could benefit from further developments. One potential feature that may need further development is the strategy for integration into functional activities. Although the MCT protocol does include incorporation of LM and TrA activation into "dynamic functional movements of the trunk" (Richardson & Jull, 1995) the implementation of this typically involves relatively simple body postures.

2.5.2 Exercise Using the Functional Re-adaptive Exercise Device

Relatively recently, an exercise device (Figure **2-11**) was designed that purported to specifically target the deep lumbopelvic muscles during a cyclical lower limb flexion/extension perturbation (Korfmacher, Debuse, & Pinotti, 2006). The kinematics of the exercise is similar to that performed on an elliptical trainer where the feet follow an anti-phase quasi-elliptical path; however, it offers negligible external resistance to motion. As such, in order for the user to achieve a smooth motion, the frequency of

motion must be kept low. Additionally, the absence of external resistance results in the need for much greater motor control of the legs and pelvis for balance than in conventional exercise devices. As one foot moves downwards through the front of the movement cycle, the muscles in the user's rear leg have to work also to maintain a smooth and controlled motion of the lower limbs. The device may be used in either a sitting or upright standing posture with users also permitted to use the available hand rests (albeit minimal contact is encouraged, so as not to negate the need for deep lumbopelvic muscle contribution) if necessary. The resultant effect is hypothesised to be a specific automatic low level of isometric co-contraction of the LM and TrA without global muscle contraction substitution during a functionally relevant movement.

The exercise device was first examined by Debuse and colleagues (2013), and it was substantiated that this mode of exercise can induce automatic contraction of the LM and TrA. Here, the authors examined LM and TrA thickness change in a mixed gender group (6 male, 6 female) of asymptomatic individuals during exposure to a range of static and dynamic weight-bearing (standing) and non-weight-bearing (sitting) conditions. Thickness change was not significantly different in the LM ($p = 1.00$) or TrA ($p = 0.190$) during static stable and unstable standing. When a functional cyclical perturbation of the lower limbs using the exercise device was introduced, recruitment of TrA was significantly increased in both sitting ($p = 0.003$) and standing ($p < 0.001$) postures. Interestingly, exercise using the device did not significantly increase recruitment of the LM in a sitting posture ($p = 0.349$), although it was significantly increased during exercise in standing ($p = 0.006$). These findings suggest that this exercise modality may be a worthwhile inclusion in the progression strategy of MCT programmes given its functional nature and the ability to recruit LM and TrA non-

volitionally, and both concomitantly or differentially depending on the weight/non-weight-bearing and sitting/standing combination used.

Since then the authors have also recommended this modality for use as a complementary exercise therapy following long-term bed rest (Evetts et al., 2014) which is an experimental protocol commonly used as an analogue to microgravity exposure (Hides et al., 2007a) known to induce many of the dysfunctions commonly observed in people with LBP (Belavý et al., 2011; Belavý et al., 2007).

Figure 2-11. Functional Re-adaptive Exercise Device designed by Korfmacher et al. (2006) illustrating the general components of the device

2.5.3 Summary

Currently LBP is a highly prevalent and increasingly common worldwide health problem that is associated not only with a huge economic burden but also a potentially life affecting burden to the individuals troubled by it, should it follow a persistently recurrent and/or chronic clinical course. Though the body of evidence concerning this truly complex and heterogeneous syndrome has grown substantially over the years, it has thus far steadfastly resisted any individual treatment strategy. The multifaceted nature of the problem requires a multifaceted approach to rehabilitation with each aspect of the treatment strategy tailored to the particular characteristics of the individual.

Although the mode of exercise proposed here demonstrates an initial degree of promise, as yet a number of key questions remain unanswered. Firstly, the intra- and interday reliability and precision of measurement of ultrasound imaging of the LM and TrA during dynamic activities is yet to be established. Thus, the magnitudes of change required for a confident determination of change in function are unknown at this time. Secondly, the typical nature of LM and TrA function during this mode of exercise is yet to be evaluated in relation to commonly used assessment techniques such as the abdominal drawing-in manoeuvre, active straight-leg raise, and contralateral arm-lift. Thirdly, functioning of the LM and TrA during this mode of exercise and other commonly used corrective/rehabilitative techniques based on relatively static challenges to stability is yet to be compared. Fourthly, the intrinsic kinematic stability of the lumbopelvic region whilst using the exercise device has not yet been examined, that might reveal further evidence of the underlying mechanisms facilitating LM and TrA contraction. And finally, the pattern and distribution of global muscle activation during this exercise also remains to be examined at this point.

CHAPTER III
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Lumbar Multifidus and *Transversus Abdominis* Thickness during Dynamic Activities – Reliability and Precision of Measurement of Ultrasound Images

3.1 Introduction

Recently there has been considerable use of USI in the assessment of *lumbar multifidus* (LM) and *transversus abdominis* (TrA), stemming from observations of functional deficits and morphologic changes of trunk musculature in individuals with low back pain (LBP) (Hebert et al., 2009). Typical USI assessment methods of LM and TrA include linear muscle thickness measurements during both resting and contraction conditions (Ferreira, Ferreira, & Hodges, 2004; Kiesel et al., 2007a; Whittaker et al., 2007). Thickness change (i.e. increases in linear muscle thickness from resting to contraction conditions) is also commonly used and has been demonstrated to closely reflect muscle activation at low levels (<30-40 % maximal voluntary contraction) in several muscles, such as *biceps brachii*, *brachialis*, *internal oblique*, LM and TrA (Hodges et al., 2003; Kiesel et al., 2007a; McMeeken et al., 2004). This makes USI a useful non-invasive alternative to fine-wire electromyography for assessing deep lumbopelvic muscle activity (Stokes, Henry, & Single, 2003).

The Functional Re-adaptive Exercise Device (FRED) has been developed with the intention to automatically recruit the local lumbopelvic muscles, specifically the LM and TrA (Debuse et al., 2013; Korfmacher, Debuse, & Pinotti, 2006). Previously FRED exercise has been demonstrated to recruit both the LM and TrA automatically in an asymptomatic population to a greater extent than simply standing on an unstable base of support (Debuse et al., 2013), suggesting a greater degree of motor control is required during the dynamic activity.

The use of ultrasound imaging (USI) in medical diagnostics has grown rapidly since the first pioneering work demonstrated the ability to visualize changes within living

tissues (Wild & Neal, 1951). During the following 60 years, medical applications for USI have expanded greatly and allowed for its widespread use in diagnostics concerning morphologic characteristics of numerous visceral organs and soft tissues (Szabo, 2004). Although not considered a gold standard in comparison to alternate imaging techniques such as magnetic resonance imaging and x-ray computed tomography, USI has nonetheless become a valuable tool for many healthcare professionals in the assessment of soft tissue and muscle morphology and function (Teyhen, 2006). Such application of USI can be seen as early as the 1980s when the technology was used to quantify the extent of quadriceps atrophy following knee immobilisation or injury (Young et al., 1980).

With respect to determining morphological aspects of muscle tissue, Hides, Richardson and Jull (1995) investigated the validity of USI derived measurements compared with those derived from MRI. Bilateral measurements of cross-section area were made at vertebral levels from L2 to S1 in healthy females. No significant differences were found between USI and MRI, despite the inherent differences in position for imaging (prone lying for RUSI and supine lying for MRI), when researchers adhered to a strict measurement protocol. Additionally, in a recent review Koppenhaver (2009) concluded that researchers can be confident of ultrasound measures of size (thickness/cross-section area) during most sub-maximal contractions.

To date, numerous studies have examined various aspects of reliability concerning USI and the assessment of LM and TrA (Hebert et al., 2009). These studies have used exercises such as the abdominal drawing in manoeuvre (ADIM) (Teyhen et al., 2005), active straight leg raise (Koppenhaver et al., 2009a; Teyhen et al., 2009b) and contralateral arm lifting (Kiesel et al., 2007a) to preferentially activate the deep

lumbopelvic musculature To our knowledge, however, only two of these has examined the reliability and precision of USI during a dynamic activity (Bunce, Moore, & Hough, 2002; Mangum et al., 2015), where LM and TrA were evaluated whilst the transducer was held in place with a custom made belt during treadmill walking. However, neither presented data for a freehand assessment method, and Bunce and Colleagues (2002) method has been questioned in subsequent reviews (Costa et al., 2009b; Hebert et al., 2009). No studies have reported the reliability of USI measurements of LM and TrA using a freehand technique during dynamic activities such as treadmill walking or when using the FRED. The purpose of this study was to determine the intra- and interday reliability and measurement precision of the assessment of both the LM and TrA using USI during a range of static and dynamic activities (including FRED exercise) when using a freehand method.

3.2 Method

3.2.1 Design

A random order test-retest design was used to investigate the intra- and interday reliability of hand-held ultrasound imaging of LM and TrA during selected control and dynamic activities (Figure **3-1**). Participants visited the laboratory on three separate occasions with each visit separated by three days (e.g. Tuesday, Friday, and Monday). All experimental conditions were assessed during each visit by a single assessor. Following a programme of formal training this assessor had approximately 12 months of experience upon commencement of this study. To avoid systematic order effects the conditions, with the exception of the rest condition which was captured first, were conducted in a counterbalanced order (Teyhen et al., 2011). Order was established using a custom permutation list and random rank ordering system.

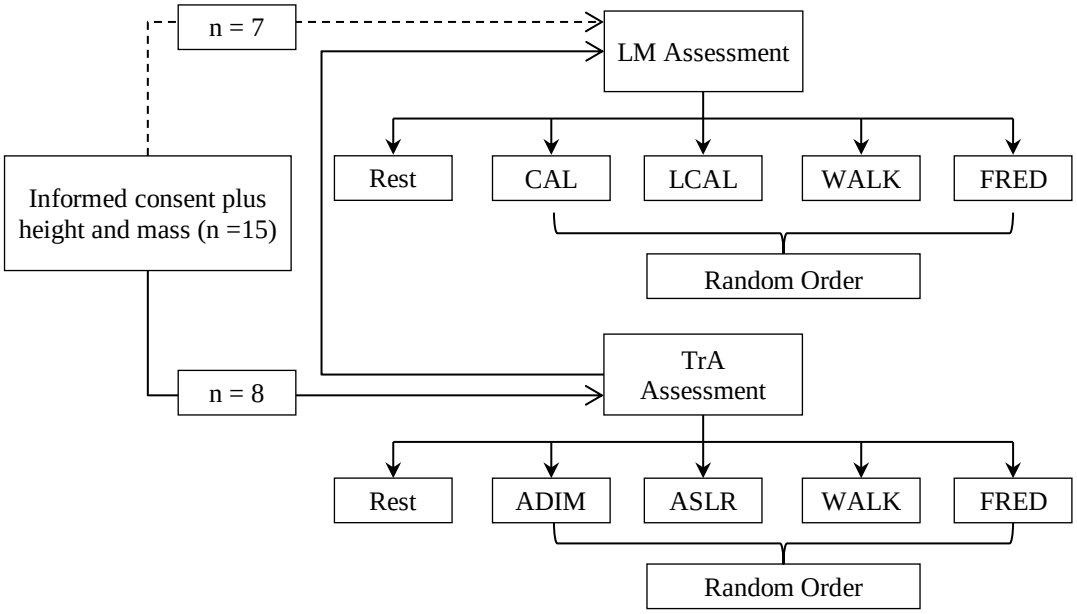

Figure 3-1. Schematic representation of reliability study experimental protocol design. CAL, contralateral arm lift; LCAL, loaded contralateral arm lift; ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise; WALK, treadmill walking; FRED, exercise device.

3.2.2 Participants

Fifteen healthy adults volunteered for this study (nine males; six females). Anthropometric characteristics of participants are shown in Table **3-1**. Participants were excluded if they had a history of LBP within the preceding six months, existing or previous musculoskeletal pathology/injury, any known neuromuscular or joint disease, previous abdominal or lumbar spine surgery, or were currently pregnant. Approval for this study was gained from the Ethics Committee of the School of Life Sciences at Northumbria University, Newcastle upon Tyne, England. All participants gave fully informed (Appendix H) written consent (Appendix I) to participate in this study.

Table 3-1. Anthropometric participant characteristics

	Height (m)	Mass (kg)	BMI ($kg \cdot m^{-2}$)	Age (years)
Female (n=6)				
Mean	1.73 ± 0.10	63.1 ± 5.1	21.2 ± 1.6	25.7 ± 1.6
Male (n=9)				
Mean	1.75 ± 0.03	81.6 ± 7.9	26.7 ± 3.1	29.7 ± 8.6
Combined (n=15)				
Mean	1.74 ± 0.07	74.2 ± 11.5	24.5 ± 3.8	28.1 ± 6.9

BMI denotes body mass index

3.2.3 Ultrasound Assessment of Lumbar Multifidus and Transversus Abdominis

A digital ultrasound imager (Technos MP, Esaote, Genoa, Italy) in B-mode was used by a single operator to collect images of the LM and TrA during each experimental condition. A 60-mm curvilinear transducer array (CA621, Esaote, Genoa, Italy) with a variable centre frequency of 2-7 MHz was used throughout. A fixed centre frequency of 5 MHz was chosen for both LM and TrA assessment in accordance with recommendations (Stokes et al., 2007) and previous literature (Hides, Cooper, & Stokes, 1992; McMeeken et al., 2004; Stokes, Rankin, & Newham, 2005). Images were optimised by manipulation of gain and digital processing parameters.

Water-soluble hypoallergenic ultrasound transmission gel (Aquasonic 100, Parker Laboratories Inc., Fairfield, New Jersey) was applied to the head of the transducer probe prior to placement onto the skin of the participant. This was repeated as and when necessary throughout each testing session. A free hand technique was used during image capture so as to minimise the impact of soft-tissue compression caused by inward pressure of the transducer (Whittaker et al., 2007). This also allowed the operator to make subtle corrections in transducer orientation, within the ranges of 9° internal/external rotation and 5° of cranial/caudal and medial/lateral tilting previously identified (Whittaker, Warner, & Stokes, 2009), to optimise image clarity.

Images of TrA were captured with the transducer head placed transversely on the antero-lateral abdominal wall superior to the iliac crest along the longitudinal midaxillary line with the muscle belly in the centre of the screen and the aponeurosis clearly visible (Teyhen et al., 2007). To control for the influence of food consumption on TrA measurements (Kordi et al., 2011) all participants were instructed to record the time of their last meal before the first visit and replicate this for subsequent visits. To control for the influence of respiration, all TrA images were captured at the end of relaxed exhalation where TrA thickness is at its greatest (Ainscough-Potts, Morrissey, & Critchley, 2006).

Images of LM were captured with the transducer head placed longitudinally along the spine, lateral of the L4 spinous process and orientated medially to identify the L4/5 facet joint (Hides, Cooper, & Stokes, 1992). The transducer was first placed longitudinally over the sacrum, in the mid-line, before slowly tracking vertically to produce a scan of the spinous processes, which resembled the 'Loch Ness Monster' (Stokes, Rankin and Newman, 2005). All images were captured unilaterally on the right hand side of the body in triplicate, and averaged to increase precision and reduce measurement error (Hebert et al., 2009). The transducer was removed and repositioned between consecutive acquisitions. Images were saved locally before being exported for offline analysis following completion of data collection.

Knowledge of local anatomy and transducer placement according to published literature ensured consistency of transducer positioning within and between trials (Hides, Cooper, & Stokes, 1992; Stokes et al., 2007; Strohl et al., 1981). Skin markings were not used as changes in posture were expected to cause movements of the skin,

thus altering their position relative to the underlying musculature (Ainscough-Potts, Morrissey, & Critchley, 2006; Coldron, Stokes, & Cook, 2003; Reeve & Dilley, 2009).

3.2.4 Experimental Protocol

During each visit, participants completed a battery of experimental conditions for assessment of both LM and TrA. *Lumbar multifidus* conditions were rest, unloaded contralateral arm lift (CAL), loaded contralateral arm lift (LCAL), treadmill walking (WALK), and FRED exercise in standing (FRED). *Transversus abdominis* conditions were rest, abdominal drawing-in manoeuvre (ADIM), active straight leg raise (ASLR), treadmill walking (WALK), and FRED exercise in standing (FRED).

For LM assessment during the resting condition participants laid in a prone position with pillows placed under the abdomen to reduce the lumbar/sacral junction to less than 10° so that the muscles lay as horizontally as possible along the spine (Kiesel et al., 2007a; Stokes et al., 2007). During the CAL condition participants laid prone as in the rest condition with their shoulder abducted 120° and their elbow flexed 90° and instructed to raise their arm approximately 5 cm off the examination couch. LCAL arm position and movement were as in CAL, however, participants held a weight of either 0.68 or 0.9 kg in their hand dependent on their body mass (Kiesel et al., 2007a). Individuals ≤79.5 kg held a weight of 0.68 kg and individuals >79.5 kg held a weight of 0.9 kg (Kiesel et al., 2007a).

For TrA assessment during the resting condition participants lay in a supine position with their hips and knees flexed to 50 and 90°, respectively. The ADIM was also performed with the participants lying supine. They were instructed to "take a relaxed breath in and out, hold the breath out, and then draw-in your lower abdomen without

moving your spine". Alternate cues of "cut off the flow of urine" or "close your rear passage" were provided if necessary to optimise preferential activation of TrA (Koppenhaver et al., 2009a). The active straight leg raise was performed with the participants lying supine with legs extended and feet approximately 20 cm apart. On command, participants were instructed to slowly raise the leg unilateral to the image site approximately 5 cm off the examination couch and hold this position for 10 seconds (Teyhen et al., 2009b).

The WALK condition was identical for assessment of both LM and TrA. Participants walked on a treadmill at a self-selected comfortable walking speed with images captured when their right foot was in its most anterior position (i.e. heel strike). Participants were blinded to the actual walking speed selected, but their walking speed was noted and replicated at subsequent visits. A digital metronome was set to match stride frequency of each participant between visits and provided an audible indicator to the operator for image capture.

For the assessment of LM and TrA in the FRED condition, participants were instructed to self-select a movement frequency that allowed them to achieve a smooth controlled movement with minimal cephalad/caudad excursion of the torso (Debuse et al., 2013). As with WALK, images were captured when the right foot was in its most anterior position in the cycle and a digital metronome matched to the movement frequency was used to provide an audible indicator to the operator.

3.2.5 Image Analysis & Blinding

All ultrasound images were processed offline using publicly available software (ImageJ, US National Institutes of Health, available at http://rsb.info.nih.gov/ij/).

Images were magnified 200 % to ensure consistent optimal visualisation of the echogenic fascial membranes (Bunce, Moore, & Hough, 2002). Images were analysed in random order to ensure blinding of the investigator as to the test condition, participant and previous values.

Linear measurements between the posteriormost portion of L4/5 facet joint and the thoracolumbar fascia (Figure **3-2**) were taken as LM muscle thickness (Hides, Cooper, & Stokes, 1992). Muscle thickness of the TrA was taken as the linear distance between the superficial and deep hyperechoic fasciae (Figure **3-3**), perpendicular to the muscle fibres, at a standardised distance of 15mm lateral from the aponeurosis (Reeve & Dilley, 2009). Thickness change was also calculated as a percentage increase from resting measurements for each experimental condition for both LM (Kiesel et al., 2007a) and TrA (Critchley & Coutts, 2002).

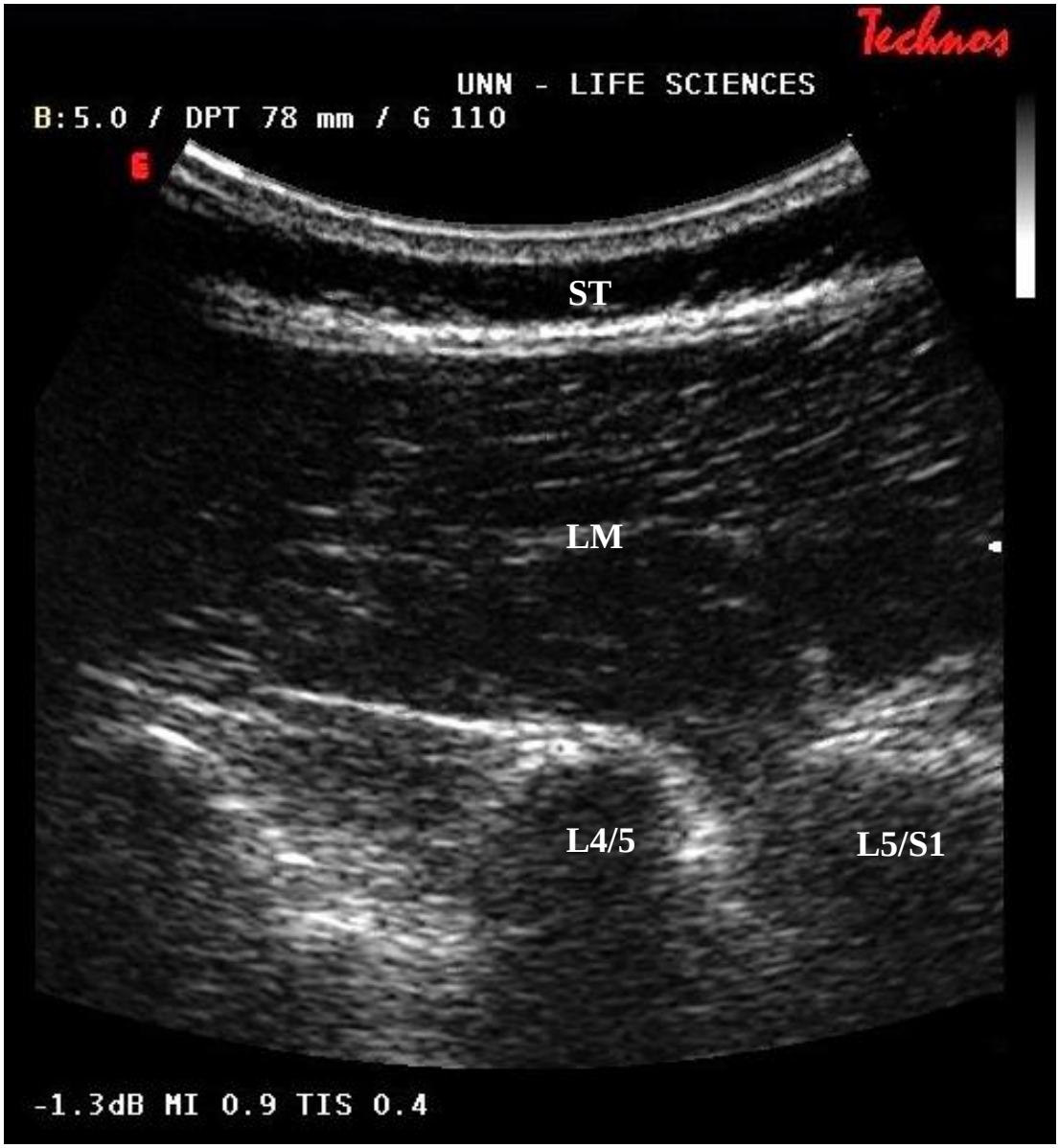

Figure 3-2. Exemplar captured ultrasound image of the longitudinal view of the lumbar vertebrae including the subcutaneous tissue (ST), *lumbar multifidus* muscle (LM), and the L4/5 and L5/S1 facet joints

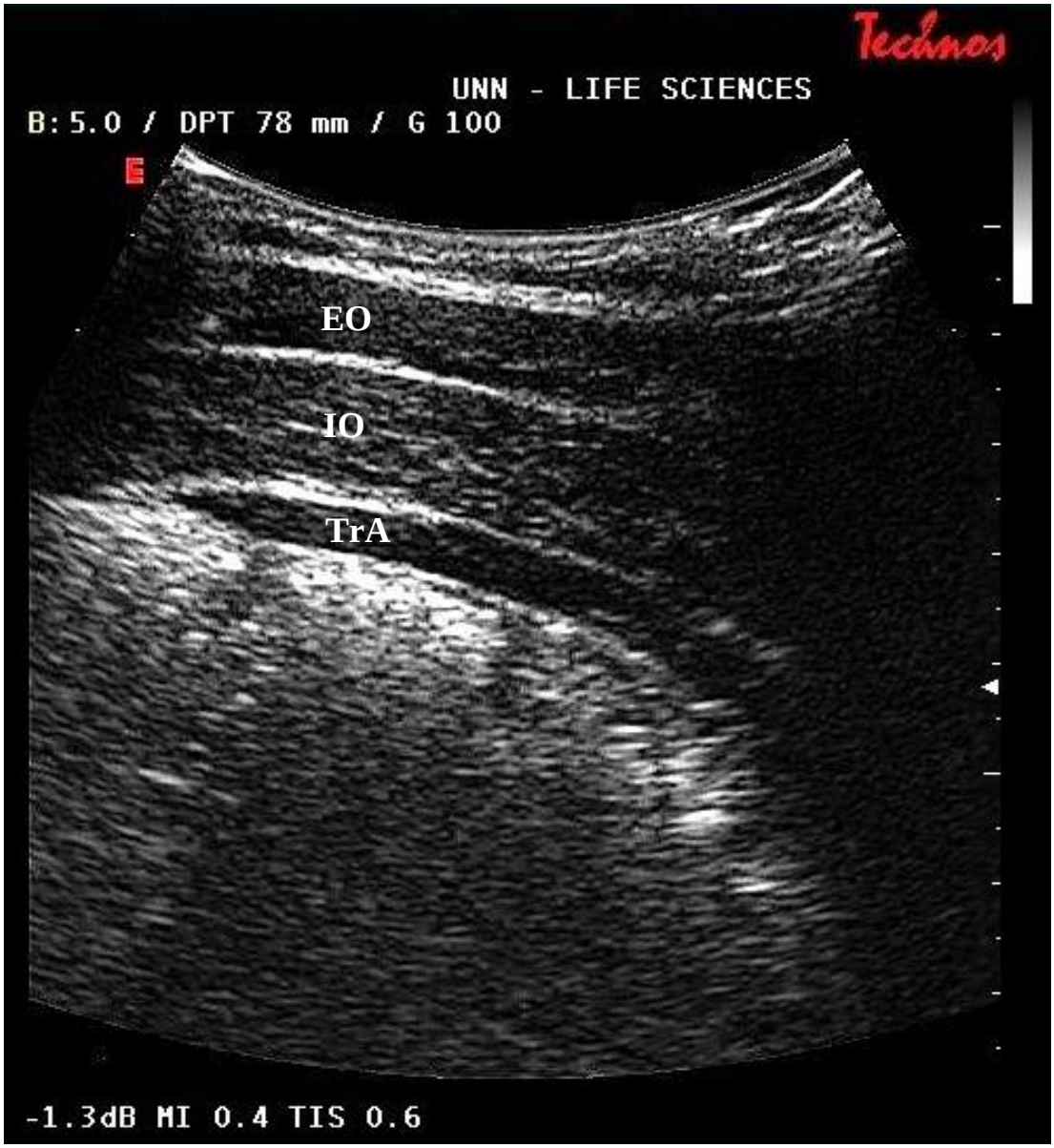

Figure 3-3. Exemplar captured ultrasound image of the anterolateral abdominal wall including the *external oblique* (EO), *internal oblique* (IO) and the *transversus abdominis* (TrA) muscles

3.2.6 Data Processing and Statistical Analysis

Two-way random effects intra-class correlation coefficients (ICC) of the three individual thickness measurements taken each day (ICC2,1) were calculated for estimation of intraday reliability of LM and TrA. Intraclass correlation coefficients were calculated separately for each day that participants attended. Interday reliability was assessed using two-way random effects ICC of thickness and thickness change

using the mean of three consecutive measurements (ICC2,3), where thickness change was given as:

$$\%Change = \left(\frac{(Contracted - Rest)}{Rest}\right) \cdot 100$$

Intraclass correlation coefficients were objectively interpreted in accordance with recommendations by both Portney and Watkins (2008) and Shrout and Fleiss (1979) where an ICC ≥ 0.9 was considered excellent, ≥ 0.75 was considered good, ≥ 0.5 was considered moderate and < 0.5 was considered poor.

Standard error of measurement (SEM) and minimum detectable change (MDC) were calculated for estimates of both intra- and interday precision of measurement for LM and TrA. Standard error of measurement was calculated as SD·√(1-ICC) and MDC was calculated as 1.96·(SEM·√2). Biases and 95 % limits of agreement (LOA) were also calculated for interday precision of measurement estimates as the mean of interday difference measurements on consecutive days ± 2SD.

One-way repeated measures analyses of variance (RM ANOVA) were used to identify any systematic order effects between days. The level of significance was set at 95 % ($p < 0.05$) for all data. Sphericity of data was assumed if Mauchly's Test was non-significant ($p > 0.05$). If this assumption was violated, adjustment was made using a Greenhouse-Geisser correction if $\varepsilon < 0.75$, alternatively a Huynh-Feldt correction was used if $\varepsilon \geq 0.75$ as recommended by Girden (1992). If the RM ANOVA identified significant interactions *post hoc* pairwise comparisons (LSD) were used to identify the location of significant differences between days.

All statistical analysis was performed within PASW Statistics v.18 (SPSS Inc., Chicago, Illinois). Summary raw data tables are available for lumbar multifidus and transversus abdominis in appendices J and K respectively.

3.3 Results

3.3.1 Intraday Reliability and Precision

Intraday reliability and precision of measurement estimates for all conditions for absolute linear muscle thickness on each of the three visits are presented in Table **3-2**. Intraday reliability estimates for LM and TrA absolute linear muscle thickness demonstrated good to excellent reliability with ICC values ranging from 0.83 to 0.97 and 0.89 to 0.97, respectively, for all conditions in each of the three images. Reliability estimates were typically lower for measurements taken during visit one in comparison to those observed on visits two and three, although generally consistent overall. Standard error of measurement for LM and TrA ranged from 1.2 to 3.8 mm and 0.3 to 0.7 mm, respectively. Minimum detectable change estimates for LM and TrA ranged from 3.2 to 10.5 mm and 0.9 to 1.8 mm across all conditions, respectively.

Table 3-2. Intraday reliability and precision of absolute linear muscle thickness using three consecutive individual measures for each assessed condition for each of the three visits

	$ICC_{2,1}$			SEM (mm)			MDC (mm)		
	Day			Day			Day		
Condition	1	2	3	1	2	3	1	2	3
LM									
Rest	0.97	0.95	0.96	1.2	1.3	1.3	3.2	3.6	3.7
CAL	0.88	0.92	0.93	2.8	2.2	2.2	7.8	6.0	6.0
LCAL	0.96	0.97	0.96	1.8	1.5	1.8	4.9	4.2	5.0
WALK	0.89	0.84	0.91	3.1	3.1	2.4	8.7	8.5	6.8
FRED	0.83	0.84	0.89	3.8	3.4	3.3	10.5	9.4	9.1
TrA									
Rest	0.96	0.97	0.96	0.4	0.3	0.3	1.0	0.9	0.9
ADIM	0.91	0.92	0.93	0.6	0.5	0.5	1.6	1.5	1.4
ASLR	0.96	0.97	0.97	0.5	0.3	0.4	1.3	1.0	1.0
WALK	0.89	0.93	0.93	0.7	0.5	0.6	1.8	1.5	1.5
FRED	0.97	0.97	0.95	0.4	0.4	0.5	1.2	1.2	1.4

Abbreviations: ICC, intraclass correlation coefficient; SEM, standard error of measurement; MDC, minimum detectable change; CAL, contralateral arm lift; LCAL, loaded contralateral arm lift; WALK, treadmill walking; FRED, exercise device; ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise;

Intraday reliability and precision of measurement estimates for all conditions for relative linear muscle thickness change on each of the three images are presented in Table **3-3**. Intraday reliability estimates for LM and TrA relative linear muscle thickness demonstrated moderate to excellent reliability with ICC values ranging from 0.59 to 0.95 and 0.52 to 0.97, respectively, for all conditions across the three images. Standard error of measurement for LM and TrA ranged from 2.8 to 7.6 % and 5.2 to 13.6 %, respectively. Minimum detectable change estimates for LM and TrA ranged from 7.9 to 21.0 % and 13.4 to 37.8 % across all conditions, respectively.

Table 3-3. Intraday reliability and precision of relative linear muscle thickness change using individual measures for each assessed condition for each of the three visits

	$ICC_{2,1}$			SEM (%)			MDC (%)		
	Day			Day			Day		
Condition	1	2	3	1	2	3	1	2	3
LM									
CAL	0.91	0.79	0.83	3.3	3.5	3.7	9.0	9.8	10.1
LCAL	0.91	0.95	0.90	3.8	2.8	3.5	10.6	7.9	9.6
WALK	0.80	0.89	0.88	6.3	6.1	5.1	17.6	16.9	14.2
FRED	0.59	0.83	0.73	7.6	6.3	6.5	21.0	17.4	18.1
TrA									
ADIM	0.67	0.80	0.62	13.6	8.3	8.3	37.8	23.1	23.0
ASLR	0.88	0.97	0.96	6.9	4.8	5.2	19.1	13.4	14.4
WALK	0.70	0.83	0.52	11.1	8.0	7.9	30.8	22.1	21.9
FRED	0.81	0.89	0.81	6.4	6.2	7.2	17.8	17.2	20.0

Abbreviations: ICC, intraclass correlation coefficient; CI, confidence interval; SEM, standard error of measurement; MDC, minimum detectable change; ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise; CAL, contralateral arm lift; LCAL, loaded contralateral arm lift; FRED, exercise device

3.3.2 Interday Reliability and Precision

Interday reliability and precision estimates for absolute linear muscle thickness LM and TrA are presented in Table **3-4**. Interday reliability estimates for LM and TrA absolute linear muscle thickness demonstrated excellent reliability, with ICC values ranging between 0.93 to 0.99 and 0.94 to 0.99, respectively. Standard error of measurement between days for LM and TrA ranged from 1.2 to 2.3 mm and 0.2 to 0.5 mm, respectively. Minimum detectable change estimates between days ranged from 3.4 to 6.5 mm and 0.2 to 1.4 mm, respectively.

Table 3-4. Interday reliability and precision of absolute linear muscle thickness using a mean of three measures for each assessed condition for each of the three visits.

	$ICC_{2,3}$			Bias (95 CI) ± 95 % LOA		SEM (mm)		MDC (mm)	
Condition	All	D1-D2	D2-D3	D1-D2	D2-D3	D1-D2	D2-D3	D1-D2	D2-D3
LM									
Rest	0.99	0.96	0.96	-0.4 (-1.4-0.6) ± 3.5	0.4 (-0.5-1.4) ± 3.4	1.2	1.2	3.4	3.4
CAL	0.93	0.95	0.93	0.4 (-0.3-1.0) ± 2.3	-0.8 (-1.7-0.1) ± 3.3	1.8	2.1	4.9	5.9
LCAL	0.98	0.96	0.96	-0.1 (-0.6-0.4) ± 1.8	0.8 (0.4-1.3) ± 1.6	1.8	1.8	5.0	4.9
WALK	0.94	0.93	0.93	-1.7 (-3.4-0.1) ± 6.0	-1.9 (-3.7-0.0) ± 6.4	2.3	2.1	6.2	5.8
FRED	0.95	0.93	0.94	2.4 (0.9-3.9) ± 1.4	0.8 (-0.7-2.2) ± 5.1	2.3	2.3	6.5	6.2
TrA									
Rest	0.99	0.99	0.99	0.1 (0.0-0.3) ± 0.5	-0.2 (-0.3-0.0) ± 0.6	0.2	0.2	0.5	0.5
ADIM	0.98	0.97	0.98	0.1 (-0.2-0.5) ± 1.4	0.0 (-0.3-0.4) ± 1.1	0.3	0.3	0.9	0.8
ASLR	0.98	0.97	0.99	-0.1 (-0.4-0.2) ± 1.0	0.2 (0.1-0.3) ± 0.4	0.4	0.2	1.0	0.6
WALK	0.94	0.97	0.94	-0.1 (-0.6-0.3) ± 1.5	0.0 (-0.3-0.3) ± 1.0	0.3	0.5	0.9	1.4
FRED	0.98	0.98	0.98	0.1 (-0.2-0.3) ± 0.2	0.1 (-0.1-0.3) ± 0.7	0.4	0.3	1.0	0.9

Abbreviations: ICC, intraclass correlation coefficient; CI, confidence interval; LOA, limits of agreement; SEM, standard error of measurement; MDC, minimum detectable change; D1-3, Day 1-3, ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise; CAL, contralateral arm lift; FRED, exercise device

Interday reliability and precision of measurement estimates for all conditions for relative linear muscle thickness change on each of the three visits are presented in Table **3-5**. Intraday reliability estimates for LM and TrA for relative linear muscle thickness change demonstrated good to excellent reliability with ICC values ranging from 0.79 to 0.90 and 0.79 to 0.90, respectively, for all conditions across the three visits. Standard error of measurement for LM and TrA ranged from 4.3 to 6.4 % and 7.4 to 10.1 %, respectively, between days 1 and 2. Standard error of measurement reduced for both LM and TrA between days 2 and 3 to between 3.9 to 5.3 % and 3.6 to 8.7 %, respectively. Minimum detectable change estimates for LM and TrA ranged from 11.8 to 17.7 % and 20.4 to 28.1 % across all conditions between days 1 and 2, respectively, reducing to between 9.5 to 14.8 % and 9.9 to 24.2 % between days 2 and 3, respectively.

Table 3-5. Interday reliability and precision of linear muscle thickness change (normalised to resting thickness) using a mean of three measures for each assessed condition for each of the three visits

	$ICC_{2,3}$			Bias (95 CI) ± 95 % LOA		SEM (%)		MDC (%)	
Condition	All	D1-D2	D2-D3	D1-D2	D2-D3	D1-D2	D2-D3	D1-D2	D2-D3
LM									
CAL	0.79	0.77	0.76	2.4 (-1.9-6.8) ± 15.5	-4.6 (-8.7--0.6) ± 14.4	4.3	3.9	12.0	10.8
LCAL	0.90	0.89	0.82	1.0 (-3.4-5.5) ± 15.8	1.0 (-4.2-6.3) ± 18.6	4.3	3.4	11.8	9.5
WALK	0.84	0.85	0.82	-3.2 (-9.8-3.4) ± 23.3	-9.0 (-17.0--0.7) ± 29.5	6.4	5.1	17.7	14.0
FRED	0.84	0.78	0.79	10.6 (4.2--7.2) ± 22.8	-0.2 (-7.7-7.2) ± 26.2	6.4	5.3	17.7	14.8
TrA									
ADIM	0.87	0.82	0.89	-1.1 (-8.7-6.55) ± 27.0	4.7 (-0.8-10) ± 19.8	7.5	4.6	20.7	12.7
ASLR	0.9	0.84	0.91	-4.7 (-12.0-2.8) ± 26.5	7.5 (3.0-11.9) ± 15.8	9.3	4.1	25.8	11.3
WALK	0.79	0.74	0.77	-8.4 (-19-1.6) ± 35.7	3.2 (-4.5-10.8) ± 27.1	10.1	8.7	28.1	24.2
FRED	0.88	0.81	0.93	-3.1 (-11.0-4.4) ± 26.5	7.5 (2.4-13) ± 18.2	7.4	3.6	20.4	9.9

Abbreviations: ICC, intraclass correlation coefficient; CI, confidence interval; LOA, limits of agreement; SEM, standard error of measurement; MDC, minimum detectable change; D1-3, Day 1-3, ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise; CAL, contralateral arm lift; FRED, exercise device

3.3.3 Systematic Order Effects

No interaction effects were found between days for absolute LM thickness in either Rest or CAL. Interaction effects for absolute LM thickness were found between days in LCAL ($F_{2,28} = 7.068$; $p = 0.003$), WALK ($F_{2,28} = 8.932$; $p = 0.001$), and FRED ($F_{2,28} = 0.858$; $p = 0.435$). No interaction effects were found between days for LM thickness change in either CAL or LCAL. Interaction effects for LM thickness change measurements between days were found in WALK ($F_{2,28} = 6.368$; $p = 0.005$) and FRED ($F_{2,28} = 8.628$; $p = 0.001$). Locations of significant differences identified from *post hoc* pairwise comparisons are illustrated in Figure **3-4**.

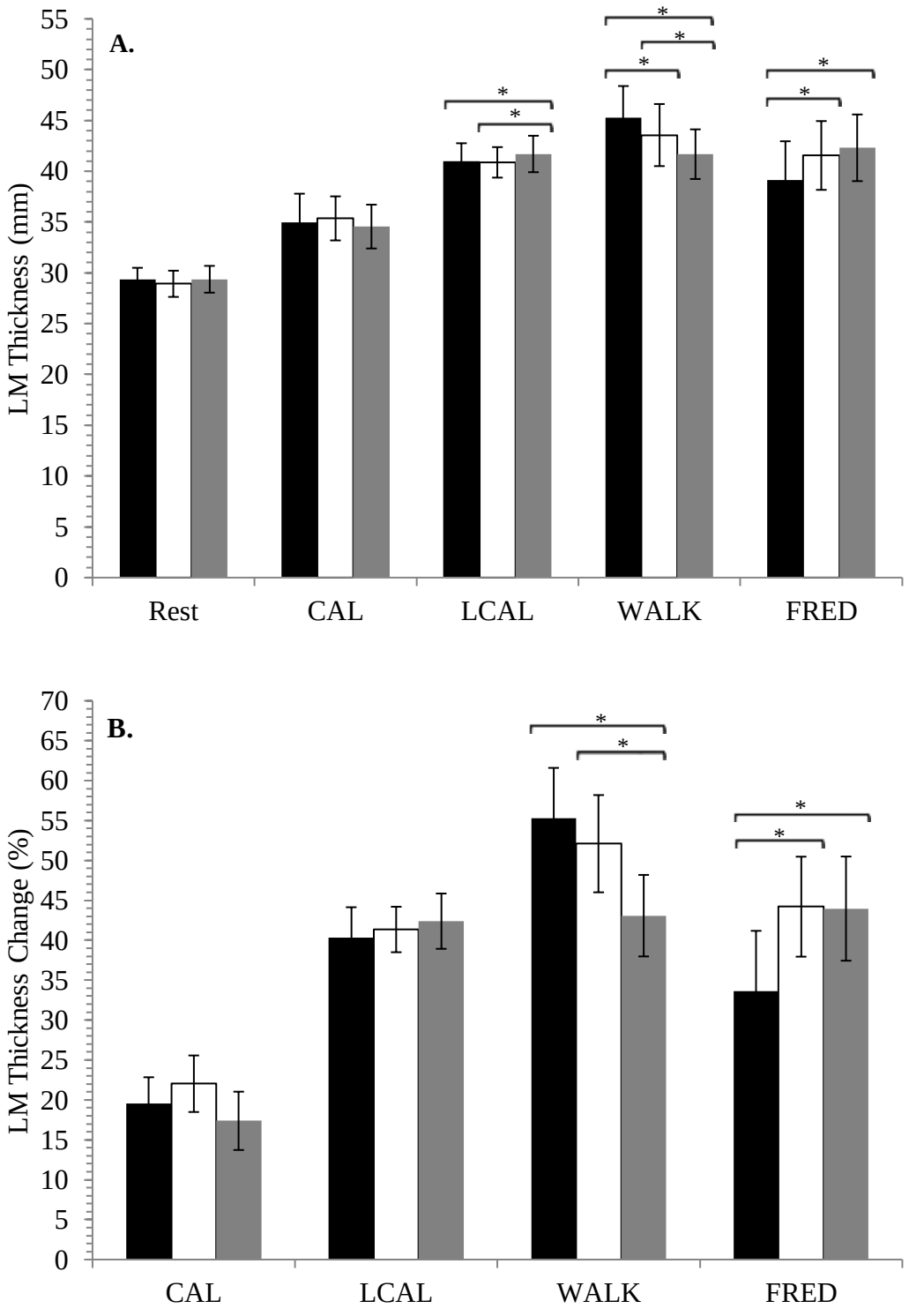

Figure 3-4. Absolute thickness (A) and percentage thickness change relative to resting thickness (B) of the *lumbar multifidus* (LM) during each experimental condition (CAL, contralateral arm lift; LCAL, contralateral arm lift with external load; WALK, treadmill walking; FRED, exercise device) across days one (black bars), two (white bars) and three (grey bars). Error bars indicate intraday standard error of measurement

No interaction effects were identified between days for any absolute TrA thickness measurements in any conditions. No interaction effects were found between days for TrA thickness change in ADIM, WALK, or FRED. An interaction effect for TrA thickness change measurements between days was found in the ASLR ($F_{2,28}$ = 3.833; p = 0.034) and FRED ($F_{2,28}$ = 8.628; p = 0.001). Locations of significant differences identified from *post-hoc* pairwise comparisons are illustrated in Figure **3-5**.

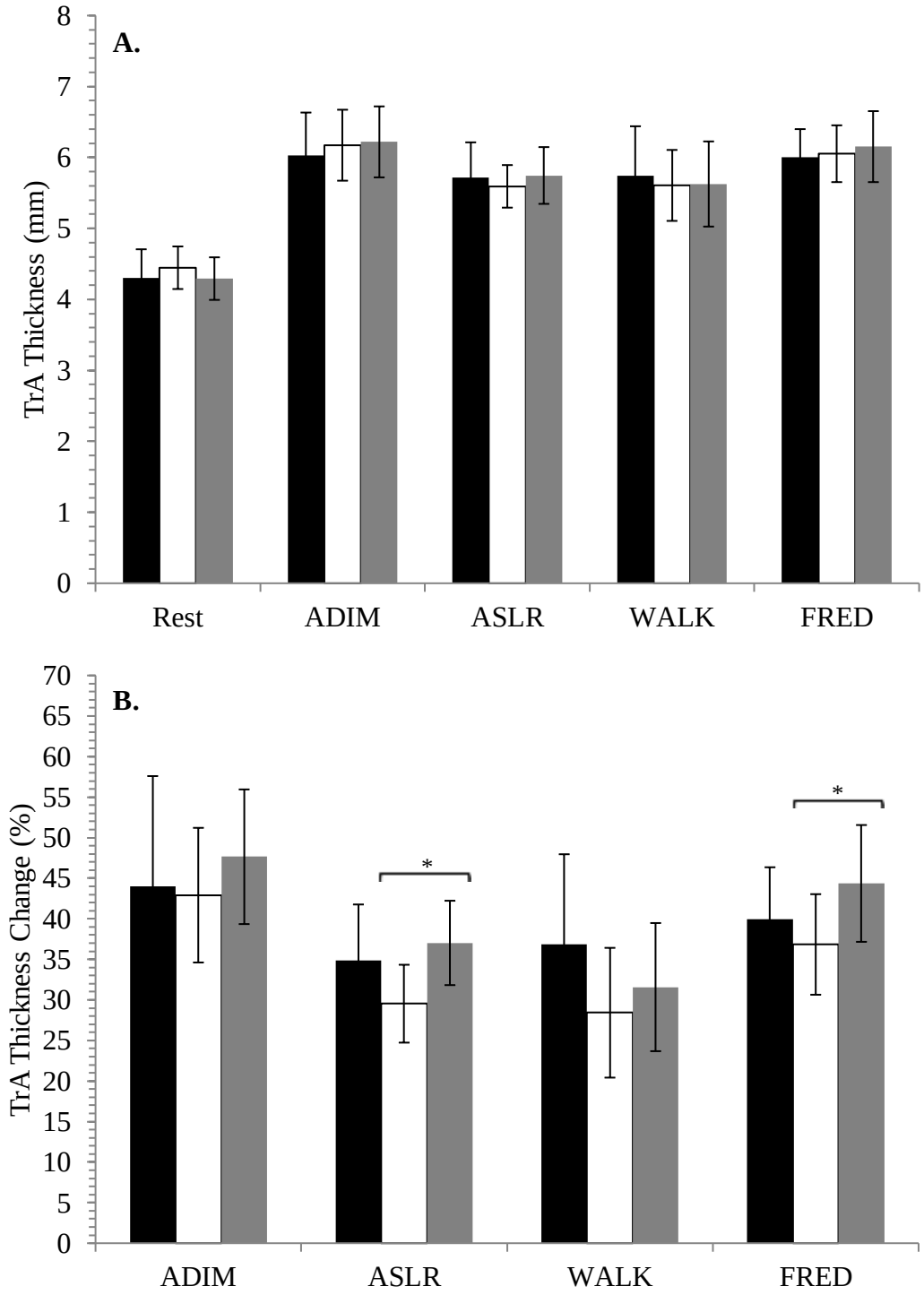

Figure 3-5. Absolute thickness (A) and percentage thickness change relative to resting thickness (B) of the *transversus abdominis* (TrA) during each experimental condition (ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise; WALK, treadmill walking; FRED, exercise device) across days one (black bars), two (white bars) and three (grey bars). Error bars indicate intraday standard error of measurement

3.4 Discussion

This study aimed to investigate the intra- and interday reliability and precision of linear thickness and thickness change measurements of the LM and TrA during dynamic and control activities using a freehand USI technique. The key findings of this study were that LM and TrA typically demonstrated good (ICC ≥ 0.75) to excellent (ICC ≥ 0.9) intrarater reliability for both intra- and interday measurements of absolute linear muscle thickness across all conditions. Normalised thickness change, expressed relative to resting values, also demonstrated good reliability between days with ICCs in excess of 0.75 across all conditions. One further observation was that of consistent reductions between days one-two and two-three in both standard error of measurement (range [4.3-10.1 % vs. 3.4-8.7 %], respectively) and minimum detectable change (range [11.8-28.1 % vs. 9.5-24.3 %], respectively).

3.4.1 Reliability and Precision of Ultrasound Imaging during Static Conditions

Upper limb movement during standing has previously been shown to actively elicit recruitment of the deep and superficial fibres of the LM, contralateral to the movement (Moseley, Hodges, & Gandevia, 2002), a finding also consistent with prone lying (Kiesel et al., 2007a) in asymptomatic individuals. Kiesel et al. (2007a) also examined the effect of external load during upper limb movement and noted a contraction intensity of approximately 30 % of maximal contraction capacity during loaded contralateral arm lifts. A number of studies have since implemented loaded contralateral arm lifts as a method of assessing functional changes in LM muscle thickness in asymptomatic individuals (Larivière et al., 2013; Sions et al., 2014; Sweeney, O'Sullivan, & Kelly, 2014; Teyhen et al., 2012; Teyhen et al., 2011) as well as individuals with low back pain (Hebert et al., 2010; Koppenhaver et al., 2011;

Koppenhaver et al., 2009a; Larivière et al., 2013; Sweeney, O'Sullivan, & Kelly, 2014; Zielinski et al., 2013).

The single-rater intraday reliability estimates for absolute thickness measurements presented within the current study are consistent with those studies including an asymptomatic sample group. Larivière et al. (2013) reported an intraclass correlation coefficient (ICC) and standard error of measurement (SEM) of 0.94 and 1.5 mm respectively, corresponding closely with the values obtained across all three days in this study (ICC day one = 0.96, day two = 0.97, day three = 0.96; SEM day one = 1.8, day two = 1.5, day three = 1.8). In terms of thickness change expressed relative to resting measurements, however, Larivière et al. (2013) reported significantly lower reliability estimates (ICC = 0.61) than those found here (ICC = 0.90), although precision estimates were largely similar (SEM = 5.8 % vs 4.3 %). Unfortunately, the authors did not directly report minimum detectable change (MDC) values for either absolute or relative thickness change. However, given the relationship between SEM and MDC [$MDC = 1.96 \cdot (SEM \cdot \sqrt{2})$] it is likely that there would be similar agreement for MDC with the current data.

There is no published literature reporting interday reliability of absolute and relative thickness change of the LM muscle in asymptomatic individuals. In comparison to the findings of Koppenhaver et al. (2009a), however, general consistency with this study is again demonstrated in terms of ICC, SEM and MDC for both absolute (0.97, 1.1 mm, and 3.1 mm vs. 0.98, 1.8 mm, and 5.0mm) and relative (0.79, 4.0 %, and 11 % vs. 0.9, 4.3 % and 11.8 %) thickness changes of the LM during loaded contralateral arm lifts.

Intraday reliability estimates for absolute TrA thickness measurements during the ADIM were consistent with previous literature. Koppenhaver et al. (2009b) reported ICCs greater than 0.9, as in the current study. Their ICC was slightly higher than that in the current study (ICC = 0.97) which was reflected in the reduced SEM and MDC reported. Hides et al. (2007b) reported a lower ICC of 0.8, which could be explained by their use of a novice rater that was newly trained in the use of USI for the assessment of TrA muscle thickness. For relative TrA muscle thickness, Koppenhaver et al. (2009a) reported excellent reliability (based on ICC) compared to the moderate to good reliability apparent in the current data.

Reliability estimates for absolute TrA thickness during the ASLR (ICC = 0.96-0.97) were in line with previous studies, with both Teyhen et al. (2009b) and Koppenhaver et al. (2009b) reporting ICCs of 0.96. For relative TrA thickness during the ASLR, excellent reliability was observed which was in line with previous reports (Koppenhaver et al., 2009a), although Koppenhaver et al. (2009a) reported a higher SEM.

Koppenhaver et al. (2009b) and Hides et al. (2007b) reported interday reliability estimates for absolute TrA muscle thickness. During the ADIM, the present data showed better reliability than both these studies, with greater ICC and lower SEM. Similarly during the ASLR, an increased ICC and reduced SEM was found in the current study compared to Koppenhaver et al. (2009b).

3.4.2 Reliability and Precision of Ultrasound Imaging during Dynamic Movements

As walking is arguably one of the most common functional activities of daily living it is surprising that, to date, only one study has explored the use of ultrasound imaging in this context (Bunce, Moore, & Hough, 2002). Here, the authors examined TrA muscle function during treadmill walking in asymptomatic participants whilst using a custom-built belt to secure the transducer in place, thus allowing hands-free gathering of ultrasound images. This method has similarly been employed elsewhere, for example in the investigation of gastrocnemius muscle function (Fukunaga et al., 2001). The larger spatial excursions of the lower limbs arguably necessitate the use of a fixed transducer holder in this instance. However, the lumbar spine and abdominal wall do not typically experience such extremes of motion, in comparison.

Intraday reliability estimates of absolute linear thickness of the TrA, when performing USI freehand (i.e. without the use of a belt to secure the transducer in place), showed good to excellent reliability across all three days during both treadmill walking (ICC day one = 0.89; day two = 0.93; day three = 0.93) and during exercise using the device (ICC day one = 0.97; day two = 0.97; day three = 0.95). Precision estimates are also generally consistent across all three days during both treadmill walking (SEM and MDC day = 0.7 mm and 1.8 mm; day two = 0.5 mm and 1.5 mm; day three = 0.6 mm and 1.5 mm) and during exercise using the device (SEM and MDC day = 0.4 mm and 1.2 mm; day two = 0.4 mm and 1.2 mm; day three = 0.5 mm and 1.4 mm).

Bunce and colleagues (2002) reported marginally lower reliability estimates for treadmill walking (ICC = 0.88), alongside precision estimates (SEM = 0.56 mm) consistent with those observed in the current investigation. Notably, both the reliability and precision estimates for the TrA were superior during FRED exercise compared to

those observed during treadmill walking. This may be a consequence of the chosen image acquisition point (heel strike) for WALK, namely the propagation of impact forces (Jonsson, 1970) and associated axial rotation (Thorstensson et al., 1982) - a feature not present in FRED due to the absence of impact forces and more controlled motion.

To date, this is the only study to include ultrasound thickness measurements of the LM during dynamic activities. Absolute intraday reliability was good across all three days during WALK (ICC day one = 0.89; day two = 0.84; day three = 0.91) and FRED (ICC day one = 0.83; day two = 0.84; day three = 0.89). Expectedly, precision estimates were larger during both WALK (SEM and MDC day one = 3.1 mm and 8.7 mm; day two = 3.1 and 8.5 mm; day three = 2.4 mm and 6.8 mm) and FRED (SEM and MDC day one = 3.8 mm and 10.5 mm; day two = 3.4 and 9.4 mm; day three = 3.3 mm and 9.1 mm) in comparison to the control conditions (Rest, CAL and LCAL), where physical movement is much more restricted.

Relative thickness changes are arguably the most relevant for assessment of change in functioning across time. However, these measures incorporate the error associated with both resting and contracted measurements (Koppenhaver et al., 2009a). It is not surprising, therefore, that when expressed in such a manner, relative intraday LM thickness changes typically demonstrate reduced reliability estimates during both WALK (ICC = 0.80) and FRED (ICC = 0.59). However, this difference results in a relative SEM difference of only 1.3% between the two conditions. Considering the relationship between ICC and standard deviation in the determination of SEM, this suggests the data for the FRED condition were more homogenous than that of the WALK condition.

3.4.3 Systematic Order Effects

No systematic effects were observed for any condition when assessing absolute TrA muscle thickness that would suggest any learning effect on the part of the imager or participants. For LM, however, some systematic effects were observed, especially in the dynamic conditions of WALK and FRED. For walking, a systematic reduction was seen in absolute and relative LM thickness across the three days. It is unlikely that this was due to a learning effect of the participants, as walking was a routine activity for them. These systematic reductions could therefore be due to a learning effect of the imager. It should be noted, however, that the interday reliability was shown to be excellent for walking. In FRED exercise, an increase in muscle thickness (absolute and relative) was observed consistently between days one and two, but no change was seen between days two and three. This suggests that the participants became familiarised with FRED exercise during the first session, which could have caused greater activation of LM on the second and third days. In both instances, however, the magnitudes of change observed were both within the respective MDC thresholds and, thus, in practice, would not be considered as a true change.

3.5 Limitations

This study took measurements of TrA and LM muscle thickness from a relatively small sample of healthy individuals. In symptomatic individuals, it can be more difficult to obtain reliable measurements of muscle thickness during contraction due to the altered motor control seen and the difficulty that symptomatic participants can have in recruiting TrA and LM (Richardson & Hides, 2004; Richardson & Jull, 1995; Van, Hides, & Richardson, 2006). The reliability estimates presented here are for a single imager, limiting the generalisability of the findings to the wider group of USI users.

However, this is the first study to have reported on the intra- and interday reliability of USI using a freehand technique in dynamic conditions. It also took measurements on three separate days to determine interday reliability. Other studies have typically taken measurements over only two days (Hides et al., 2007b).

Furthermore, this study presents the results from a relatively low sample size of only 15 participants. Typically, a minimum sample size of 20 participants is recommended for the purposes of a reliability study (Atkinson and Nevill, 2001). This should again be considered when attempting to generalise the results of this study.

3.6 Conclusion

Intraday reliability was found to be moderate to excellent for a range of dynamic and control conditions for both absolute and relative thickness measurement of LM and TrA. Minimum detectable change in LM and TrA absolute muscle thickness measurements within-day was lower than for relative muscle thickness measurements. Interday reliability was found to be good to excellent across all conditions for both absolute and relative thickness measurements. Minimum detectable change between days was also found to be lower for absolute than for relative muscle thickness measurements. These findings support the use of freehand USI for the assessment of lumbopelvic muscle thickness during dynamic activities such as treadmill walking and FRED exercise. The minimum detectable change values reported also provide a useful reference for use in future studies investigating lumbopelvic muscle activity using USI.

CHAPTER IV

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FRED Exercise vs. Overground Walking: An Ultrasound Assessment of Muscle Function

4.1 Introduction

Exercise on the FRED which is the subject of this thesis (Debuse et al., 2013; Korfmacher, Debuse, & Pinotti, 2006) is hypothesised to recruit LM and TrA. However, as of yet this muscle recruitment has not been studied in relation to common clinical tests or walking as a fundamental activity of daily living. Having established the reliability of freehand USI to assess TrA and LM thickness in a range of activities in the previous chapter, the aim of this chapter is to examine the contraction of LM and TrA during general clinical tests and dynamic movement in walking and FRED exercise using USI.

Acute, recurrent and chronic LBP are increasingly associated with changes in both the function and morphology of the deep muscles of the trunk including both the *lumbar multifidus* (LM) and the *transversus abdominis* (TrA). Such observations include atrophy of the LM at multiple vertebral levels (Danneels et al., 2000; Hides et al., 2008; Hides et al., 1994; Kjaer et al., 2007), reduced activity of the LM (Kiesel et al., 2007b; MacDonald, Moseley, & Hodges, 2010; Sihvonen et al., 1997) and TrA (Ferreira, Ferreira, & Hodges, 2004), delayed activity of the LM (MacDonald, Moseley, & Hodges, 2009) and TrA (Hodges & Richardson, 1996), and a shift from tonic to phasic activation (Saunders, Coppieters, & Hodges, 2004) of the TrA.

Key clinical tests of TrA function include the abdominal drawing-in manoeuvre (ADIM) (Hodges, Richardson, & Jull, 1996; Teyhen et al., 2005), and the active straight leg raise (ASLR) (Richardson et al., 2002; Teyhen et al., 2009b), both of which are used to assess the activity of this deep local abdominal muscle that is known to augment intersegmental spinal stability by increasing intra-abdominal pressure

(Cresswell, Grundström, & Thorstensson, 1992; Hodges et al., 2001; Stokes, Gardner-Morse, & Henry, 2010) and tension on the thoracolumbar fascia (Barker et al., 2006).

Function of the LM is commonly assessed by a more subjective approach involving a conscious isometric contraction of the lumbar paraspinal muscles that can be assessed by either manual palpation or USI (Hides et al., 2000; Hides et al., 2011b; Van, Hides, & Richardson, 2006; Wallwork et al., 2009). Given that approximately 70 % of individuals are either 'unable' to consciously produce this contraction or the resulting contraction is 'poor' (Hides et al., 2011b), the recently proposed contralateral arm lift (CAL) (Kiesel et al., 2007a) has provided an alternative means to assess LM muscle function without the reliance on potentially troublesome conscious/voluntary contraction. Additionally, a variation of the CAL that includes an external load (LCAL) gives a potential indirect measure of muscle function relative to maximum voluntary contractions as measured by intramuscular electromyography, at least over a small range of submaximal intensities (Kiesel et al., 2007a).

One approach for physiotherapists involved in delivering therapeutic exercise for low back pain that has been the focus of continued experimental investigation is motor control training (MCT) (Hides et al., 2010; Richardson, Hodges, & Hides, 2004; Richardson et al., 1999). Motor control training involves 'teaching' individuals to voluntarily and isometrically contract the LM and TrA, firstly independent of each other in lying before progressing to sitting and standing with co-contraction of LM and TrA over a period of six weeks (Hides et al., 2008). The effectiveness of MCT in chronic non-specific LBP was recently examined in a systematic review (Macedo et al., 2009) which concluded that MCT was more effective than minimum intervention and beneficial when supplemented with additional therapies. Given that LBP has

previously been shown to negatively impact on lumbopelvic coordination during dynamic activities such as walking and running (Seay, Van Emmerik, & Hamill, 2011; van der Hulst et al., 2010) it is surprising that such functional actions are not fully integrated within MCT programmes despite propositions that they should be (Hodges & Cholewicki, 2007; Hodges et al., 2013; Richardson, Hodges, & Hides, 2004).

4.2 Method

All data presented in this chapter were collected using the protocol that is described in the previous chapter (Chapter III). The data from the first visit of participants were used. It was noted in this chapter that the ICC, SEM and MDC of the majority of measurements taken improved on subsequent days, however current practice does not typically involve familiarisation visits. As such, day one was chosen to more closely reflect real-world practice.

Ultrasound images were taken of the LM and TrA as participants performed a series of exercises, both static and dynamic in nature. For assessment of the LM the conditions included Rest, CAL, LCAL, WALK, and FRED, whilst for assessment of the TrA conditions were Rest, ADIM, ASLR, WALK and FRED.

Please see Chapter III for a full description of the procedures used.

4.2.1 Statistical Analysis

Differences in linear thickness change of the LM and TrA between conditions were assessed using one-way repeated measures analyses of variance (RM ANOVA). The level of significance was set at 95 % ($p < 0.05$) for all data. Sphericity of data was

assumed if Mauchly's Test was non-significant ($p > 0.05$). If this assumption was violated, adjustment was made using a Greenhouse-Geisser correction if $\varepsilon < 0.75$, alternatively a Huynh-Feldt correction was used if $\varepsilon \geq 0.75$ as recommended by Girden (1992). In the presence of a significant interaction of condition revealed by the RM ANOVA *post hoc* pairwise comparisons (LSD) were used to identify the location of significant differences between conditions.

All statistical analyses were performed within PASW Statistics v.18 (SPSS Inc., Chicago, Illinois). As per Chapter III, summary raw data tables are available for lumbar multifidus and transversus abdominis in appendices J and K respectively.

4.3 Results

The pattern of thickness change expressed relative to resting values of the LM was that it was greatest in the WALK condition (55.26 ± 14.36), followed by the LCAL (40.32 ± 12.81), FRED (33.62 ± 14.85) and CAL conditions (19.59 ± 10.30). A significant main effect of contraction condition (CAL, LCAL, WALK and FRED) was observed ($F_{3,42} = 41.000$; $p < 0.001$) with all conditions significantly different ($p \leq 0.001$) from one another (Table **4-1** and Figure **4-1**). No adjustments were necessary, as Mauchly's test was non-significant ($p > 0.05$).

Table 4-1. Individual pairwise comparisons between contraction conditions for *lumbar multifidus* thickness change expressed relative to resting/relaxed measurement. Units are percentage change (±SD).

	Mean1 (±SD)	Mean2 (±SD)	Mean Difference (± 95CI)	P value
CAL-LCAL	19.59 (±10.30)	40.32 (±12.81)	20.73 (13.79 to 27.67)	<0.001*
CAL-WALK		55.26 (±14.36)	35.67 (26.90 to 44.44)	<0.001*
CAL-FRED		33.62 (±14.85)	14.03 (6.95 to 21.11)	0.001*
LCAL-WALK	40.32 (±12.81)	55.26 (±14.36)	14.94 (7.50 to 22.38)	0.001*
LCAL-FRED		33.62 (±14.85)	-6.70 (-9.70 to -3.70)	<0.001*
WALK-FRED	55.26 (±14.36)	33.62 (±14.85)	-21.64 (-28.56 to -14.72)	<0.001*

CAL, contralateral arm lift; LCAL, loaded contralateral arm lift; WALK, treadmill walking; FRED, exercise device; SD, standard deviation; CI, confidence interval.

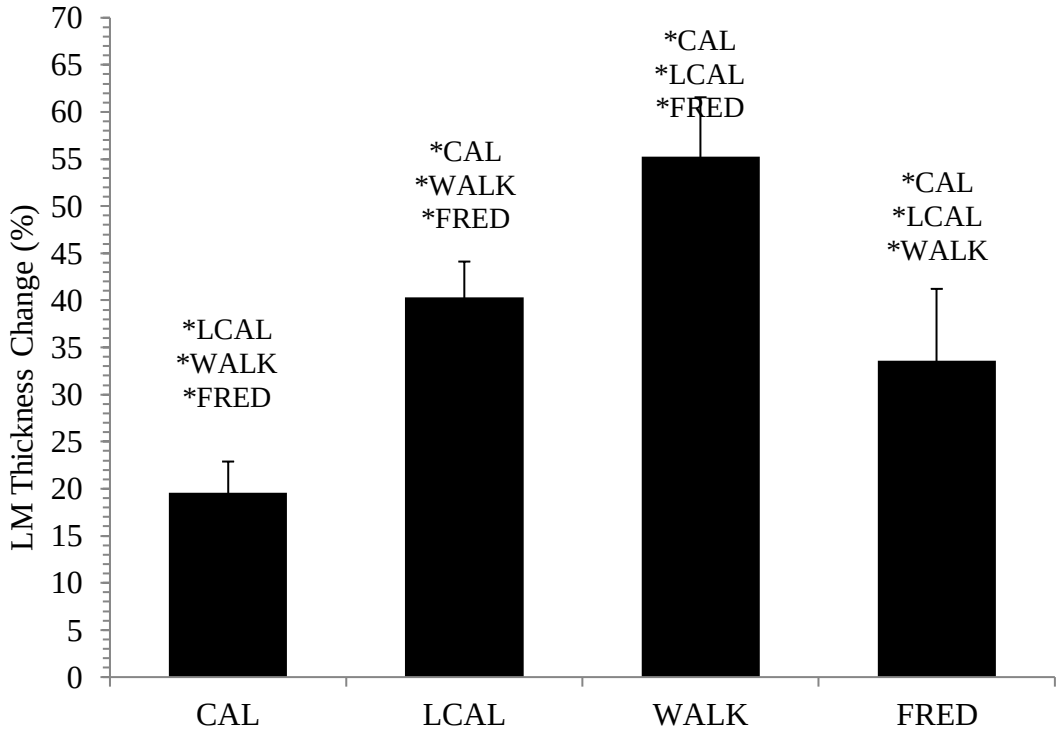

Figure 4-1. Change in muscle thickness of the *lumbar multifidus* at the L4/5 vertebral level, expressed relative to resting thickness for each of the experimental conditions (CAL-contralateral arm lift; LCAL-loaded contralateral arm lift; WALK-treadmill walking; FRED-exercise device) *denotes pairwise significant difference between that particular condition and each of the named conditions. Error bars represent the standard error of the measurement between days one and two as identified previously in chapter three.

The pattern of thickness change expressed relative to resting values of the TrA was that no condition was significantly different ($p > 0.05$) from the other, with no interaction effect of contraction condition (ADIM, ASLR, WALK and FRED) ($F_{3,42} = 1.018$; $p = 0.394$). Overall results are shown in Table **4-2** and Figure **4-2**. No adjustments were necessary, as Mauchly's test was non-significant ($p > 0.05$).

Table 4-2. Individual pairwise comparisons between contraction conditions for *transversus abdominis* thickness change expressed relative to resting/relaxed measurement. Units are percentage change (±SD).

	Mean1 (±SD)	Mean2 (±SD)	Mean Difference (± 95 % CI)	P value
ADIM-ASLR	43.99 (±16.89)	34.87 (±19.80)	-9.12 (-18.58 to 0.32)	0.061
ADIM-WALK		36.86 (±20.15)	-7.14 (-18.21 to 3.94)	0.195
ADIM-FRED		39.95 (±14.85)	-4.04 (-17.01 to 8.93)	0.521
ASLR-WALK	34.87 (±19.80)	36.86 (±20.15)	1.99 (-10.85 to 14.83)	0.748
ASLR-FRED		39.95 (±14.85)	5.08 (-6.84 to 17.01)	0.383
WALK-FRED	36.86 (±20.15)	39.95 (±14.85)	-3.09 (-15.21 to 9.02)	0.598

ADIM, abdominal drawing-in manoeuvre; ASLR, active straight leg raise; WALK, treadmill walking; FRED, exercise device; SD, standard deviation; CI, confidence interval.

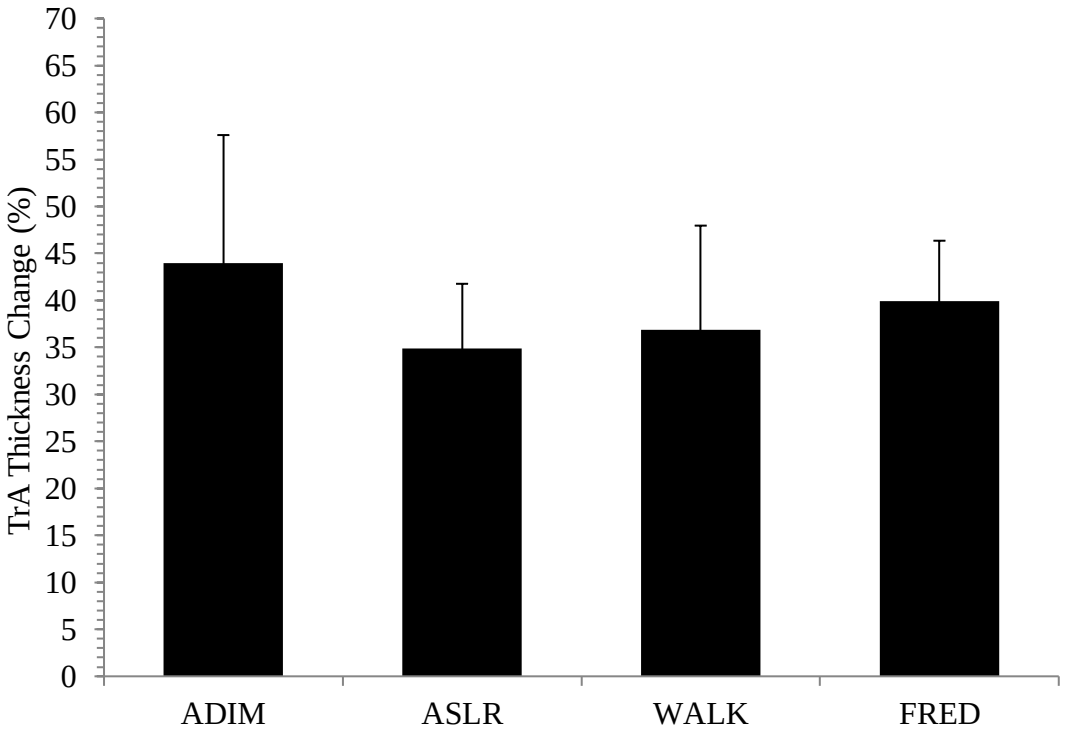

Figure 4-2. Change in muscle thickness of the *transversus abdominis*, expressed relative to resting thickness for each of the experimental conditions (ADIM-abdominal drawing-in manoeuvre; ASLR-active straight leg raise; WALK-treadmill walking; FRED-exercise device). Error bars represent the standard error of the measurement between days one and two as identified previously in chapter three.

4.4 Discussion

This chapter reported the examination of LM and TrA contraction in a healthy asymptomatic cohort during a range of static clinical tests (ADIM, ASLR, CAL and LCAL) and two dynamic exercise modalities (WALK and FRED) using freehand USI. The key findings from this investigation were that all contraction conditions successfully resulted in active relative thickness change of LM (range = 19.59 ± 10.30 % to 55.26 ± 14.36 %) and TrA (range = 34.87 ± 19.80 % to 43.99 ± 16.89 %). In the case of LM contraction, there was a notable interaction effect of contraction condition ($F_{3,42}$ = 41.000; p < 0.001) that was not present in TrA ($F_{3,42}$ = 1.018; p = 0.394).

Static upper extremity lifting tasks can be used as a method to assess non-volitional recruitment of the LM (Kiesel et al., 2007b) as opposed to voluntary recruitment strategies such as isometric swelling (Debuse et al., 2013; Van, Hides, & Richardson, 2006; Wallwork et al., 2009). Additionally, isometric swelling results in a much 'weaker' contraction of LM in comparison to those elicited during upper limb movement, when assessed by way of thickness change. For example, Debuse et al. (2013) reported a mean LM thickness change of 7.8 ± 13.2 % during swelling, whereas during unloaded and contralateral arm lifts thickness change is much greater, ranging from approximately 32 to 48 % (Kiesel et al., 2007b). In the current chapter, the contralateral arm lift and loaded contralateral arm lift conditions elicited significantly different changes in LM thickness of 19.59 ± 10.30 % and 40.32 ± 12.81 %, respectively. This is consistent with previous observations (Kiesel et al., 2007b; Teyhen et al., 2012).

The ADIM is included in many lumbar stabilisation training programmes as a mechanism to facilitate coactivation of the TrA and LM (O'Sullivan, Twomey, & Allison, 1998; Richardson et al., 2002). It is intended to preferentially recruit the TrA over the more superficial lateral abdominal wall muscles. It has also been shown to be able to differentiate between individuals with acute, sub-chronic, chronic (Kiesel et al., 2007b) and experimentally induced (Kiesel et al., 2008) LBP, highlighting potential for use as a muscle function assessment strategy. In the current study, the relative thickness change of 43.99 ± 16.89 % shown during ADIM is consistent with the previously reported values of 41 ± 22 % from Beazell and colleagues (2011), 65.5 ± 27.7 % from Teyhen and colleagues (2009a), and 52 ± 26.6 % from Gorbet and colleagues (2010). It is, however, noticeably lower than the 80.8 ± 39.0 % and 127 ± 89 % change in thickness reported by Koppenhaver et al. (2009a) and Teyhen et al.

(2005), respectively. In both of these instances, however, mean absolute thickness of the TrA during the rest condition (3.3 and 2.1 mm) was considerably less than in the current study. This could be explained (at least partially) by the authors not controlling for food intake prior to data collection. Such a procedure has subsequently been shown to significantly reduce resting thickness by approximately 23-27 % (Kordi et al., 2011).

From initial use as a clinical test of sacroiliac joint mobility in people with pelvic girdle pain (Mens et al., 1999) the active straight leg raise has recently been used in the assessment of deep abdominal muscle function by way of ultrasound (Teyhen et al., 2009b) and intramuscular EMG (Hu et al., 2012). During an active straight leg raise the TrA serves to compress the ilia against the sacrum, providing force closure between the two surfaces, thus increasing stiffness and reducing shear stresses (Richardson et al., 2002). Normative data from a large sample (n=340) cross-sectional study suggested that in a healthy asymptomatic mixed gender cohort thickness change of the TrA during the active straight leg raise is approximately 10-12 % (Teyhen et al., 2012). Thickness change of the TrA during the active straight leg raise in the current study (34.87 ± 19.80 %) is greater than those reference values, but broadly consistent with the previously published values of 23.7 ± 3.0 by Teyhen et al. (2009b).

Despite the ability of exercises such as the ADIM, CAL and ASLR to promote recruitment of LM and TrA, they are relatively static and have little similarity with most activities of daily living that are more dynamic. Walking has been proposed previously to be of potential benefit to people with low back pain as it is arguably the most functionally relevant exercise to most people (Hendrick et al., 2010). Walking, however, has not consistently been shown to be of benefit in this population (Joffe et

al., 2002; Mirovsky et al., 2006; Taylor, Evans, & Goldie, 2003; Torstensen et al., 1998).

It is also important to consider how FRED promotes recruitment of LM and TrA in comparison to the more static exercises (ADIM, ASLR, CAL, LCAL) used routinely in the assessment and rehabilitation of these muscles in people with low back pain and other conditions as well as walking. FRED exercise produced almost twice the increase in LM thickness than the contralateral arm lift. In comparison to the loaded contralateral arm lift, the increase in LM thickness seen during FRED exercise was not quite as large.

To date, no other study has reported the change in thickness of the LM during walking. Treadmill walking elicited the greatest LM thickness change of 55.26 ± 14.36 %, which was higher than that seen during FRED exercise (33.62 ± 14.85 %). Whilst, to date, only one study has provided data for the thickness change of LM when exercising on the FRED (Debuse et al., 2013), their findings (28.1 ± 7.5 %) are consistent with the observations made here. Walking is known to bi-phasically activate the LM during both ipsilateral and contralateral heel strikes (Anders et al., 2007; Dofferhof & Vink, 1985; Thorstensson et al., 1982) to approximately 37 % of MVC (Kim et al., 2012). Considered alongside the previous report that the loaded contralateral arm lift produces a contraction intensity of approximately 32 % MVC (Kiesel et al., 2007a), it would be reasonable to assume that the thickness changes during FRED exercise observed here represent contraction intensities similar to that of the loaded contralateral arm lift. Whilst walking elicited the greatest thickness increase of all conditions, this could simply be due to the measurement being taken at heel strike where there is a peak in LM activity (Anders et al., 2007; Dofferhof & Vink, 1985; Thorstensson et al., 1982).

For TrA, FRED exercise produced increases in thickness that were similar to all other conditions. Only one study has previously investigated thickness change of the TrA during treadmill walking (Bunce, Moore, & Hough, 2002). However, only reliability metrics (ICC, SEM, CV) were presented. No data were given as to absolute or relative thickness change parameters. In the current investigation, relative thickness change of the TrA was similar during both the WALK (36.86 ± 20.15 %) and FRED (39.95 ± 14.85 %) conditions. Debuse and colleagues (2013) currently provide the only available dataset for thickness change of the TrA during FRED, with reported values of 71.8 ± 29.3 %. A number of methodological differences exist, however, between the current study and the previous one by Debuse et al. (2013) that may help explain this difference. The previous study measured the change in TrA at the thickest part of the muscle observed on ultrasound whereas the current one used a standardised distance of 15mm from the aponeurosis (Reeve & Dilley, 2009). However, to date no studies have directly examined the impact of such a difference in measurement location. Additionally, and perhaps most consequentially, the authors of the previous study did not include specific information concerning the movement frequency whilst using the exercise device, simply stating instead that this did not exceed 1 Hz (Debuse et al., 2013). In the current study, however, this was significantly lower at 0.52 ± 0.1 Hz. It is likely that the lower thickness change observed here is a result of this reduction in movement frequency, and inferentially a reduction in the challenge to spinal stability.

4.4.1 Limitations

Data analysed and presented within this chapter were obtained during the first visit of participants in the study outlined in Chapter III. In that chapter, however, day-to-day variation in the thickness change of the LM was observed. Specifically, LM thickness

change decreased across three visits during the WALK condition and increased between days one and two in the FRED condition. However, the magnitudes of changes observed were within the respective MDC thresholds for both conditions and thus in practice would not be considered as a true change.

Furthermore, in the current study the measurement location used for the determination of TrA thickness was at a standardised distance of 15mm from the aponeurosis. Whilst this position does standardise the distance from the aponeurosis it may not standardise the location anatomically, dependent upon initial muscle size. Possible alternatives include taking the measurement at the point of maximum thickness (Debuse et al. 2013), at 50% of the muscle length (Teyhen et al. 2007), or at multiple points along the length of the muscle (Ferreira et al. 2004). Currently however no data exist directly examining the potential impact of such a difference in measurement location.

4.5 Conclusion

The data presented in this chapter demonstrate that dynamic functional activities may be a potentially useful assessment strategy of LM and TrA contraction (as derived from changes in muscle thickness) alongside more commonly used static techniques, such as loaded and unloaded contralateral arm lifts, the abdominal drawing-in manoeuvre and the active straight leg raise. The benefit is that this strategy is dynamic in nature and could be used to highlight motor control dysfunction relative a more complex movement pattern, as opposed to a relatively simple one. It also provides evidence that LM recruitment during FRED exercise is of a sufficiently low intensity to be representative of deep muscle action rather than a global superficial action. Given that the required lumbopelvic stability during activities of everyday living often requires

such a strategy, this method of assessment in people with, or at risk of, LBP could be extremely useful within practice. As yet, however, the utility of FRED exercise as a therapeutic intervention strategy remains unknown and should be investigated.

CHAPTER V

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Lumbar Multifidus and *Transversus Abdominis* Contraction during Static and Dynamic Stability Challenges

5.1 Introduction

Previously, it has been demonstrated that FRED exercise can facilitate recruitment of the LM and TrA (Debuse et al., 2013). In Chapters III and IV it was evidenced that reliability and precision of USI assessment of LM and TrA thickness in this study were consistent with published data and that FRED exercise is comparable to a range of typical LM and TrA functional assessment/rehabilitation strategies.

The aim of the study reported in this chapter was to examine the activity of deep (LM and TrA) and superficial (EO and IO) lumbopelvic muscles during commonly used exercises to 'improve core strength' and FRED exercise with a particular focus on the recruitment of deep versus superficial muscles, using USI.

As outlined on pages 65-67 in the Literature Review (Chapter II), FRED exercise shares important aspects with the motor control training (MCT) approach to LBP rehabilitation. In general, MCT encompass a fundamental strategy of retraining individuals to achieve spinal stability by restoring local lumbopelvic muscle function and activity appropriate to the specific task demands (Hodges, Ferreira, & Ferreira, 2009; Richardson, Hodges, & Hides, 2004), rather than a simple 'splinting' response where hyperactivity of the superficial trunk muscles is predominant and increases spinal stiffness (Bergmark, 1989; Gardner-Morse, Stokes, & Laible, 1995; Radebold et al., 2000). Specifically, MCT aims to 1) restore coordinated control of the trunk, particularly the dependence of activation of LM and TrA over the superficial musculature, 2) promote anticipatory and tonic activity during static and dynamic tasks, and 3) progress to daily function. This is typically facilitated with the use of repeated isolated isometric voluntary contractions of the TrA sustained for 10 seconds

(Koumantakis, Watson, & Oldham, 2005; Miller et al., 2005; O'Sullivan, Twomey, & Allison, 1997; Shaughnessy & Caulfield, 2004) with progression regularly involving additional load through limb movement (O'Sullivan, Twomey, & Allison, 1997; Rasmussen-Barr, Nilsson-Wikmar, & Arvidsson, 2003; Shaughnessy & Caulfield, 2004). Additionally, the use of various other methods to further increase the demands for reactive stability by creating an unstable base of support have been suggested including exercise balls, balance boards, Bodyblades and Thera-Bands (Hodges et al., 2013) . However, several authors (Hamlyn, Behm, & Young, 2007; Marshall & Murphy, 2005) have questioned the skill transference as a result of those exercises, due to their predominantly static nature of the exercises.

5.2 Method

5.2.1 Design

A random order single-session within-subject repeated measures design was used to investigate the activity of LM, TrA, IO and EO during a range of upright postural stability challenges. Participants visited the laboratory on a single occasion and were assessed in all experimental conditions during this visit. To avoid any potential systematic order influences, the conditions, with the exception of the rest condition which was captured first, were conducted in a randomised order (Teyhen et al., 2011). Order was established using a custom permutation list and random rank ordering system (as used previously in Chapters III and IV).

5.2.2 Participants

Twelve healthy adult males (mean ± SD age: 25.8 ± 2.7 years, body mass: 78.3 ± 9.2 kg, height: 1.78 ± 0.08 m, and body mass index: 24.7 ± 2.3 $kg \cdot m^{-2}$) volunteered for

this study. Participants were excluded if they had a history of LBP within the preceding six months, existing or previous musculoskeletal pathology/injury, any known neuromuscular or joint disease, or previous abdominal or lumbar spine surgery.

Approval for this study was gained from the Ethics Committee of the School of Life Sciences at Northumbria University, Newcastle upon Tyne, England. All participants gave fully informed (Appendix L) written consent (Appendix M) to participate in this study.

5.2.3 Assessment of Lumbar Multifidus and Lateral Abdominal Wall Musculature

As detailed in previous chapters (Chapters III and IV), a digital ultrasound imager (Technos MP, Esaote, Genoa, Italy) in B-mode (brightness mode) was used by a single operator to collect images of the LM, TrA, IO and EO.

5.2.4 Experimental Protocol

During their visit participants were required to complete a battery of experimental conditions during which LM, TrA, IO and EO muscle activity were assessed (Figure **5-1**). Experimental conditions were upright sitting on a stable surface (Sit), upright sitting on a gym ball to create an unstable surface (Gym), upright standing on a stable surface (Stand), upright standing on a wobble board to create an unstable surface (Wobble) and using the exercise device in sitting (FREDSit) and standing (FREDStand). Each of the conditions was reported relative to the resting position.

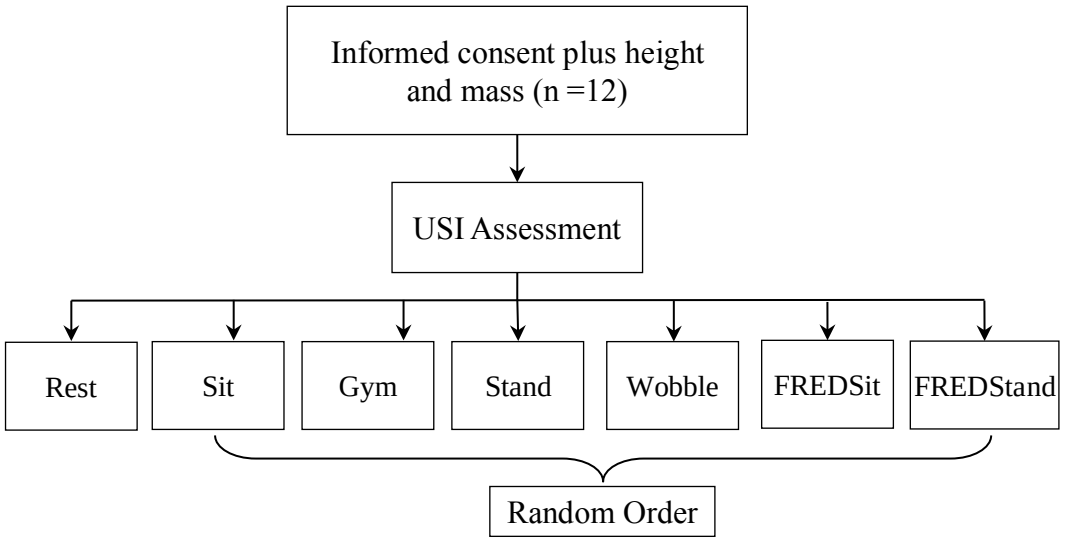

Figure 5-1. Schematic representation of experimental protocol. Following supine resting measurements (Rest) ultrasound images were taken in a randomised order of the *lumbar multifidus, transversus abdominis*, and *internal* and *external oblique muscles* during stable sitting (Sit) and standing conditions (Stand) as well as unstable sitting on a gym ball (Gym) and standing on a wobble board (Wobble) and dynamically unstable conditions using the exercise device in sitting (FREDSit) and standing (FREDStand).

During the resting and FRED conditions, LM and TrA were imaged in accordance with the method described in Chapter II, with the only exception being that of FREDSit, where participants were instructed to use the available seat on the exercise device (Debuse et al., 2013).

During the sitting condition, participants were seated with a neutral erect posture on a perching stool (Nottingham Rehab Supplies, Nottingham, UK) with the height of the stool adjusted between participants such that the hips and knees were flexed to 90°. A neutral erect posture was defined as a position in which the participant had a neutral pelvic tilt, neutral lumbar lordosis, and neutral thoracic kyphosis as assessed by visual inspection (O'Sullivan et al., 2002). Participants placed their arms across their chest with hands loosely placed on the contralateral shoulders so as to avoid obstructing the

placement of the transducer during imaging (Ainscough-Potts, Morrissey, & Critchley, 2006), and positioned their feet on the floor, shoulder width apart and facing forwards.

For the Gym ball condition, participants sat on a 65 cm diameter gym ball with a neutral erect spinal posture with hands placed on contralateral shoulders. Participants were positioned such that the hips and knees were flexed to 90° as in the Sit condition.

In the Stand condition, participants stood on the (stable) laboratory floor in an erect upright posture such that an imaginary vertical line would approximately intersect the lateral malleolus, greater trochanter and acromion (O'Sullivan et al., 2002). Feet were positioned approximately shoulder width apart facing forwards.

In the Wobble condition, participants stood aligned as in the Stand condition, but on a wobble board (Sissel 3080, Sissel UK Ltd, Mytholmroyd, UK) consisting of a 400 mm diameter rigid board set atop a hemisphere, resulting in a maximum omnidirectional tilt of 23.5°.

5.2.5 Image Analysis & Blinding

All ultrasound images were analysed in accordance with the methods described in the previous chapters (Chapters III and IV). The only exception being the additional inclusion of the IO and EO taken as the linear distance between the superficial and deep hyperechoic fascial lines perpendicular to the muscle fibre direction, at a standardised distance of 15mm medial from the aponeurosis (Reeve & Dilley, 2009) for each individual muscle (Figure **5-2**).

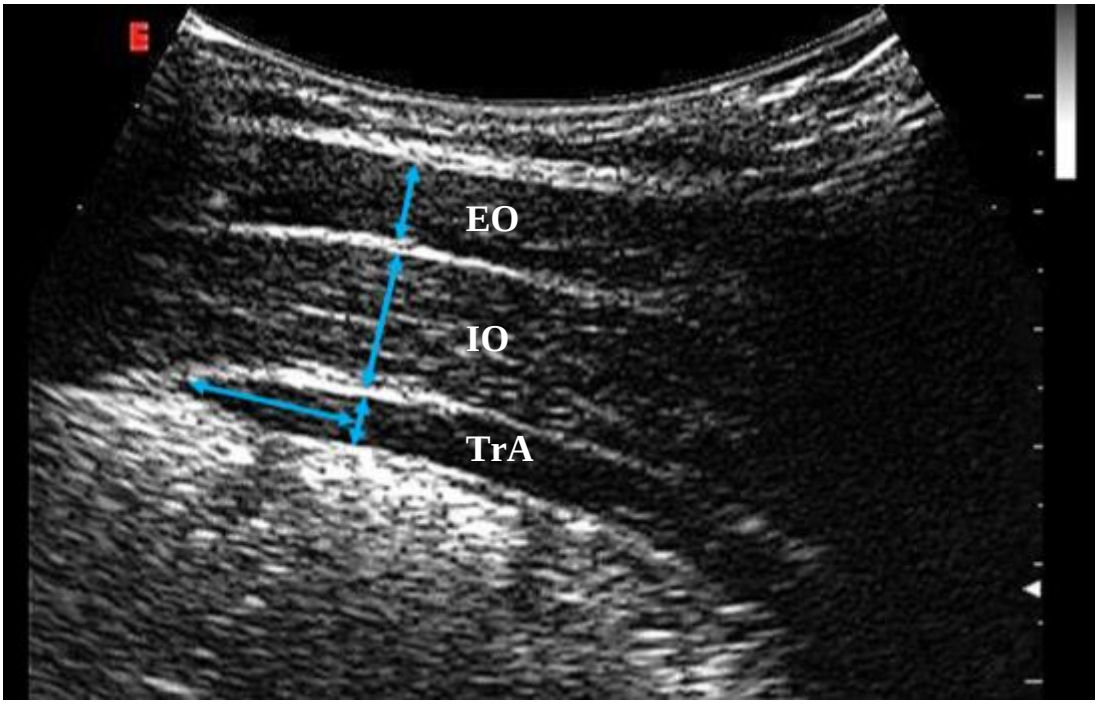

Figure 5-2. Exemplar captured ultrasound image of the anterolateral abdominal wall including the *external oblique* (EO), *internal oblique* (IO) and the *transversus abdominis* (TrA) muscles. Blue lines denote the standardised distance from the aponeurosis and the orientation of measurement.

Thickness change was expressed as a percentage increase from resting measurements for each experimental condition for both LM (Kiesel et al., 2007a) TrA, IO and EO (Critchley & Coutts, 2002) and given as:

$$\%Change = \left(\frac{(Contracted - Rest)}{Rest}\right) \cdot 100$$

Additionally a preferential contraction (TrA Pref) metric was calculated representing the coactivation of the TrA relative to the IO and EO muscle group, representing the proportional difference of muscle thickness relative to the resting and contracted conditions (Teyhen et al., 2005).

$$TrA\ Pref = \left(\frac{TrA\ contracted}{TrA + IO + EO\ contracted}\right) - \left(\frac{TrA\ rest}{TrA + IO + EO\ rest}\right) \cdot 100$$

Thus, the difference between the two proportions gives the relative change in the proportion of TrA thickness relative to the total lateral abdominal wall muscle thickness. Therefore, values greater than zero illustrate a contraction with the majority of change in muscle thickness occurring in the TrA, whereas a negative value represents a greater relative change in EO and IO thickness.

5.2.6 Statistical Analysis

One-way analyses of variance with repeated measures (RM ANOVA) were used to examine data for the presence of any main effects between changes in muscle thickness and experimental condition (1x6) for each individual muscle as well as for TrA Pref. The level of significance was set at 95 % ($p < 0.05$) for all data. Sphericity of data was assumed if Mauchly's Test was non-significant ($p > 0.05$). If this assumption was not met, adjustment was made using a Greenhouse-Geisser correction if $\varepsilon < 0.75$, or a Huynh-Feldt correction if $\varepsilon \geq 0.75$ as recommended by Girden (1992) If significant main effects were revealed by the RM ANOVA, *post hoc* pairwise comparisons (LSD) were used to identify the location of any significant differences between conditions.

All statistical analysis was performed within PASW Statistics v.18 (SPSS Inc., Chicago, Illinois). Summary raw data tables for lumbar multifidus and transversus abdominis, internal obliques, and external obliques are available in appendices N and O respectively.

5.3 Results

5.3.1 Lumbopelvic Muscle Activity

A significant main effect of condition was present for relative LM thickness change ($F_{(2.581, 28.389)} = 19.709$, $p < 0.001$). With Mauchly's test of sphericity indicating a violation of this assumption ($X^2(14) = 28.95$, $p = 0.013$) degrees of freedom were adjusted using Greenhouse-Geisser estimates ($\varepsilon = 0.516$). Subsequent pairwise comparisons (Table **5-1** and Figure **5-3**) revealed that all standing conditions elicited a significantly greater ($p < 0.05$) relative thickness change than sitting condition equivalents, regardless of surface lability.

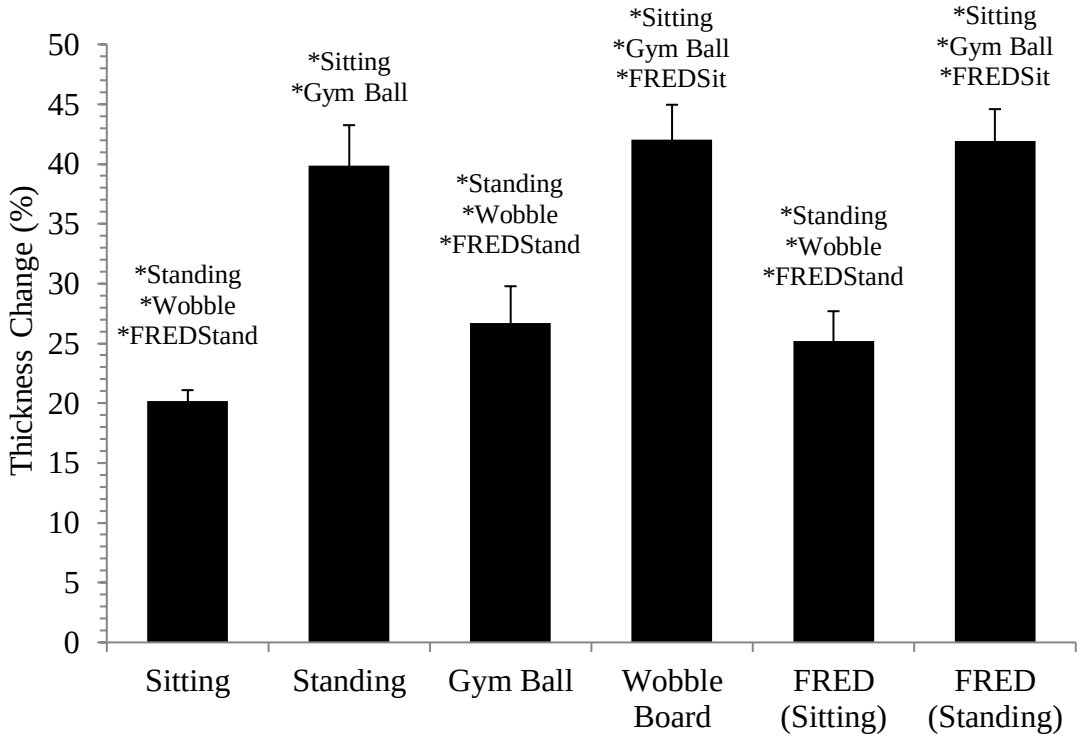

Figure 5-3. Thickness change of the *lumbar multifidus*, expressed relative to the resting condition for each individual experimental condition. *denotes pairwise significant difference between that particular condition and each of the named conditions.

Table 5-1. Pairwise comparisons of thickness change of the LM relative to the resting condition (percentage thickness change) between experimental exposures. A negative mean difference value indicates the initial condition (Condition 1) was less than the comparative condition (Condition 2).

Condition 1	Condition 2	Mean1 (±SD)	Mean2 (±SD)	Mean Difference (± 95 % CI)	*P*
Sitting	Standing	20.16 (±0.94)	39.85 (±3.40)	-19.69 (-28.61 to -10.78)	0.001*
	Gym Ball		26.70 (±3.08)	-6.54 (-13.59 to 0.50)	0.157
	Wobble		42.02 (±2.94)	-21.86 (-29.02 to -14.70)	<0.001*
	FRED (Sitting)		25.20 (±2.49)	-5.04 (-10.47 to 0.37)	0.156
	FRED (Standing)		41.92 (±2.66)	-21.76 (-27.80 to -15.74)	<0.001*
Standing	Gym Ball	39.85 (±3.40)	26.70 (±3.08)	13.15 (2.40 to 23.89)	0.021*
	Wobble Board		42.02 (±2.94)	-2.17 (-6.90 to 2.56)	0.335
	FRED (Sitting)		25.20 (±2.49)	14.65 (7.16 to 22.14)	0.001*
	FRED (Standing)		41.92 (±2.66)	-2.07 (-8.01 to 3.85)	0.457
Gym Ball	Wobble Board	26.70 (±3.08)	42.02 (±2.94)	-15.32 (-22.40 to -8.22)	0.001*
	FRED (Sitting)		25.20 (±2.49)	1.50 (-5.46 to 8.46)	0.647
	FRED (Standing)		41.92 (±2.66)	-15.22 (-23.82 to -6.63)	0.002*
Wobble Board	FRED (Sitting)	42.02 (±2.94)	25.20 (±2.49)	16.82 (10.70 to 22.93)	<0.001*
	FRED (Standing)		41.92 (±2.66)	0.10 (-6.17 to 6.36)	0.975
FRED (Sitting)	FRED (Standing)	25.20 (±2.49)	41.92 (±2.66)	-16.72 (-223.67 to -9.78)	<0.001*

* indicates a significant difference between conditions at the $p < 0.05$ level

5.3.1.1 Transversus Abdominis, Internal Oblique and External Oblique Activity

The assumption of sphericity was violated in both the TrA ($X^2(14) = 24.53$, $p = 0.045$) and IO ($X^2(14) = 36.987$, $p = 0.001$), therefore, degrees of freedom were adjusted using Greenhouse-Geisser estimates of sphericity ($\varepsilon = 0.520$ [TrA] and 0.548 [IO]). This violation was not present for EO ($X^2(14) = 19.529$, $p = 0.158$), thus degrees of freedom were unadjusted. A significant interaction effect of condition was found only in the thickness change of the TrA ($F_{(2.601, 28.613)} = 7.006$, $p = 0.002$). Significant interaction effects of condition were not present in either IO ($F_{(2.738, 30.123)} = 1.700$, $p = 0.191$) or EO ($F_{(5,55)} = 3.381$, $p = 0.098$). All static conditions resulted in similar magnitudes of thickness change (range = 27.08 ± 5.05 to 38.09 ± 6.52) with the only conditions

sensitive to posture and lability effects were those of FREDSit and FREDStand (Figure **5-4** and Table **5-2**).

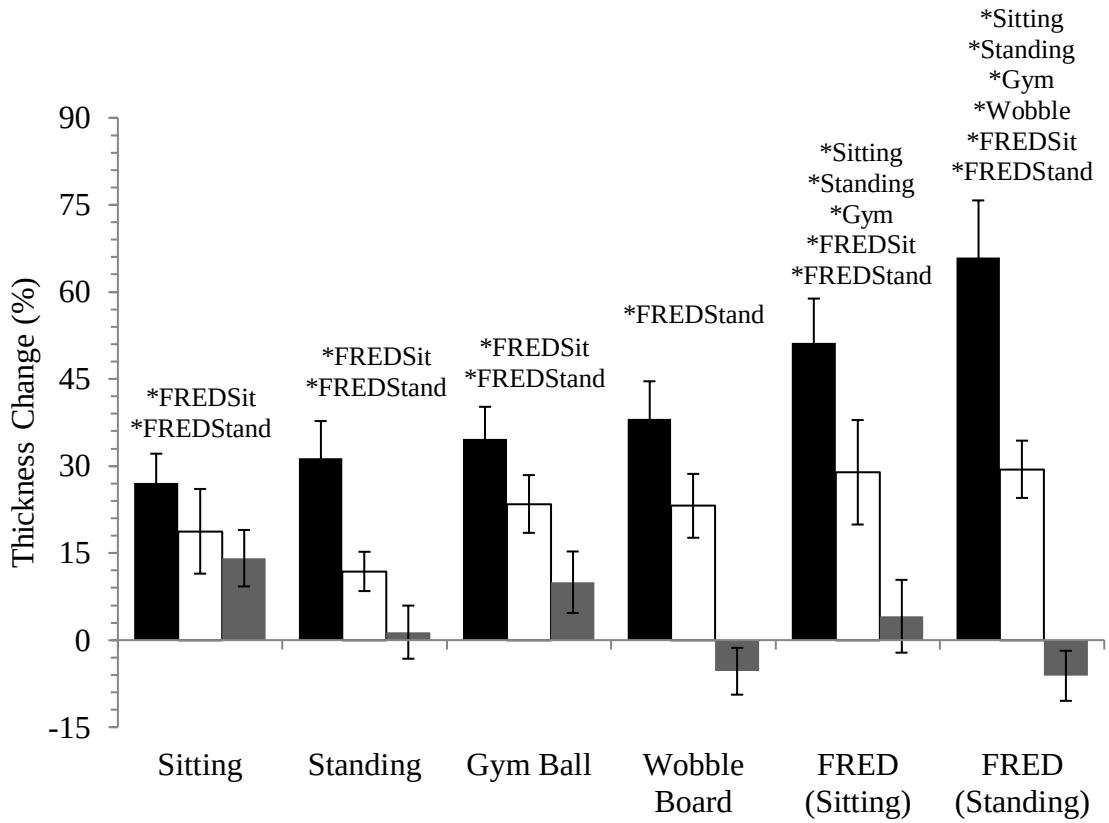

Figure 5-4. Thickness change of the *transversus abdominis* (black bars), *internal oblique* (white bars) and *external oblique* (grey bars) expressed relative to the resting condition, as a result of each individual experimental exposure. *denotes pairwise significant difference between that particular condition and each of the named conditions.

Table 5-2. Pairwise comparisons of thickness change of the TrA relative to the resting condition (percentage thickness change) between experimental exposures

Condition 1	Condition 2	Mean1 (±SD)	Mean2 (±SD)	Mean Difference (± 95 % CI)	*P*
Sitting	Standing	27.08 (±5.05)	31.36 (±6.39)	-4.28 (-19.07 to10.51)	0.537
	Gym Ball		34.71 (±5.49)	-7.63 (-18.00 to 2.73)	0.133
	Wobble Board		38.09 (±6.52)	-11.01 (-27.98 to 5.96)	0.181
	FRED (Sitting)		51.25 (±7.60)	-24.17 (-41.56 to -6.78)	0.011 *
	FRED (Standing)		65.95 (±9.81)	-38.88 (-64.25 to -13.50)	0.006 *
Standing	Gym Ball	31.36 (±6.39)	34.71 (±5.49)	-3.35 (-17.25 to 10.55)	0.606
	Wobble Board		38.09 (±6.52)	-6.73 (-16.97 to 3.51)	0.176
	FRED (Sitting)		51.25 (±7.60)	-19.89 (-38.55 to -1.22)	0.039 *
	FRED (Standing)		65.95 (±9.81)	-34.59 (-54.50 to -14.69)	0.003 *
Gym Ball	Wobble Board	34.71 (±5.49)	38.09 (±6.52)	-3.38 (-17.99 to 11.24)	0.621
	FRED (Sitting)		51.25 (±7.60)	-16.54 (-31.92 to -1.15)	0.037 *
	FRED (Standing)		65.95 (±9.81)	-31.24 (-54.29 to -8.20)	0.012 *
Wobble Board	FRED (Sitting)	38.09 (±6.52)	51.25 (±7.60)	-13.16 (-28.80 to -2.48)	0.091
	FRED (Standing)		65.95 (±9.81)	-27.87 (-47.13 to -8.60)	0.009 *
FRED (Sitting)	FRED (Standing)	51.25 (±7.60)	65.95 (±9.81)	-14.70 (-29.16 to -0.25)	0.047 *

* indicates a significant difference between conditions at the $p < 0.05$ level

5.3.1.2 Preferential Contraction

For TrA Pref the assumption of sphericity had been violated ($X^2(14) = 24.654$, $p = 0.044$), and therefore, degrees of freedom were corrected using Greenhouse-Geisser estimates ($\varepsilon = 0.507$). A significant effect was found between conditions ($F_{(2.537, 27.907)} = 18.433$, $p < 0.001$), with subsequent pairwise comparisons presented in Figure **5-5** and Table **5-3**. These pairwise comparisons highlight significantly greater ($p < 0.05$) preferential contraction of TrA in FREDStand (5.69 ± 1.06) than all other comparative conditions (range = 1.52 ± 0.76 to 3.10 ± 0.64), with the only exception being that of FREDSit (3.89 ± 0.85).

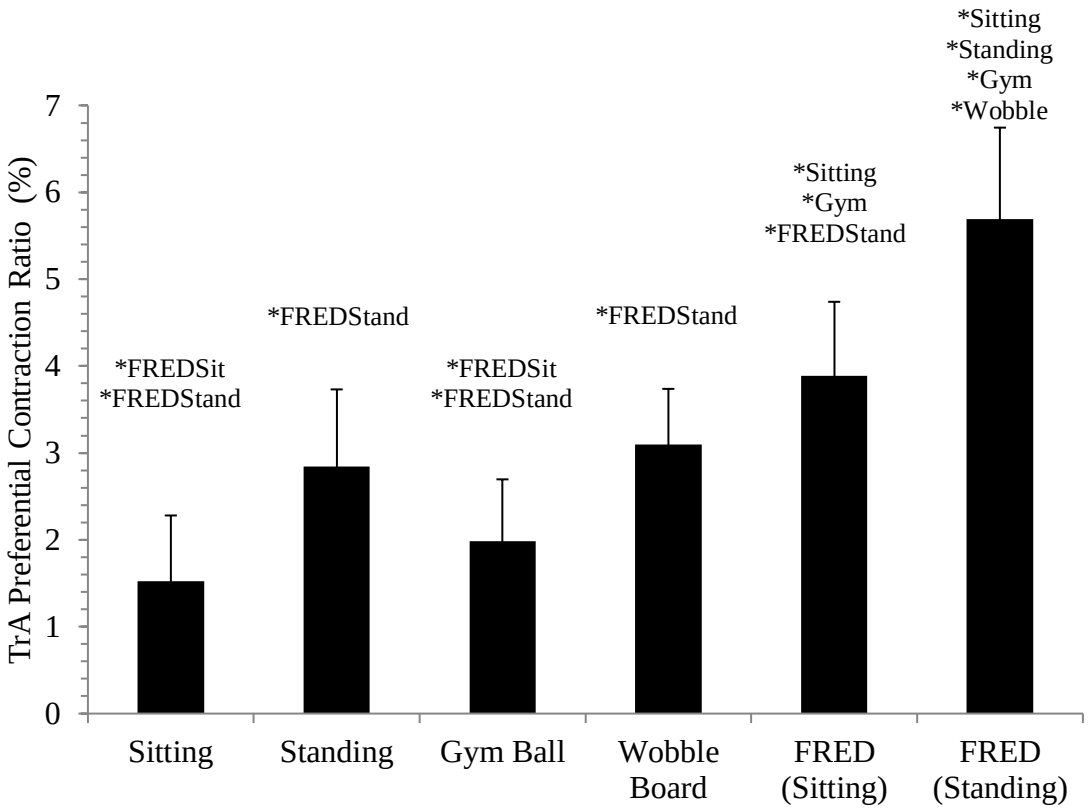

Figure 5-5. Preferential contraction of the *transversus abdominis* as a result of each individual experimental exposure. *denotes pairwise significant difference between that particular condition and each of the named conditions.

Table 5-3. Pairwise comparisons of the *transversus abdominis* preferential contraction metric between experimental exposures.

Condition 1	Condition 2	Condition 1 Mean (±SD)	Condition 2 Mean (±SD)	Mean Difference (± 95 % CI)	*P*
Sitting	Standing	1.52 (±0.76)	2.85 (±0.88)	-1.33 (-3.73 to 1.08)	0.267
	Gym Ball		1.98 (±0.71)	-0.46 (-1.72 to 0.79)	0.449
	Wobble		3.1 (±0.64)	-1.58 (-3.54 to 0.39)	0.113
	FRED (Sitting)		3.89 (±0.85)	-2.37 (-4.42 to -0.31)	0.028*
	FRED (Standing)		5.69 (±1.06)	-4.17 (-7.15 to -1.18)	0.011*
Standing	Gym Ball	2.85 (±0.88)	1.98 (±0.71)	0.87 (-1.52 to 3.25)	0.442
	Wobble Board		3.1 (±0.64)	-0.25 (-1.71 to 1.2)	0.711
	FRED (Sitting)		3.89 (±0.85)	-1.04 (-3.79 to 1.71)	0.422
	FRED (Standing)		5.69 (±1.06)	-2.84 (-5.49 to -0.19)	0.038*
Gym Ball	Wobble Board	1.98 (±0.71)	3.1 (±0.64)	-1.12 (-2.87 to 0.64)	0.189
	FRED (Sitting)		3.89 (±0.85)	-1.91 (-3.41 to -0.40)	0.018*
	FRED (Standing)		5.69 (±1.06)	-3.71 (-6.42 to -0.98)	0.012*
Wobble Board	FRED (Sitting)	3.1 (±0.64)	3.89 (±0.85)	-0.79 (-2.91 to 1.33)	0.430
	FRED (Standing)		5.69 (±1.06)	-2.59 (-4.63 to -0.55)	0.017*
FRED (Sitting)	FRED (Standing)	3.89 (±0.85)	5.69 (±1.06)	-1.80 (-3.90 to -0.30)	0.086

* indicates a significant difference between conditions at the $p < 0.05$ level

5.4 Discussion

This chapter aimed to examine the activity of key lumbopelvic muscles, namely the LM, TrA, IO and EO, during a series of challenges to upright postural stability. The key findings from this investigation were that all stability challenges successfully induced non-volitional co-contraction of both the LM (range = 20.16 ± 0.94 % to 42.02 ± 2.94 %) and TrA (range = 27.08 ± 5.05 to 65.95 ± 9.81 %). Additionally, it was observed that the LM followed a pattern where all standing conditions elicited greater recruitment than sitting conditions, with no additional effect of surface lability. Contrastingly, the TrA only demonstrated an effect of surface instability during FRED conditions. Lastly, and arguably of most importance, the preferential contraction ratio

of the TrA was found to be greatest (5.69 ± 1.06 %) during FRED exercise in standing in comparison to all other static stability challenges (range = 1.52 ± 0.76 to 2.85 ± 0.88 %), suggesting that FRED exercise is able to preferentially recruit the deep lumbopelvic muscles (TrA) in comparison to the superficial lumbopelvic muscles (IO and EO).

5.4.1 Surface Stability and Muscle Recruitment in Sitting and Standing

In the study reported in this chapter, LM thickness change was similar in magnitude between stable (20.16 ± 0.94 %), unstable (26.70 ± 3.08 %) and dynamically unstable (25.20 ± 2.49 %) sitting conditions. In standing, LM thickness change also did not differ between stable (39.85 ± 3.40 %), unstable (42.02 ± 2.94 %) and dynamically unstable (41.92 ± 2.66 %) conditions. This is contrary to previously published literature that suggests some activities performed on labile surfaces increase muscle activation compared to similar activities on more stable surfaces (Grenier, Vera-Garcia, & McGill, 2000; Marshall & Murphy, 2005). In a study by McGill and colleagues (2006), no differences were found in mean amplitude EMG between sitting on a stable surface and sitting on an exercise ball, whereas in studies by Gregory et al. (2006) and Kingma and Van Dieén (2009), significant increases were observed. A key distinction between these studies, however, is that only McGill et al. (2006) compared exercise ball sitting with unsupported stable sitting. The others both compared exercise sitting with sitting on a chair with a backrest. Thus, with respect to investigations of a similar nature to the current study (Gregory, Dunk, & Callaghan, 2006; Kingma & van Dieën, 2009; McGill, Kavcic, & Harvey, 2006), the findings presented here are consistent with the view that unstable sitting does not significantly increase trunk muscle activation when compared to stable sitting (O'Sullivan et al., 2013; O'Sullivan et al., 2006a).

The findings of this study indicate that similar mechanisms apply to TrA, IO and EO. Thickness of TrA was not different between stable (27.08 ± 5.05 %) and unstable (34.71 ± 5.49 %) sitting conditions, nor was it different during stable (31.36 ± 6.39 %) and unstable (38.09 ± 6.52 %) standing. Interestingly, however, there was a significant increase during the FREDSit (51.25 ± 7.60 %) and FREDStand (65.95 ± 9.81 %) conditions. This is also consistent with the findings of Debuse et al. (2013) who reported a similar pattern of recruitment. This would suggest that, irrespective of surface lability or the inclusion of a dynamic movement, the contribution of the LM to maintenance of spinal stability is consistent, whilst the cyclical movement of the lower limbs does increase the need for additional stabilisation via supplementary contribution of the TrA.

5.4.2 Preferential Activation of Transversus Abdominis

Motor control exercises (Richardson & Jull, 1995) have been shown to be effective in improving pain and function (Ferreira et al., 2006; Hides et al., 2008; Hides et al., 2012; Kriese et al., 2010; Rackwitz et al., 2006) in individuals with evidence of 'insufficient' neuromuscular control of intersegmental spinal stability. A key component of this approach is enabling contraction of the TrA preferentially of the EO and IO, therefore facilitating preferential activation of deep over superficial musculature (Richardson & Jull, 1995). In the current chapter, the TrA Pref during exercising in both sitting and standing conditions was 3.89 ± 0.85 % and 5.69 ± 1.06 %, respectively. This is marginally lower than previously reported values for TrA Pref during abdominal hollowing exercises that range from 6 ± 3 % (Mannion et al., 2008) to 9 ± 5 % (Teyhen et al., 2005). It should, however, be noted that in both of these instances abdominal hollowing was performed in a supine position. When performing abdominal hollowing in supine and standing positions, Manshadi et al. (2011) reported

a significant reduction in TrA Pref of approximately 66 % from supine to standing (6 ± 4 % in supine versus 2 ± 6 % in standing). This would seem to indicate that FRED exercise recruits TrA more preferentially than one of the key components (voluntary activation) of specific stabilisation exercise programmes (Ferreira et al., 2007; Hides et al., 2010; Hides, Jull, & Richardson, 2001; Hides & Stanton, 2014; O'Sullivan, Twomey, & Allison, 1997; Richardson & Jull, 1995) in upright postures. Conventionally, this is achieved via voluntary activation training using the abdominal drawing-in manoeuvre, thus the dynamically unstable FRED exercise investigated here is more effective in this regard, at least in the current sample population.

5.4.3 Concomitant Activation of Lumbar Multifidus and Transversus Abdominis

Whilst it is proposed that coactivation of the LM and TrA is required for lumbar stability, and is advocated during motor control exercise programmes (Hides, Jull, & Richardson, 2001), such concomitant recruitment may be dependent on task complexity (McCook, Vicenzino, & Hodges, 2009) and not necessarily be obligatory for all tasks (MacDonald, Moseley, & Hodges, 2006). During all included conditions within the current study, this co-activation strategy was evident, though increases in thickness were observed in the LM from sitting to standing positions and in the TrA during dynamic activities only. This is consistent with the theory that increases in lumbopelvic stability are attributable more to modulations in TrA than LM recruitment, where activity of the LM is sufficient for the required upright posture (McCook, Vicenzino, & Hodges, 2009). Although it should also be noted that LM and TrA were not measured simultaneously and as such, this will require future confirmation via either simultaneous ultrasound imaging or electromyography.

5.5 Limitations

Although consistent with the methods of Ainscough-Potts et al. (2006) and Rasouli et al. (2011), only a single diameter gym ball was used within the current study. Variances in participant height meant standardisation of hip and knee alignment could only be partially achieved in some individuals, unlike the stable seated condition where an adjustable stool mitigated this problem. Therefore, direct comparison with the stable seated condition should be done with caution.

This investigation used relative changes in muscles thickness of the TrA, IO and EO in order to assess the contributions of the anterolateral abdominal wall muscles simultaneously. Whilst this USI method has previously been shown to be valid for TrA and IO (Hodges et al., 2003; McMeeken et al., 2004), the same cannot be said for EO (Brown & McGill, 2010; Hodges et al., 2003; John & Beith, 2007). This is likely due to the laminate-like organisation of the abdominal wall resulting in complex patterns of deformation that differ dependent upon the task and, therefore, the relative contributions of individual muscles (Brown & McGill, 2010). As such, it is possible the activity of EO could be greater than the negligible contribution observed here, however, this requires further investigation using EMG during FRED exercise.

5.6 Conclusion

The study reported in this chapter shows that the muscle recruitment strategy during dynamic unstable exercise conditions fulfils a large number of the proposed key requirements for retraining of the deep lumbopelvic musculature. These include contraction of the LM and TrA at low levels of contraction, with preferential activation of TrA (deep) over the more superficial IO and EO muscles. Of particular consequence

is likely to be the similar magnitude of LM contraction observed during all standing conditions as well as the increased preferential contraction of the TrA as the challenge to stability increased. This suggests a stabilising strategy that is beyond basic spinal stiffening obtained through co-contraction of superficial flexors and extensors that increases spinal load (Gardner-Morse & Stokes, 1998; Granata & Marras, 2000). Whilst it is apparent that some of the comparative conditions included also meet some of the proposed requirements, the dynamic unstable exercise on the FRED is likely to be beneficial for use in the rehabilitation of motor control in people with LBP (where motor control deficits are apparent). What is yet to be determined, however, is whether this increased activity of the TrA can be explained by the underlying kinematics of FRED exercise.

CHAPTER VI
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Examination of 3-Dimensional Trunk, Pelvic and Lower Limb Kinematics during FRED Exercise and Walking

6.1 Introduction

In the previous chapters (Chapter IV and V), it was shown that FRED exercise is potentially useful in the rehabilitation of motor control in people with LBP, primarily due to its fulfilment of various proposed requirements for retraining of the deep lumbopelvic musculature.

In Chapter IV FRED exercise was shown to activate LM and TrA in a non-volitional manner (without conscious effort, prior training or instruction) and doing so at a level of approximately 30% MVC. Recruitment of the TrA was found to be similar during common static challenges to stability such as the use of wobble boards and gym balls, whereas there was a significant increase during both the sitting and standing FRED conditions. This suggests that the cyclical movement of the lower limbs as well as an upright weight-bearing posture during FRED exercise increase the need for additional stabilisation via contraction of the TrA. Additionally, magnitudes of LM contraction were similar during all standing conditions (and significantly increased from sitting conditions), irrespective of the challenge to stability, suggesting a stabilising strategy beyond basic spinal stiffening obtained through co-contraction of superficial flexors and extensors.

Debuse et al. (2013) implied that tonic muscle activity is likely to be responsible for the stable lumbopelvic region during FRED exercise. However, no information was provided on lumbopelvic and lower limb kinematics of the user while exercising to identify how the exercise device promoted lumbopelvic stability and, thus, tonic muscle activity.

The aim of the study reported in this chapter was to compare lower limb, pelvic and trunk kinematics during FRED exercise (FRED) and overground walking (WALK), with a particular focus on the level of lumbopelvic stability in both activities, using 3-dimensional motion analysis.

6.2 Method

6.2.1 Participants

Sixteen healthy adult male volunteers (mean ± SD age: 26.5 ± 3.38 years, body mass: 82.18 ± 7.21 kg, height: 1.78 ± 0.05 m, and body mass index: 25.89 ± 2.16 $kg \cdot m^{-2}$) with no recent history of LBP, gait impairments, or other conditions affecting their ability to walk or exercise, agreed to participate in this study. Participants gave their fully informed (Appendix P) written consent (Appendix Q) to take part. Approval for this study was gained from the Ethics Committee of the School of Life Sciences at Northumbria University, Newcastle upon Tyne, England prior to data collection.

6.2.2 Three-Dimensional Motion Capture

Three-dimensional trajectories of 37 retro-reflective markers (Ø = 14 mm) were captured at a sampling frequency of 200 Hz using a 12-camera near-infrared motion capture facility (MX T20, Vicon Motion Systems, Oxford, UK). Markers were placed in accordance with a standard full-body (Figure **6-1**) model (Plug-in-Gait, Vicon Motion Systems, Oxford, UK), which consists of a 15-segment rigid-linked model of the head, thorax, pelvis, and bilateral upper arms, forearms, hands, thighs, lower legs and feet. Only the segmental orientations of the thorax, pelvis, thighs and lower legs were subsequently used for analysis.

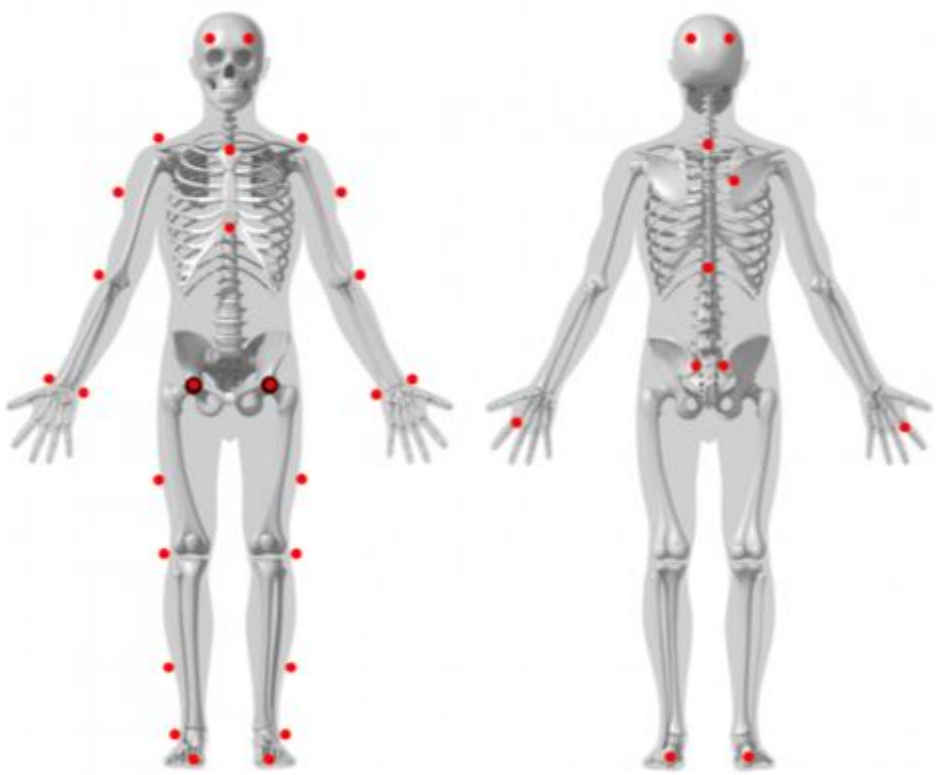

Figure 6-1. Full body plug-in gait marker placements

The motion capture system was calibrated before all testing sessions using a standard dynamic protocol, with a 5-marker calibration wand (Vicon Motion Systems, Oxford, UK). System calibration was accepted when the image error of all 12 cameras was less than 0.2 mm.

Body mass, height and anthropometric measurements, including leg length (anterior superior iliac spine to medial malleolus), ankle widths and knee widths, necessary for the correct operation of the model used were taken in triplicate and the mean value used for subsequent analysis.

6.2.3 Experimental Protocol

Participants completed an overground walking (WALK) condition and a FRED exercise (FRED) condition in a counterbalanced random order within a single session.

In the WALK condition, participants were asked to walk along a level 7.5 m walkway, instrumented with embedded force plates (OR6-7, AMTI, Watertown, Massachusetts, USA), at a self-selected comfortable speed. Starting positions were adjusted individually to ensure that 'clean' foot contacts with the force plates could be achieved without direct targeting by the participant. A minimum of 10 trials were completed, before six trials - without evidence of targeting - were selected for subsequent analysis.

In the FRED condition participants were given an initial five-minute period to familiarise themselves with the exercise device. Following this, 30 seconds of trajectory data were captured during exercise in standing. Subsequently, six cycles were chosen at random for analysis. All participants were given standardised instructions on the correct use of the device emphasising the need for a 'slow controlled movement' whilst maintaining 'an upright posture' during each cycle.

6.2.4 Data Processing and Reduction

Marker trajectories collected during WALK and FRED trials were reconstructed and processed within Vicon Nexus (1.7, Vicon Motion Systems, Oxford, UK). Lost or obscured trajectory segments were interpolated using a quintic-spline function for gaps less than or equal to 10 frames (0.05 s) or a pattern fill function for gaps greater than 10 frames, which uses the trajectory of a marker with a similar predicted displacement trajectory. Marker trajectories were then low pass filtered at 5 Hz using a fourth-order zero lag Butterworth filter (Saunders et al., 2005).

Key "gait cycle" phases (stance and swing) were demarcated for both the WALK and FRED conditions using discrete gait cycle events. Heel strikes and toe offs during WALK were detected using the vertical component of the ground reaction force

obtained from the force plates embedded flush with the walkway surface at the centre of the calibrated capture volume. When using the FRED, the feet remain in contact with the foot plates at all times during the motion cycle. Therefore, data collected during FRED exercise were divided into a 'stance' and 'swing' phase based on the trajectory of a marker placed on the front corner of the foot plate: stance was defined as the most anterior to the most posterior foot plate position, and swing was from the most posterior to most anterior foot plate position.

Three-dimensional angular displacements for the trunk (thorax relative to the pelvis), pelvis (relative to the room, rather than a relative position between body segments), hip (pelvis relative to the thigh) and knee (thigh relative to the lower leg) were time normalised to cycle duration in 2 % increments (51 data points from 0-100 %) for the right sided cycles of both WALK and FRED conditions. Angular range of motion (ROM) was calculated as the maximum minus the minimum joint angle achieved within one cycle. This was done for each of the six trials and averaged within each participant, and then between all participants in both conditions. The mean angular position of each segment or joint was determined as the average of each angle throughout the gait cycle for WALK and FRED. The difference in mean angular positions, or offset, between WALK and FRED was calculated.

6.2.5 Statistical Analysis

Data for each variable were checked for normality of distribution using Q-Q and box plots. For variables that were normally distributed, paired samples t-tests were used to compare ROM and mean angular position between conditions with significance set at $p < 0.05$. For variables that were not normally distributed, Wilcoxon signed rank tests

were used. Confidence intervals (95 %) were also calculated for each pairwise comparison. All statistical analyses were performed using SPSS (version 19).

6.3 Results

6.3.1 Spatiotemporal Characteristics

All spatiotemporal data were normally distributed. Statistically significant differences were observed in all six spatiotemporal parameters (Table 6-1). The FRED condition was characterised by reduction in cadence ($t(15) = 21.220$, $p < 0.001$), stride length ($t(15) = 14.041$, $p < 0.001$), stride duration ($t(15) = 26.380$, $p < 0.001$), speed ($t(15) = 20.506$, $p < 0.001$), and effective 'stance' phase ($t(15) = 15.354$, $p < 0.001$) compared to those observed during WALK. Step width was significantly greater in the FRED condition compared to WALK ($t(15) = 2.662$, $p < 0.05$).

Table 6-1. Spatiotemporal characteristics of overground walking and FRED exercise in standing.

Gait Parameter	Overground Walking		Exercise Device		Mean Difference	
	Mean	±1SD	Mean	±1SD	(95 % CI)	*P* value
Cadence (steps·min^{-1})	110.7	7.2	71.3	2.7	-39.4 (-43.4 to -35.45)	<0.001*
Stride Length (m)	1.41	0.09	1.10	0.00	-0.31 (-0.35 to -0.26)	<0.001*
Stride Duration (s)	1.09	0.07	1.69	0.06	0.60 (0.55 to 0.65)	<0.001*
Speed (m·s^{-1})	1.30	0.13	0.65	0.03	-0.65 (-0.71 to -0.58)	<0.001*
Step Width (m)	0.20	0.03	0.23	0.05	0.03 (0.01 to 0.06)	0.018*
Stance Phase (%)	59.54	1.66	49.45	2.26	-10.09 (-11.49 to -8.69)	<0.001*

* indicates a significant difference between conditions at the $p < 0.05$ level

6.3.2 Kinematics

All angular ROM data were normally distributed with the exception of the hip in the transverse plane. Angular ROM was found to be similar between FRED and WALK

conditions for the trunk in the sagittal ($t(15) = 1.622$, $p = 0.126$) and frontal ($t(15) = 1.203$, $p = 0.248$) planes. It was also similar for the pelvis in the sagittal ($t(15) = 1.607$, $p = 0.129$) and frontal ($t(15) = 0.213$, $p = 0.834$) planes. In the transverse plane, ROM was significantly reduced for the trunk ($t(15) = 8.513$, $p < 0.001$) and the difference between FRED and WALK approached statistical significance in the pelvis ($t(15) = 1.854$, $p = 0.083$) (Table 6-2).

Table 6-2. Angular range of motion of the trunk, pelvis, hip, and knee in all three planes during overground walking and using the exercise device, also including the mean difference between the two conditions. (SD = standard deviation, CI = confidence interval)

Gait Parameter	Overground Walking		Exercise Device		Mean Difference	
	Mean	±1SD	Mean	±1SD	(95 % CI)	*P* value
Sagittal Plane						
Trunk	3.93	1.80	3.01	1.67	-0.92 (-0.29 to 2.14)	0.126
Pelvis	2.89	0.78	3.69	1.91	0.8 (-1.86 to 0.26)	0.129
Hip	42.54	3.96	33.38	2.28	-9.16 (6.50 to 11.81)	<0.001*
Knee	59.88	4.03	45.22	6.02	-14.66 (10.97 to 18.36)	<0.001*
Frontal Plane						
Trunk	12.59	3.26	11.21	4.42	-1.39 (-1.07 to 3.84)	0.248
Pelvis	8.29	3.33	8.09	2.70	-0.20 (-1.85 to 2.26)	0.834
Hip	12.67	3.44	8.77	4.64	-3.90 (0.67 to 7.14)	0.021*
Knee	16.50	5.91	9.42	5.22	-7.08 (4.93 to 9.22)	<0.001
Transverse Plane						
Trunk	12.55	3.85	3.92	1.14	-8.63 (6.47 to 10.79)	<0.001
Pelvis	12.00	3.28	9.25	4.18	-2.75 (-0.41 to 5.92)	0.083
Hip	16.93	7.34	8.87	2.73	-8.06 (4.95 to 11.17)	<0.001[a]*
Knee	20.66	5.37	10.59	3.96	10.07 (7.06 to 13.09)	<0.001*

[a] indicates that these data were not normally distributed. * indicates a significant difference between conditions at the $p < 0.05$ level

All mean angular position data were normally distributed with the exception of the pelvis and hip in the transverse plane. The pelvis was significantly tilted anteriorly for the FRED condition compared to WALK with an offset of 6.49° ($t(15) = 4.697$, $p < 0.001$) (Table 6-3). Hip ROM was significantly reduced in the FRED condition compared to WALK in the sagittal ($t(15) = 7.359$, $p < 0.001$), frontal ($t(15) = 2.572$, p

= 0.021) and transverse (Z = 3.516, p < 0.001) planes (Table 6-2). Knee ROM was also reduced in FRED in the sagittal (t(15) = 8.463, p < 0.001), frontal (t(15) = 7.041, p < 0.001) and transverse (t(15) = 7.120, p < 0.001) planes. The hip (t(15) = 13.297, p < 0.001) and knee (t(15) = 19.878, p < 0.001) were both more flexed throughout the gait cycle in the FRED condition than in WALK, with offsets of 22.31° and 24.11°, respectively, which were significant (Table 6-3). Despite the reduced ROM, peak knee and hip angles occurred at a similar point in the gait cycle for WALK and FRED.

Table 6-3. Mean angular position of the trunk, pelvis, hip and knee in all three planes during overground walking and exercise in the standing position on the device.

Gait Parameter	Overground Walking		Exercise Device		Mean Difference	
	Mean	±1SD	Mean	±1SD	(±95 % CI)	*P* value
Sagittal Plane						
Trunk	-5.37	6.15	-5.43	6.66	0.06 (-3.44 to 3.56)	0.970
Pelvis	9.06	4.06	15.55	6.18	-6.49 (-9.43 to -3.54)	<0.001*
Hip	18.30	5.56	40.61	6.62	-22.31(-25.88 to -18.73)	<0.001*
Knee	26.28	4.62	50.39	6.69	-24.11 (-26.69 to -21.52)	<0.001*
Frontal Plane						
Trunk	-0.41	1.68	0.53	2.05	-0.94 (-2.27 to 0.38)	0.150
Pelvis	-0.25	1.23	-0.33	1.83	0.08 (-0.69 to 0.85)	0.827
Hip	-0.14	2.02	-0.91	2.33	0.77 (-0.13 to 1.67)	0.088
Knee	2.99	3.92	0.44	7.53	2.55 (-0.37 to 5.48)	0.082
Transverse Plane						
Trunk	-2.11	1.86	-1.70	2.03	-0.41 (-1.25 to 0.42)	0.311
Pelvis [a]	-0.65	2.26	-1.63	2.93	0.98 (-0.05 to 2.01)	0.056
Hip [a]	8.77	8.35	2.55	5.59	6.22 (1.77 to 10.66)	0.010*
Knee	-8.77	9.14	1.16	8.58	-9.94 (-12.45 to -7.42)	<0.001*

[a] indicates that these data were not normally distributed. * indicates a significant difference between conditions at the p < 0.05 level

6.4 Discussion

The aim of the investigation reported in this chapter was to compare lower limb and trunk kinematics during FRED exercise (FRED), and overground walking (WALK). The key findings of this study were that the lumbopelvic region was at least as stable whilst exercising on the FRED as in overground walking. In the transverse plane, reduced ROM was observed during FRED compared to WALK. This stable lumbopelvic region was achieved over a dynamically moving base of support, where the ROM of the knees and hips was lower in FRED than in WALK. All spatiotemporal variables were significantly reduced in FRED compared to WALK, suggesting a slower, more controlled motion. Trunk motion in the sagittal and frontal planes demonstrated similar ranges for both FRED and WALK. In the transverse plane, a reduced ROM was observed for FRED. Similar observations were made for the pelvis in terms of ROM, although in the transverse plane, a smaller reduction in range of motion was found for WALK.

As a fundamental human activity, walking has previously been investigated as an intervention strategy in the treatment of LBP (Joffe et al., 2002; Mirovsky et al., 2006; Taylor, Evans, & Goldie, 2003; Torstensen et al., 1998). However, heterogeneity of study design and methodological quality have contributed to inconsistent findings (Hendrick et al., 2010). Of these studies only Torstensen et al. (1998) and Taylor et al. (2003) used walking independently, while Joffe et al. (2002) and Mirovsky et al. (2006) combined walking with bodyweight support and traction, respectively. Notwithstanding the lack of evidence supporting walking as an effective intervention strategy for low back pain, the movement itself, involving control of trunk and pelvis motion during lower limb movements, is known to contribute to recruitment of the

TrA and LM (Saunders, Rath, & Hodges, 2004; Saunders et al., 2005). Importantly, walking tends to be advocated by health care professionals in line with recommendations that ordinary physical activities should be continued as much as possible in order to aid recovery from LBP and prevent long-term disability (van Tulder et al., 2000).

Similarities observed in both trunk and pelvic ROM between FRED and WALK in the sagittal and frontal planes suggest that the exercise device may be similar to walking, in terms of enabling tonic recruitment of the TrA and LM. Previously, Saunders et al. (2004; 2005) reported tonic TrA but phasic LM activity at walking speeds comparable to those reported here. However, no data were presented describing changes in activity amplitude, if any, within each gait cycle. The phasic activity of LM previously reported during walking (Saunders, Rath, & Hodges, 2004) could be a factor explaining the questionable effectiveness of walking as a successful intervention for LBP (Hendrick et al., 2010). The reduced transverse ROM, in FRED compared with WALK seen in the current study could further indicate facilitation of greater tonic activity of the local lumbopelvic muscles (Richardson & Jull, 1995) when using the FRED than in overground walking. If this reduced axial rotation results in more tonic recruitment of LM at a segmental level, then this could mean that FRED exercise is a more successful intervention for LBP than walking. Current research within our group is exploring differences in lumbopelvic muscle recruitment between the exercise device and walking using intramuscular electromyography. Future studies in symptomatic populations are required to examine the clinical effectiveness of FRED exercise.

No angular offsets were found between FRED and WALK for the trunk or pelvic position in all three planes, with the exception of a greater degree of anterior pelvic tilt

in the FRED condition. Influences of anterior pelvic tilt (O'Sullivan et al., 2006b) and accompanying lordotic spinal posture (Claus et al., 2009), similar in magnitude to that observed within this investigation, have previously been shown to recruit both the superficial and deep fibres of the LM to approximately 30-40 % of maximal voluntary isometric contraction capabilities, a range known to facilitate deep muscle recruitment (McArdle, Katch, & Katch, 1991). Thus, this angular offset could be beneficial for the recruitment of the LM, provided care is taken to avoid over-recruitment of the superficial fibres of LM.

Hip and knee joints were more flexed throughout the gait cycle in FRED than during WALK. The increase in hip flexion was partly due to the angular definition being relative to a perpendicular axis of the pelvis. Therefore, the observed increase in anterior tilt creates a greater degree of flexion at the hip. The increased flexion of the knee throughout the motion cycle during FRED is linked to the reduced stride length that was caused by the mechanical constraints of the device. As a result of reduced stride length, and being instructed to minimise cephalad/caudad excursion, participants did not reach full knee extension during the 'stance phase' of the motion cycle on FRED, as is normally seen in walking. What was apparent for knee and hip motion in the sagittal plane, was that the change in angle throughout the gait cycle showed a more sinusoidal pattern in FRED compared to WALK. This more regular movement pattern could contribute, to some extent, to more continuous/tonic muscle recruitment, a key training requirement of the deep lumbopelvic muscles (Richardson & Jull, 1995).

In recent years there has been a drive for training interventions for the muscles of the lumbopelvic region to be made more functional (Hodges, 2011). A number of studies have brought into question the transferability of any training effects seen following

less functional activities such as gym ball training where the base of support is simply unstable (Drake et al., 2006). Debuse et al. (2013) demonstrated that the local lumbopelvic muscles were recruited to a greater extent with lower limb movement and an unstable base of support than with standing still on an unstable base of support (i.e. no voluntary lower limb movement). While overground walking involves lower limb movement, it does not usually involve an unstable base of support. During FRED exercise the requirement to control the descent of the "front" leg by gradually unloading the "back" leg within each motion cycle may result in greater recruitment of the local lumbopelvic muscles than overground walking.

6.5 Limitations

The study reported in this chapter has a number of limitations. Firstly, it examined relative motion between the pelvis and trunk. In order to gain a better understanding of how FRED exercise influences the kinematics of the lumbopelvic region, a more detailed model of the thoracic and lumbar spine is needed. This would enable vertebral motion to be evaluated at a segmental level. Participants were asked to walk at their preferred walking speed. Due to the nature of the exercise device, movements were slower compared to walking. Saunders et al. (2005) reported reduced axial rotation of the spine when walking slower. Thus, slow walking could lead to similar kinematics that were observed for the exercise device, and this should be explored further. However, walking slower does not involve an unstable base of support, or the complex motor control associated with FRED exercise, both of which could contribute to increased local lumbopelvic muscle recruitment.

6.6 Conclusion

Key differences between FRED exercise and overground walking included reduced transverse plane range of trunk motion with respect to the pelvis, a more anteriorly tilted pelvis, reduced stride length, and reduced knee and hip range of motion in the sagittal plane. The greater anterior tilt of the pelvis potentially moved the pelvis into a more advantageous position for the recruitment of TrA and LM. In Chapter IV, this distinction was not observed as TrA contraction was similar during both treadmill walking and FRED exercise and LM activation was greater during overground walking. However, it should be noted that due to the nature of the study design in Chapter IV, contraction of both muscles was only assessed at one key point in the movement cycle; at heel strike in treadmill walking and the equivalent position during FRED exercise. Therefore, future studies should attempt to address this limitation with continuous assessment of muscle activity during both conditions.

CHAPTER VII

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Examination of Key Lumbopelvic Muscle Activity during FRED Exercise and Walking using Surface Electromyography

7.1 Introduction

In recent years there has been a drive to make rehabilitative interventions more functional in nature (Hodges & Cholewicki, 2007; Hodges et al., 2013; Richardson, Hodges, & Hides, 2004) and whilst some exercise approaches have been shown to recruit LM and/or TrA, many are limited in their functional relevance to activities of daily living such as walking and maintaining an upright posture against gravity. Many of these interventions also require conscious voluntary contraction from the participant to activate the local muscles which is known to be difficult (Van, Hides, & Richardson, 2006). As arguably the most functionally relevant activity to most people, walking has been suggested as a potential therapeutic intervention for LBP (Joffe et al., 2002; Taylor, Evans, & Goldie, 2003). However, conflicting evidence has been presented in the literature, and a recent meta-analysis suggested that walking is not effective in this context (Hendrick et al., 2010). For a functionally relevant intervention to be developed, however, it should consider the key elements of walking. In its simplest form, walking consists of a relatively stable upper body positioned above a moving base of support.

Debuse et al. (2013) demonstrated that FRED exercise recruited both LM and TrA without a conscious/voluntary contraction. FRED exercise was compared to overground walking using three-dimensional motion analysis in Chapter VI and was found to achieve a more stable lumbopelvic region, as apparent by attenuated axial rotation of the spine. This more stable lumbopelvic region could be indicative or greater tonic muscle activity which has been shown previously to be a key characteristic for training local muscles (Richardson & Jull, 1995). However, the

influence of the exercise device on lumbopelvic muscle activity throughout a complete foot movement cycle, or in comparison to walking, has not yet been determined.

The aims of the investigation reported in this chapter were to determine (1) lumbopelvic muscle recruitment over a complete foot movement cycle, and (2) differences in lumbopelvic muscle activity between overground walking and FRED exercise.

7.2 Method

7.2.1 Participants

Fifteen healthy adult male volunteers (mean ± SD age: 24.93 (± 3.92) years, body mass: 83.03 (± 7.21) kg, height: 1.78 (± 0.05)) with no recent history of LBP, gait impairments, or other conditions affecting their ability to walk or exercise, agreed to participate in this study. Approval for this study was gained from the Ethics Committee of the School of Life Sciences at Northumbria University, Newcastle upon Tyne, England, and all participants gave their fully informed (Appendix P) written consent (Appendix Q) to take part.

7.2.2 Experimental Protocol

Participants completed an overground walking (WALK) condition and a FRED exercise (FRED) condition in a counterbalanced random order within a single session. In the WALK condition participants were asked to walk along a level 7.5 m walkway, instrumented with embedded force plates (OR6-7, AMTI, Watertown, Massachusetts, USA), at a self-selected comfortable speed. Starting positions were adjusted individually to ensure that 'clean' foot contacts with the force plates could be achieved

without direct targeting by the participant. A minimum of 10 trials were completed, before six trials - without evidence of targeting - were randomly selected for subsequent analysis.

Prior to data collection for FRED exercise, participants were given an initial five-minute period to familiarise themselves with the device. During this time, they were also given guidance as to the intended use with instructions to keep their feet in contact with the foot plates at all times, and to self-select a movement frequency that allowed them to achieve a smooth controlled movement with minimal up/down excursion of the torso (Debuse et al., 2013). Following this, data were collected during FRED exercise in standing for 30 seconds. Subsequently, six cycles were chosen at random for analysis. A cycle was defined, as per previous chapters, as one complete revolution of the feet beginning and ending with their right foot at the most anterior point in the cycle.

7.2.3 Equipment

For the WALK trials, heel strike and toe off was identified by two force platforms (OR6-7, AMTI, Watertown) embedded within the walkway and flush with the surface in the centre of the laboratory. Raw data signals were amplified (gain=1000, MSA-6, AMTI, Watertown), and sampled at 2000 Hz by a data acquisition card (MX, Vicon Motion Systems, Oxford).

For the FRED trials, 3-dimensional trajectories of retro-reflective markers (ø = 14mm) placed on the side of the right foot plate were tracked and sampled at 200 Hz using a 12-camera motion capture system (MX T20, Vicon Motion Systems, Oxford, UK). The beginning and end of each movement cycle were determined by the most anterior

and posterior positions of these markers. As in the kinematics study reported in Chapter VI, rearward movement of the foot plate (i.e. when the foot progressed backwards under the body) was considered comparable to the stance phase of the walking gait cycle. Similarly, forward movement of the foot plate (i.e. when the foot progressed forwards under the body) was considered comparable to the swing phase of the walking gait cycle.

Myoelectric activity of the anterolateral abdominal, lumbopelvic and lower limb musculature were collected using surface EMG (sEMG). Electrodes were placed unilaterally on the right *lumbar multifidus* (LM), *erector spinae* (ES), *internal oblique* (IO), *external oblique* (EO), *rectus abdominis* (RA), *vastus medialis* (VM), *biceps femoris* (BF), *medial gastrocnemius* (MG) and the *tibialis anterior* (TA).

Prior to electrode placement, all sites were shaved and exfoliated using abrasive gel (Nuprep, Weaver & Company, Bromley, UK). Surface contaminants were then removed with isopropyl alcohol swabs so that skin impedance was less than 5 kΩ. Circular self-adhesive Ag/AgCL electrodes (diameter = 34 mm, sensing area = 13.2 mm^2, measurement area = 154 mm^2) with a conductive wet gel (Blue Sensor S, Ambu, Ballerup, Denmark) were placed in a bipolar configuration (Table **7-1**) in accordance with existing protocols (Hermens et al., 2000; Ng, Kippers, & Richardson, 1998) with an inter-electrode distance of 20 mm. Although potential for EMG signal crosstalk exists in each muscle, previous investigations using the same electrode placement configuration have shown this to be insignificant for the anterolateral abdominal muscles (Cholewicki & McGill, 1996; Cholewicki & Van Vliet 2002; Floyd & Silver, 1950) and between the spinal extensors (Vink, Van Der Velde, & Verbout, 1987).

Additionally, a series of signal integrity verifications were performed to further minimise any potential effect (Table **7-1**).

Table 7-1. Surface electromyography electrode placement guidelines.

Muscle	Placement	Activity integrity confirmation procedure
LM	Parallel to a line connecting the posterior superior iliac spine and L1-L2 interspinous space at the level of the L5 spinous process. (O'Sullivan et al., 2006b)	Trunk extension and full forward flexion. Locations amended if electrical silence in ES with continued activity in LM was not observed at full flexion
ES	*Iliocostalis lumborum pars thoracis*. Midway between the midline and axillary line of the torso at the level of the L1 spinous process (O'Sullivan et al., 2006b)	
IO	1-cm medial to the anterior superior iliac spine (O'Sullivan et al., 2006b)	Lift right shoulder and point it towards left hip (Goldman et al., 1987)
EO	Just below the rib cage, along a line connecting the most inferior costal margin and the contralateral pubic tubercle (O'Sullivan et al., 2006b)	Lift left shoulder and point it towards right hip (Goldman et al., 1987)
RA	1 cm above the umbilicus and 2-cm lateral to midline in a vertical orientation (O'Sullivan et al., 2006b)	Raise both legs from a supine (Goldman et al., 1987)
VM	5-cm from the superior medial side of the patella along a line medially oriented at an angle of 50◦ with respect to the anterior superior iliac spine(Rainoldi, Melchiorri, & Caruso, 2004)	Seated isometric knee extension
BF	35 % distance from the ischial tuberosity to the lateral side of the popliteus cavity, starting from the ischial tuberosity (Rainoldi, Melchiorri, & Caruso, 2004)	Prone isometric knee flexion
MG	At 50 % of the distance between the medial side of the popliteus cavity to the medial side of the Achilles tendon insertion, starting from the Achilles tendon (Rainoldi, Melchiorri, & Caruso, 2004)	Standing plantarflexion of the ankle
TA	15 % percentage distance from the tuberosity of tibia to the inter-malleoli line, starting from the tuberosity of tibia (Rainoldi, Melchiorri, & Caruso, 2004)	Seated isometric dorsiflexion of the ankle

LM, *lumbar multifidus*; ES, *erector spinae*; IO, *internal oblique*; EO, *external oblique*; RA, *rectus abdominis*; VM, *vastus medialis*; BF, *biceps femoris*; MG, *medial gastrocnemius*; TA, *tibialis anterior*

Raw signals from each muscle were pre-amplified (gain=1000, common mode rejection ratio >100dB) and sent telemetrically with a fixed latency of 16ms to a data receiver (Myon RFTD-E16, Myon AG, Baar, Switzerland) before being amplified and sampled at 2000Hz by a data acquisition card (MX, Vicon Motion Systems, Oxford, UK). All data were collected synchronously and stored in specialist software (Nexus 1.7, Vicon Motion Systems, Oxford, UK) for subsequent analysis.

7.2.4 Data Processing and Reduction

Myoelectric data for each muscle were processed within Nexus (1.7, Vicon Motion Systems, Oxford), using the ProEMG plugin (ProEMG, Pro Physics AG, Zurich, Switzerland). All channels were band-pass filtered (Butterworth 2nd order, 10Hz–350Hz), linear enveloped with a root mean square (RMS) with a fixed window width of 40ms (Polcyn et al., 1998) and time normalised to one complete right gait/movement cycle. All signals for both WALK and FRED movement cycles were then amplitude normalised to the peak RMS EMG amplitude from the WALK trials for each muscle (Stackhouse et al., 2007). Time and amplitude normalised RMS EMG signals for each muscle for each participant were then ensemble averaged for either WALK (six gait cycles for the right leg) or FRED (six right leg movement cycles). Mean EMG was calculated from the amplitude and time normalised RMS EMG curves for each muscle.

Baseline EMG for each muscle, collected during a 30s trial where the participant was supine and all muscles were relaxed, was filtered as described above for WALK and FRED trials, before being full-wave rectified. The mean and standard deviation (SD) of the rectified baseline EMG for each muscle was determined. The timing of muscle activity onset and cessation were then determined by the points at which the RMS

EMG signals raised above or dropped below the mean plus two SDs of the baseline rectified signal, respectively (Morey-Klapsing, Arampatzis, & Bruggemann, 2004). The proportion of the gait cycle that each muscle was active in WALK and FRED was subsequently determined as the duration of activity over the total cycle duration.

7.2.5 Statistical Analysis

Data for mean RMS EMG and the proportion of the gait cycle that each muscle was active were checked for normality using the Shaprio-Wilk test, Q-Q plots and box plots. For variables that were normally distributed, mean EMG and activity duration data were compared between WALK and FRED using paired samples t-tests. For data that were not normally distributed, WALK and FRED were compared using Wilcoxon signed rank tests. Confidence intervals (95 %) were also determined for each variable for each muscle between WALK and FRED. The level of significance was set at 95 % ($p < 0.05$) for all data, and data were analysed using PASW Statistics v.18 (SPSS Inc., Chicago, Illinois, United States).

7.3 Results

Data not normally distributed for mean RMS EMG included LM, ES, TA and BF, and for the proportion of the gait/movement cycle the muscle was active included IO, RA, TA and VM.

During walking, LM showed phasic patterns of activity with peaks in activity around the start and midway through the gait cycle (Figure **7-1**). In FRED exercise a more tonic pattern of activity was seen. Mean LM activity (Table 7-2) during FRED exercise

was significantly greater than in OW (Z = -3.067, p = 0.020), as was the proportion of the gait cycle for which LM was active (t(14)=6.618, p < 0.001) (Table 7-3).

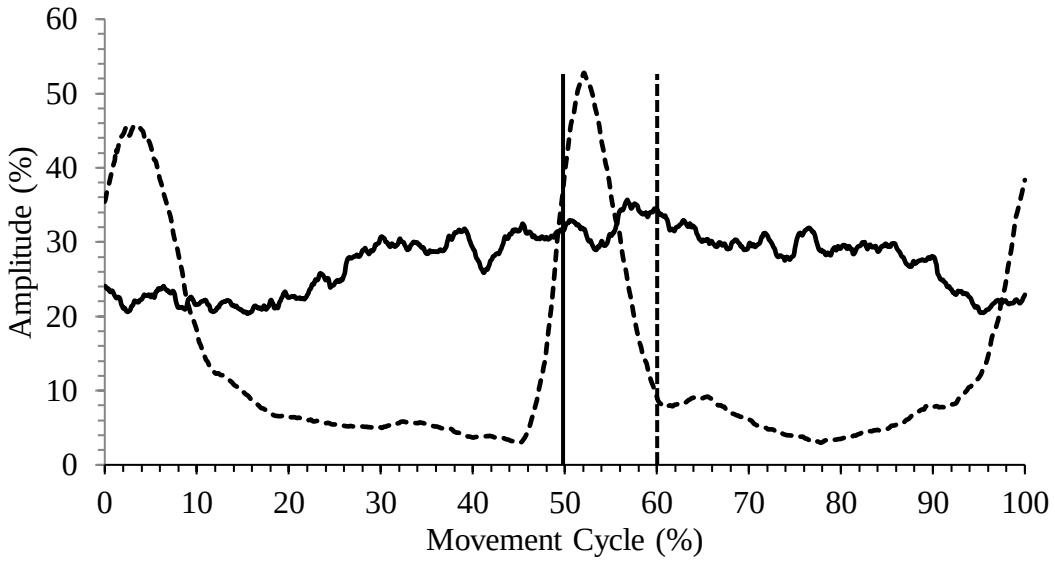

Figure 7-1. Normalised RMS EMG shown for *lumbar multifidus* in walking (---) and exercise device (—) conditions, over one gait cycle

Like LM, ES showed peaks of muscle activity at the start and approximately half way through the gait cycle in WALK (Figure 7-2). In FRED, ES muscle activity was lowest at the start and end of the movement cycle, increasing gradually towards maximum activity around half way through the movement cycle. Unlike WALK, no discernible peaks in activity were observed for ES during FRED trials. Mean ES activity (Table 7-2) was significantly greater in FRED compared to WALK (Z = -2.897, p = 0.004), and ES was active for a significantly greater proportion of the gait cycle during FRED (t(14) = 3.313, p = 0.005) (Table 7-3).

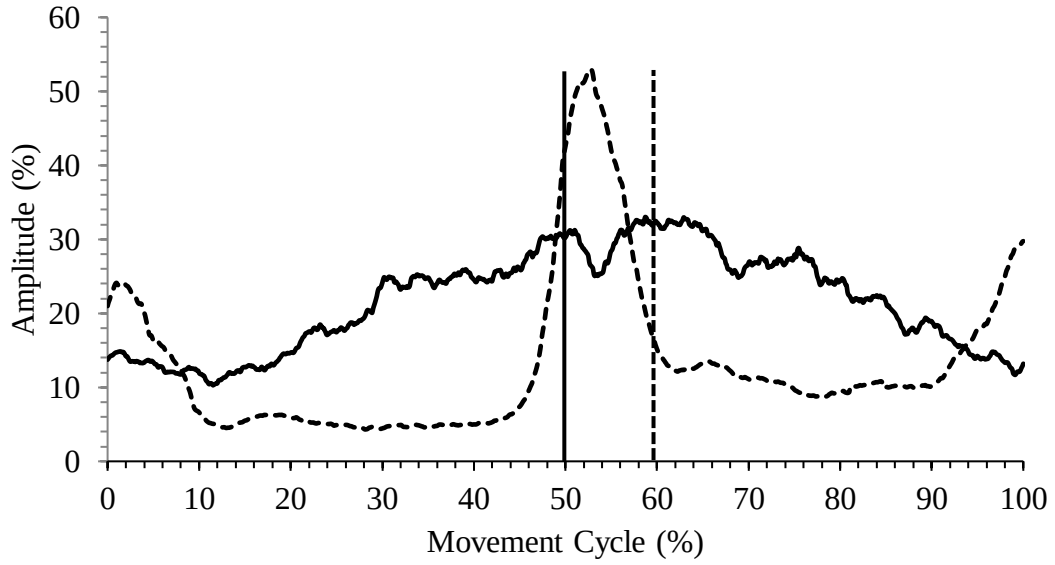

Figure 7-2. Normalised RMS EMG shown for *erector spinae* in walking (---) and exercise device (—) conditions, over one gait cycle

Similar to LM and ES, IO showed a change from phasic activity during WALK to more tonic activity during FRED (Figure **7-3**). The mean level of activity in the IO was significantly lower in FRED than in WALK ($t(14) = 4.694$, $p < 0.001$) (Table 7-2), although both conditions showed IO activity above baseline levels for the majority of the gait cycle, with no difference seen between conditions ($Z = -0.764$, $p = 0.445$) (Table 7-3).

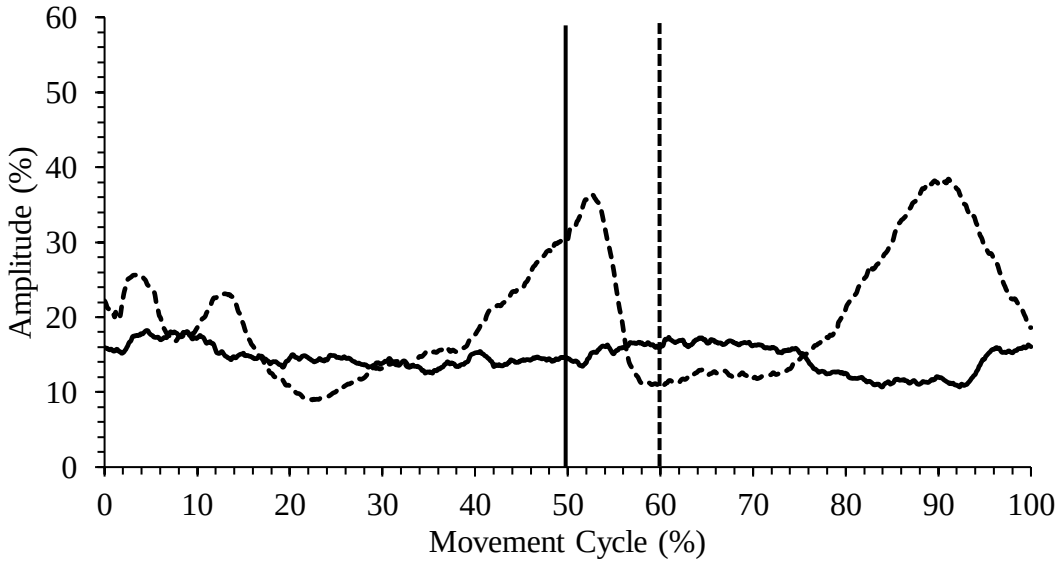

Figure 7-3. Normalised ensemble RMS EMG shown for *internal oblique* in walking (---) and exercise device (—) conditions, over one gait cycle.

During WALK, peaks in EO activity were seen at, or just prior to, half way through and towards the end of the gait cycle (Figure **7-4**), although these peaks were not as prominent as those seen in LM and ES (Figure **7-1** and Figure 7-2). In FRED, more tonic activity was observed in EO than in WALK. Mean EO activity was similar between WALK and FRED ($t(14) = 1.931$, $p = 0.074$) (Table 7-2), although it was active for a significantly reduced proportion of the movement cycle in FRED ($t(14) = 2.741$, $p = 0.016$) (Table 7-3).

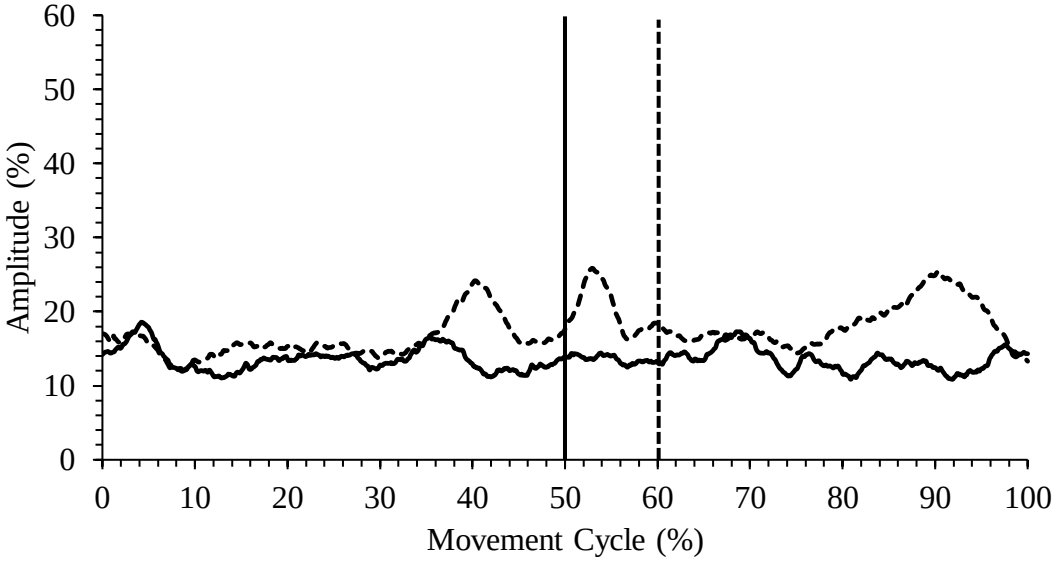

Figure 7-4. Normalised ensemble RMS EMG shown for *external oblique* in walking (---) and exercise device (—) conditions, over one gait cycle.

Walking resulted in three slight peaks in RA activity at the start, just before half way through, and towards the end of the gait cycle (Figure 24C). During FRED, a more tonic pattern of RA activity was observed with no discernible peaks in activity. Mean RA EMG activity was significantly reduced in FRED compared to WALK ($t(14) = 4.164$, $p = 0.001$) (Table 7-2). Despite the apparent tonic activation of RA in FRED, the level of activity did not exceed the threshold defined for muscle activation, and

was therefore active for a significantly reduced proportion of the gait/movement cycle compared to WALK (Z = 2.803, p = 0.005) (Table 7-3).

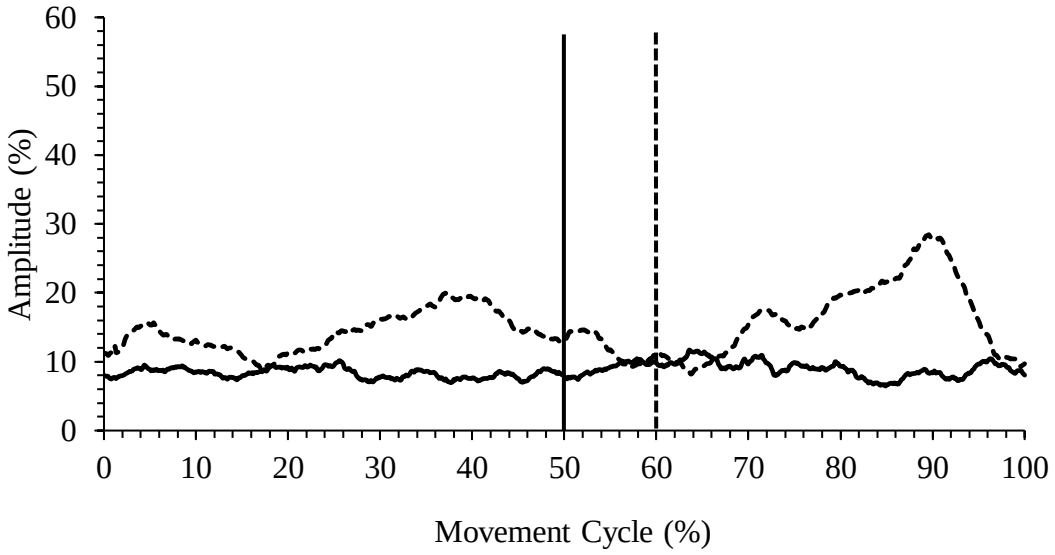

Figure 7-5. Normalised ensemble RMS EMG shown for *rectus abdominis* in walking (---) and exercise device (—) conditions, over one gait cycle.

In WALK, TA and MG were active for similar proportions of their respective movement cycles (Table 7-3) when compared to FRED exercise, although mean RMS amplitude was significantly greater (Table 7-2) in WALK for both TA (Z = -2.045, p = 0.041) and MG (t(14) = 4.100, p = 0.001). Bi-phasic activity was observed in TA in WALK with discernible peaks present at toe-off and in preparation for heel-strike. Only mono-phasic muscle activity was seen in TA during FRED with the lowest activity observed at the start and end of the movement cycle, increasing gradually towards maximum activity around half way through (Figure **7-6**A). In MG mono-phasic activity was observed in both conditions with increases in activity seen during the first half of the respective cycles only (Figure **7-6**B).

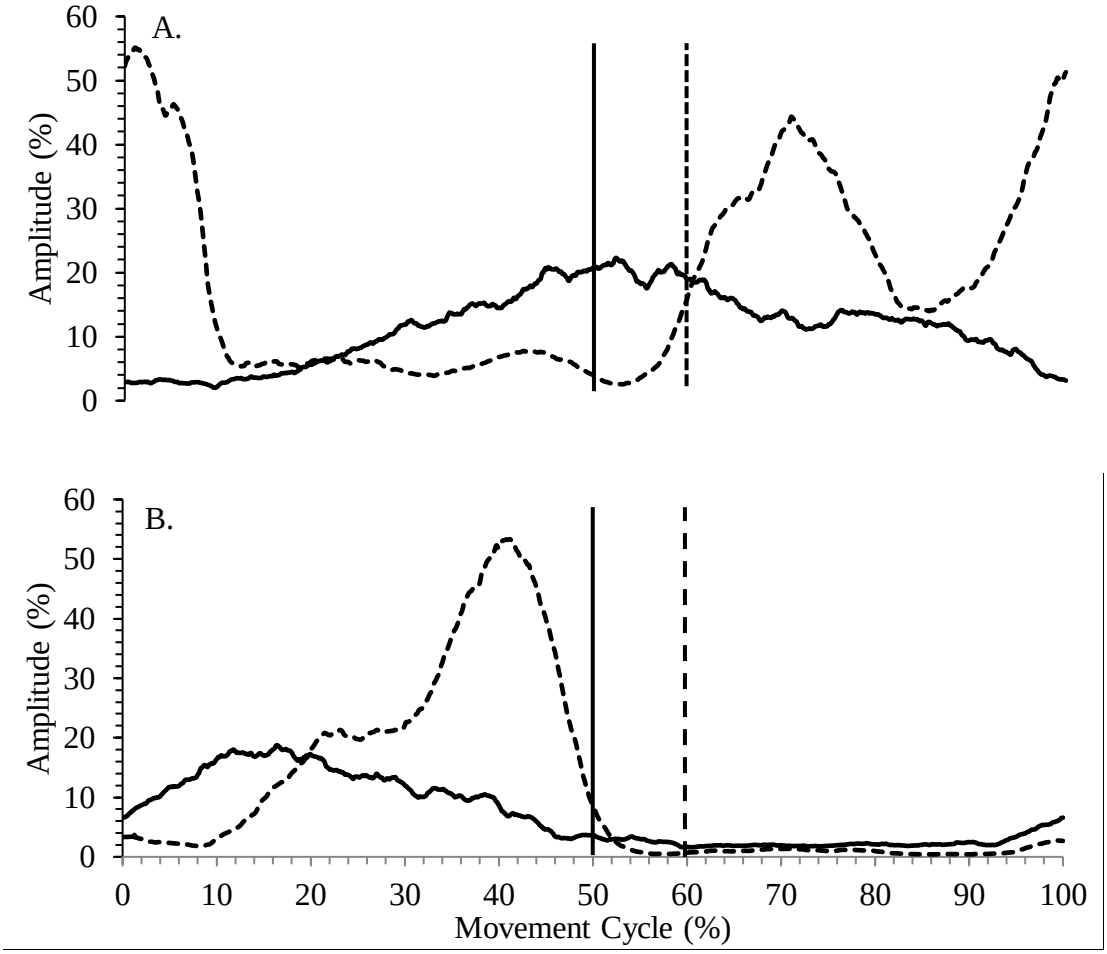

Figure 7-6. Normalised ensemble RMS EMG shown for the *tibialis anterior* (A) and *medial gastrocnemius* (B) in walking (---) and exercise device (—) conditions, over one movement cycle.

In FRED, VM ($Z = -3.233$, $p = 0.001$) and BF ($Z = -3.046$, $p = 0.009$) were active for a significantly greater proportion of their respective movement cycles (Table 7-3), although mean RMS amplitude (Table 7-2) was only greater in the VM ($t(14) = -5.805$, $p < 0.001$). Patterns of activation were similar for BF in both conditions, though with noticeably greater increases in activity during walking in preparation for the stance phase (Figure **7-7**B). Patterns of activation for VM were clearly different between conditions with a peak of activity just after heel strike during WALK that does not exist during FRED exercise, followed by a period of very low activity. During FRED exercise a gradually building very high peak of activity was observed which coincided

with the approach to the beginning of the movement cycle (i.e. when the foot was approaching its most forward position) (Figure **7-7**A).

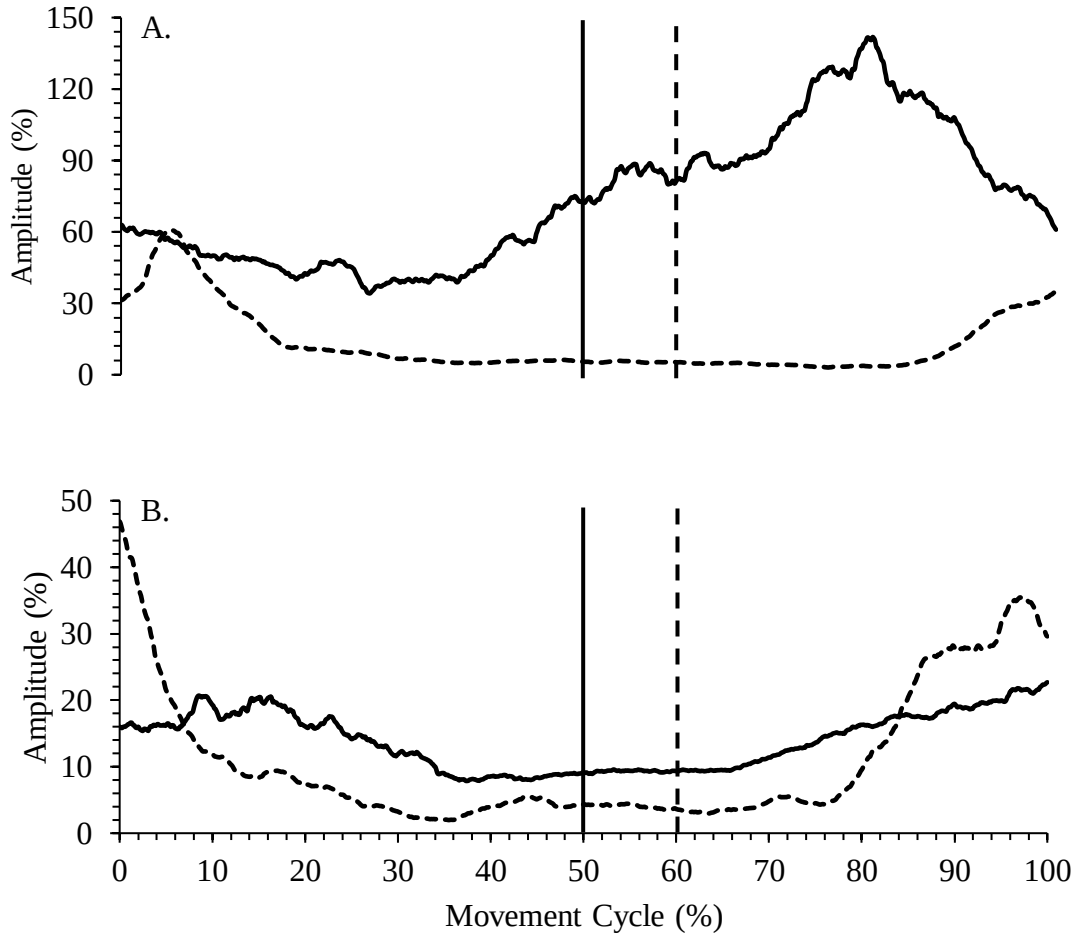

Figure 7-7. Normalised ensemble RMS EMG shown for the *vastus medialis* (A), and *biceps femoris* (B) in walking (---) and exercise device (—) conditions, over one movement cycle.

Table 7-2. Mean RMS EMG amplitude over one gait cycle

Muscle		Overground Walking		Exercise Device		Mean Difference †	95 % Confidence Interval		P value
		Mean	*SD*	Mean	*SD*		Lower	Upper	
Lumbopelvic	LM	13.5	3.98	27.38	15.61	13.87	4.92	22.82	0.020*
	ES	9.42	4.13	21.70	17.92	12.27	1.84	22.70	0.004*
	IO	19.96	4.45	14.62	4.82	-5.35	-7.79	-2.90	<0.001*
	EO	17.42	8.67	13.55	8.11	-3.87	-8.16	0.43	0.074
	RA	15.06	3.82	8.67	5.62	-6.34	-9.69	-3.10	0.001*
Lower Limb	TA	18.54	3.20	11.43	10.62	7.11	-13.30	-0.97	0.026*
	MG	11.15	3.13	7.06	4.82	4.09	-6.23	-1.95	0.001*
	VM	13.45	4.96	74.17	39.02	-60.72	38.3	83.2	<0.001*
	BF	11.11	2.41	11.31	6.57	-0.20	-2.71	3.11	0.884

† - positive mean difference indicates increase in exercise device condition compared to walking; * = significant at $p < 0.05$ level; LM = *lumbar multifidus*, IO = *internal oblique*, ES = *erector spinae*, EO = *external oblique*, RA = *rectus abdominis*.

Table 7-3. Percentage of gait/movement cycle that each muscle is active

Muscle		Overground Walking		Exercise Device		Mean Difference†	95 % Confidence Interval		P value
		Mean	SD	Mean	SD		Lower	Upper	
Lumbopelvic	LM	33.42	21.61	78.52	30.52	45.10	30.48	59.72	<0.001*
	ES	17.69	12.44	49.56	41.94	31.87	11.24	52.50	0.005*
	IO	79.52	28.93	79.57	35.34	0.05	-7.54	7.65	0.445
	EO	33.93	35.66	8.51	12.88	-25.42	-45.32	-5.53	0.016*
	RA	18.16	23.61	0.00	0.00	-18.16	-31.79	-4.54	0.005*
Lower Limb	TA	74.87	30.12	63.25	37.12	-11.62	-31.9	8.64	0.239
	MG	46.83	8.60	53.50	25.71	6.67	-7.55	20.90	0.331
	VM	39.33	29.15	95.96	8.97	56.63	41.50	71.76	<0.001*
	BF	38.17	10.11	58.92	24.88	20.75	6.14	35.36	0.009*

† - positive mean difference indicates increase in exercise device condition compared to walking; * = significant at $p < 0.05$ level; LM = *lumbar multifidus*, IO = *internal oblique*, ES = *erector spinae*, EO = *external oblique*, RA = *rectus abdominis*.

7.4 Discussion

The aims of the study reported in this chapter were to identify lumbopelvic and lower limb muscle activity throughout a complete movement cycle during FRED exercise,

and to compare this to muscle activity during overground walking. The key findings of the study were that FRED exercise a) promoted tonic activity of the lumbopelvic musculature, as compared to WALK which resulted in phasic activity of the lumbopelvic muscles, b) resulted in greater trunk extensor than trunk flexor muscle activity as compared with WALK, and c) resulted in greater knee extensor activity as compared to WALK.

7.4.1 Phasic-to-Tonic Shift in Muscle Activity

During WALK, all muscles showed one or more distinct peaks in activity. For most muscles, these peaks occurred around the start of (heel strike), and midway through (just prior to toe off), the gait cycle. This is consistent with previous research that observed peaks in lumbopelvic muscle activity around heel strike and toe off in walking (Saunders, Rath, & Hodges, 2004). Saunders et al. (2004) observed phasic activity of LM (superficial and deep fibres), ES, IO and EO, where bursts of activity were associated with the need to maintain lumbopelvic stability at heel strike and toe off (Thorstensson et al., 1982), and the need to absorb impact forces at heel strike (Jonsson, 1970).

In the present study, during WALK, IO was active for the majority of the gait cycle, but showed biphasic modulation which has been linked to respiration and changes in trunk motion through the gait cycle (Saunders, Rath, & Hodges, 2004). During FRED, these peaks in activity were not apparent, suggesting a shift from phasic to tonic activity. Despite this shift from phasic to tonic activation of IO, the duration of IO activity was not different between WALK and FRED, with IO being active for the majority of the gait cycle. Saunders et al. (2004) also observed tonic activity in IO

during walking with multiple bursts of increased activity throughout the gait cycle, supporting our observations during WALK.

Lumbar multifidus and ES muscles were active for a significantly longer proportion of the gait/movement cycle during FRED exercise compared to walking. Chapter VI reported that FRED exercise results in a greater degree of anterior pelvic tilt compared to walking. Similar magnitudes of anterior tilt (O'Sullivan et al., 2006b) have been shown to recruit the deep and superficial fibres of LM to the levels required for optimal local muscle recruitment (30-40 % maximal voluntary contraction) (McArdle, Katch, & Katch, 1991). The findings presented here appear to support the notion that FRED exercise promotes optimal activation of LM for the promotion of lumbopelvic stability. The more tonic nature of ES activity in FRED throughout the movement cycle may suggest a favourable recruitment of these muscles during FRED exercise with overground walking.

Interestingly, the finding that FRED exercise promotes a phasic-to-tonic shift in lumbopelvic muscle activity compared with walking is also highly relevant to populations other than individuals with LBP, in particular, individuals recovering exposure to microgravity and long-term bed rest (LTBR) (Belavý et al., 2011; Belavý et al., 2007; Hides et al., 2007a; Hides et al., 2011a). Belavý et al. (2007) found a tonic-to-phasic shift of lumber ES activity during a lower limb perturbation (knee flexion-extension) activity following eight weeks of bed rest. This tonic-to-phasic shift in muscle recruitment patterns persisted in lumbar ES over the six-month follow-up period after the end of the bed rest trial and was even more exaggerated once participants returned to an upright posture. This lack of recovery in normal tonic activation of the spinal extensors after six months follow-up suggests that the nature

of muscle recruitment does not recover its pre-bedrest state without therapeutic intervention. Similarly, Hodges and Moseley (2003) showed a link between LBP and reduced tonic activity in the deep lumbopelvic muscles.

Walking has been proposed as a therapeutic intervention for LBP (Joffe et al., 2002). However, there is a lack of consensus about the effectiveness of walking in reducing LBP (Hendrick et al., 2010). The data presented here, and in previous literature (Saunders, Rath, & Hodges, 2004), suggest that the lack of improvement in LBP seen when using walking therapeutically could be due to the mainly biphasic activity of the lumbopelvic muscles. Richardson and Jull (1995) have argued that for optimal lumbopelvic stability, the deep lumbopelvic muscles need to be recruited tonically. The promotion of tonic activation seen during FRED exercise is likely to make FRED exercise more effective for rehabilitation following LBP, LTBR and long-term microgravity exposure than walking.

7.4.2 Promotion of Spinal and Knee Extensor Muscle Activity

Mean levels of muscle activity over one movement cycle were significantly increased in LM and ES in FRED compared to WALK. Conversely, mean activity in RA and IO were significantly reduced in FRED compared to WALK. Previous studies investigating the influence of pelvic tilt on LM activity (O'Sullivan et al., 2006b) showed that LM is much more likely to be recruited in a position of anterior pelvic tilt. This means that the level of anterior tilt seen when using the FRED (Chapter VI) could be optimal for recruitment of LM. This is also of great relevance to people recovering from LTBR and astronauts. Buckey (2006) described that the posture of astronauts is characterised by increased trunk and limb flexion and points to a selective atrophy of the spinal extensors. The same effect has also been noted following LTBR. Hides et

al. (2007a) investigated the influence of LTBR on lumbopelvic muscle size using magnetic resonance imaging, and found selective atrophy of the spinal extensor muscles, in particular LM. In fact, the reduction in LM cross-sectional area seen post LTBR is similar to its response to LBP (Hides, Richardson, & Jull, 1996; Hides et al., 1994). The spinal flexor muscles including *psoas*, *external oblique* and *rectus abdominis*, however, were found to increase their cross sectional area following LTBR (Hides et al., 2007a). What is likely to compound the problem further is that LM is not only atrophied following LTBR and long-term microgravity exposure, it is also in a stretched position (Belavý et al., 2008) which inhibits it from being recruited at all (Comerford & Mottram, 2001). The fact that FRED exercise results in a lumbopelvic position that is particularly conducive to effective LM recruitment and activity (O'Sullivan et al., 2006b) in asymptomatic volunteers, may offer an advantage in this respect.

In an attempt to address the clear need to prevent the spinal extensor muscle atrophy seen in LTBR, Belavý et al. (2008) investigated the use of combined vibration and resistive exercise. Changes in lumbopelvic muscle cross-sectional area were assessed during 8 weeks of LTBR and for six months following return to an upright posture. Significant atrophy of the spinal extensors was observed which was reduced, but not eradicated, with the use of resistive vibration exercise. Importantly, ES had recovered and, in fact, improved on, its pre-LTBR state by 28-days following the end of LTBR, while LM did not recover its pre-LTBR cross sectional area even after six months (Belavý et al., 2008). This was a similar finding to that of Hides et al. (1996), who found continued reductions in cross-sectional area of up to 10 weeks despite remission of symptoms and return to normal activities in a group of individuals with low back pain.

Similar to the effects on spinal extensor muscles, LTBR has also been found to result in differential rates of atrophy in the lower limb musculature (Belavý et al., 2009). Following 56-days of bed rest, Belavý et al. (2009) found reduced muscle volumes in all muscles within the lower limbs when measured with MRI, though rates of atrophy were greatest in the knee extensors. This finding led the authors to suggest that, in such populations, rehabilitation should particularly target these muscles. As shown here, FRED exercise incorporates this action, evidenced by the significantly greater RMS EMG amplitude of the vastus medialis during FRED (74.17 ± 39.02 %) compared to WALK (13.45 ± 4.96 %) and greater duration of activation (95.96 ± 8.97 % vs. 39.33 ± 29.15 %).

The extensor-flexor imbalance of the lumbopelvic musculature reported in both people following LTBR and in those with LBP highlights the need for a rehabilitative tool that is able to address the atrophy of the trunk extensors and counteract any increase in the size of the trunk flexors. Currently, most therapeutic interventions evaluated during LTBR and LBP studies lack functional relevance to activities of daily living. Also, to date, muscle cross-sectional area and/or thickness determined by ultrasound imaging (e.g. (Hides et al., 2008; Wallwork et al., 2009) or MRI (e.g. (Hides et al., 2008; Van, Hides, & Richardson, 2006) has been studied, but not the type of muscle activity (i.e. tonic or phasic) or whether the deep lumbopelvic muscles have regained their anticipatory action.

Greater Increase in LM than ES Activity during FRED Exercise Compared to Walking

Our findings also show the mean difference between FRED and WALK for LM was slightly greater for RMS EMG amplitude compared to ES (13.87 vs 12.27 % respectively), and was notably greater for LM compared to ES for the percentage of

movement cycle the muscle was active (45.10 % vs. 31.87 % respectively). While it was not possible to investigate TrA activity using surface EMG, Debuse et al. (2013) previously showed significantly greater TrA activity during FRED exercise than during a range of control conditions. Together with the findings for LM in this study, this may indicate that FRED exercise results in greater recruitment of deep then superficial lumbopelvic muscles in general. Deep lumbopelvic muscles are responsible for segmental spinal stability (Hodges, 1999; Hodges & Richardson, 1996; Panjabi, 1992a; Richardson et al., 2002). The fact that there is greater atrophy of the deep than superficial lumbopelvic muscles following LTBR and microgravity exposure (Belavý et al., 2011; Hides et al., 2007a; Sayson et al., 2013) has been suggested as the reason for the four fold incidence in disc prolapse in astronauts as compared to their peers (Johnston et al., 2010). Particularly, if a similar pattern of activation during FRED exercise was to be found for TrA, this could point to FRED exercise being more effective at addressing deep lumbopelvic muscle atrophy in people with LBP, following LTBR, and astronauts than conventional exercise approaches.

7.5 Limitations

In individuals with LBP, following LTBR and exposure to microgravity, the deep lumbopelvic muscles are atrophied and dysfunctional. *Transversus abdominis* and *LM* are the most widely studied and possibly the most important muscles in this context. However, as TrA is situated deep within the anterolateral abdominal wall, it cannot be studied with surface EMG. Its activity could, therefore, not be examined within this study. Despite previous reports of non-significant electrical crosstalk affecting the EMG signals measured here (Cholewicki & McGill, 1996; Cholewicki & Van Vliet 2002; Floyd & Silver, 1950; Vink, Van Der Velde, & Verbout, 1987), there is still the

possibility that the signals may have been influenced by some crosstalk. Research using indwelling EMG electrodes is warranted in order to fully validate the findings presented here, as well as to investigate TrA activity during FRED exercise.

7.6 Conclusion

The study reported in this chapter has demonstrated that FRED exercise leads to a more tonic activation of lumbopelvic muscles compared to walking. The fact that immediate exposure to this exercise modality results in a phasic-to-tonic shift in overall muscle recruitment when compared to overground walking and in a preferential activation of spinal extensors over the spinal flexors, as compared to walking suggest that the FRED could be an effective tool for use in rehabilitation of people following LTBR, in those with LBP and potentially in astronauts returning from long duration space flight. Further research is needed to evaluate the effectiveness of FRED exercise in restoring the extensor-flexor imbalance of the lumbopelvic musculature in these populations.

CHAPTER VIII
-
General Discussion

8.1 Findings and Significance

Current best practice for LBP management, following the initial triage and discounting of serious/systemic pathology (e.g. malignancy, inflammatory disorders, fractures or infections) and/or significant neurological deficits (e.g. cauda equina syndrome, sciatica or central stenosis) is guided by screening for psychosocial risk factors and addressing maladaptive beliefs and behaviours (O'Sullivan and Lin, 2014). Whilst this process is clearly necessary, it serves simply to inform primary care providers 'how' to best deliver care and does not serve to inform 'what' care is best delivered. This has resulted in a situation where a condition with a large heterogeneity of causal/contributory factors exists and an almost equally large number of management strategies are available.

With this in mind, the magnitudes of relative thickness change in the LM and TrA in a series of common clinical assessment techniques, during FRED exercise and treadmill walking were compared in Chapter IV. The key findings from this chapter were that all conditions resulted in active relative thickness change of LM and TrA with significant differences in LM between conditions that were not present in TrA. As expected, loaded contralateral arm lifts elicited significantly greater thickness changes than unloaded arm lifts. Treadmill walking elicited a significantly greater change in thickness than the loaded arm lift, but, importantly, any difference between the FRED exercise and walking were within the minimum detectable difference reported in Chapter III. As the loaded arm lift has previously been shown to elicit a contraction intensity of approximately 32 % of maximum voluntary contraction (Kiesel et al., 2007a), it is apparent that FRED exercise elicits a muscle thickness change of a similar intensity. Given that one of the fundamental principles of motor

control exercise is that the contraction intensity is low (Richardson & Jull, 1995), this finding goes some way to validating that FRED exercise recruits LM at levels conducive to training the deep lumbopelvic muscles.

As covered in Chapter II the spine is a hugely complex multi-articular system, whose typical function presupposes its stability by neutralising noxious forces and protecting adjacent segments. All whilst facilitating the transfer of forces between the upper and lower limbs during various combinations of movements executed in a range of dynamic and changeable body postures. As a result, this intrinsic system and extrinsic demand complexity may on occasion conspire to bring about the conditions that cause or contribute to the undesirable symptoms of low back pain. And in such cases, only 15% of these people with LBP symptoms have a pathoanatomical diagnosis identified (Airaksinen et al., 2006; Deyo, Rainville, & Kent, 1992; Koes, Van Tulder, & Thomas, 2006; Wand & O'Connell, 2008).

As such, it is proposed here that assessment of LM and TrA activity during FRED exercise can also function as a diagnostic/screening tool to be used alongside others such as the STarT Back Screening Tool (Hill et al., 2008) and the Short Form Orebro Musculoskeletal Pain Screening Questionnaire (Linton, Nicholas and MacDonald, 2011). Both of these are examples of validated tools designed for use in primary care settings that stratify patients into such psychological risk groups providing a basis for stratified care. However, to date, methods of stratifying patients using diagnostic imaging techniques to identify underlying pathology are particularly ineffective with clinical outcome typically not related to baseline characteristics of LM and TrA (Wong et al., 2013).

The FRED, in its simplest form, presents a functional, cyclical and low amplitude lower extremity perturbation to spinal stability. With a commonly held belief that this stability is primarily mediated by the local muscles of the trunk, that is, those with insertions and origins attached to the spinal column and therefore control single vertebral segments and are thus responsible for their stabilization and spinal instability is a direct causal factor in the development of LBP (Bergmark, 1989; Richardson and Jull, 1995). Hibbs *et al.* (2008), for example, warned that extensively recruiting the 'global' muscles would induce imbalances and these would consequently assume responsibility for stabilization of the spinal column. As a result, specific training exercises, such as motor control training, are often recommended, with these emphasising the key roles of the LM and TrA muscles.

However, a number of authors question the utility of such a dichotomous classification scheme, such as Lederman (2010) who wrote that such a classification system is anatomical but has no functional meaning and the separation of the trunk into local and global muscle systems is a reductionist fantasy. Leading to the conflicting idea that stability of the spinal column is achieved by synergistic cooperation of 'global' and 'local' muscles (Kavcic, Grenier, and McGill., 2004; Cholewicki and VanVliet, 2002; McGill et al., 2003). Cholewicki and VanVliet (2002) stated, for example, that the classification of muscles into local and global systems, as the way to discriminate between muscles responsible for inter-segmental stability and spine motion, is incorrect and instead, the trunk muscles should be seen as a functional unit with the individual contributions dependant on the motor task. A proposition supported by evidence from Kavcic, Grenier, and McGill (2004) when they quantified the importance of specific trunk muscles with regard to spine stability and found that no single muscle was dominant in ensuring the overall stability of the lumbar spine.

However, the conclusions of this work are limited to the contrived "stability" exercises tested, that included typically static postural challenges such as sitting on a gym ball, four-point kneeling and back bridges. The implications of these findings for prevention and rehabilitation of spinal stability in a population with LBP being that the clinical practice of isolated training of a specific muscle or muscle group to reduce the compressive costs of 'global' muscle contribution should be questioned. Thus, it appears justifiable to train motor patterns that involve the contribution of many important lumbar spine stabilisers

Regardless of whether or not one subscribes to the local/global classification scheme, when functioning within normal operating parameters, both systems contribute to the maintenance of spinal stability. However, when functioning outside of these normal operating parameters it is typically the local musculature that demonstrate dysfunction. It is observations such as this that have led to the increasing body of literature examining the effectiveness of exercise-based therapeutic intervention strategies targeting restoration/correction of the aforementioned dysfunctions, for example motor control training.

This model of exercise is consistent with many of the proposed 'needs' for a strategy aiming to restore proper capacity, control and function of the local stabilising elements of the lumbar spine, particularly the LM and TrA. However, despite this *a priori* face validity and a number of trials reporting on its effectiveness in LBP patient populations (Costa et al., 2009a; Ferreira et al., 2007; Hides, Jull, & Richardson, 2001; Hides, Richardson, & Jull, 1996; Hides & Stanton, 2014; Hides et al., 2012; Koumantakis, Watson, & Oldham, 2005; O'Sullivan, Twomey, & Allison, 1997; Streicher et al., 2014; Unsgaard-Tøndel et al., 2010) a recent Cochrane review (Saragiotto et al., 2016)

found moderate to high quality evidence that there is no clinically important difference between MCT and manual therapy for all follow-up periods and outcomes tested. A not entirely unsurprising finding when taking into account the aforementioned criticisms of the deep muscle concept.

With those criticisms in mind, in Chapter V, the magnitudes of relative thickness change in the anterolateral abdominal wall muscles and LM were examined. This was done in a series of challenges to postural stability that include some of the more common strategies used in motor control training such as sitting and standing on unstable surfaces, as well as during FRED exercise in both sitting and standing. The key findings from this study were that all stability challenges successfully induced non-volitional activation of both the LM and TrA muscles consistent with the requirements of motor control training. Additionally, it was found that the LM followed a pattern where all standing conditions elicited greater recruitment than sitting conditions, with no additional effect of surface instability. Contrastingly, the TrA only demonstrated an effect of surface instability during FRED exercise. Additionally, and of particular consequence, was the observation of increased preferential contraction of the TrA over the more superficial *internal* and *external oblique* muscles during FRED exercise. This suggests that a stabilising strategy beyond that of simple spinal stiffening (through co-contraction of superficial flexors and extensors) exists during FRED exercise. And when considered alongside the propositions of Cholewicki and Van Vliet (2002), Kavcic, Grenier and McGill (2004), and Lederman (2010), that the clinical practice of isolated training of a specific muscles or muscle groups should be questioned, these results suggest that FRED exercise provides more than just isolated training. Thus, FRED exercise clearly constitutes a potentially beneficial addition to existing rehabilitation approaches.

However, a potential limitation of this study was the use of ultrasound imaging. Whilst this approach was not necessarily new, with a number of previous studies relating relative change in muscle thicknesses as observed by USI to muscle activity recorded by EMG, only one study had previously used USI during a functional dynamic activity (Bunce, Moore, & Hough, 2002). Therefore, in order to ensure that any USI data collected (specifically relative thickness change of the LM and TrA) would be comparable to existing data available within the literature, a study was conducted determining the intra- and interday reliability of both static and dynamic movements.

The findings of this experiment (Chapter III) were that this ultrasound imaging approach was a reliable tool in the investigation of paraspinal and deep abdominal wall muscle function during dynamic activities and that a good to excellent level of repeatability can be achieved without the need for a rigid transducer holder. Prior to this study, authors typically reported on reliability during relatively static postures such as the abdominal drawing-in manoeuvre (Hides et al., 2007b), and upper and lower limb raises (Koppenhaver et al., 2009a). When reliability was assessed during active conditions these, too, included limited relevance to functional activities and included sitting on an exercise ball (Ainscough-Potts, Morrissey, & Critchley, 2006) or isometric flexion/extension/rotation tasks (Pietrek et al., 2000). Additionally, many investigations provided reliability and precision of measurement data for only single occasions (Kiesel et al., 2007a; Teyhen et al., 2005; Teyhen et al., 2008) or between only two separate visits (Kidd, Magee, & Richardson, 2002; Mannion et al., 2008; Rankin, Stokes, & Newham, 2006).

Finally, in Chapters VI and VII, the underlying mechanisms of action during FRED exercise were examined. The key findings of these studies were that, kinematically,

the lumbopelvic region was as stable during FRED exercise as walking in the sagittal and frontal planes, and characterised by reduced axial rotation in the transverse plane, as well as eliciting a more anteriorly tilted pelvis. Additionally, FRED exercise was also found to result in increased tonic activity of the lumbopelvic musculature, greater trunk extensor than flexor muscle activity, and greater knee extensor activity than overground walking. These features are all potentially beneficial to promoting the effective rehabilitation of the deep lumbopelvic muscles.

Taken together, these studies have shown, for the first time, at least in asymptomatic individuals, that a level of activity consistent with the proposals of the motor control framework can be achieved during a functional task (Functional Re-adaptive Exercise). At the same time, this task also addresses two of the major criticisms of motor control training, 1) that progression of the exercise programme does not fully incorporate functional movements in a standard manner and 2) that allocating the responsibility of spinal stability to a select sub-group of muscles may be an oversimplification.

8.2 Practical Implications

The traditional stance in most evidence-based treatment strategies of LBP was that as the source of pain cannot be determined for the large majority of patients presenting with low back pain. As such, these patients should be assigned to the classification of 'non-specific LBP' (Chou et al., 2007; Rossignol et al., 2007; Van Tulder et al., 2006) and be provided with generic treatment. However, in contemporary practice there has been some reconsideration of this position and an alternative proposition is to divide patients with non-specific LBP into treatment-based subgroups that inform the choice

of specific treatment for that individual (Kent and Keating, 2004; Kent, Kent and Keating, 2005).

What unifies these classification schemes, whether based on psychosocial characteristics (Vibe Fersum et al., 2013) or characteristic patterns of signs and symptoms (O'Sullivan, 2005), is an underlying hypothesis that the effect of treatment will be greater when patients receive the specific treatment that matches their subgroup. Proponents of treatment-based subgroups argue that this approach offers the possibility of much larger treatment effects than are typically observed after applying generic treatments to all patients with non-specific LBP. The argument here, being that any observed mean group treatment effects may be attenuated by the inclusion of subgroups of LBP patients for whom the treatment is not suitable and thus not effective (Delitto, 2005). However, to date, no stratification tool exists that achieves this. The findings presented in Chapters IV and V have provided a set of normative data to which individuals with LBP can now be stratified against using a reliable, non-invasive clinical assessment tool of ultrasound imaging.

Furthermore, the data presented within Chapter VII highlights the possibility that FRED exercise may also constitute a beneficial treatment modality for patients with LBP. Notable observations of atypical recruitment in individuals with LBP have previously been reported to include atrophy of the LM at multiple vertebral levels (Danneels et al., 2000; Hides et al., 1994), attenuated activity of the LM (Kiesel et al., 2007b; MacDonald, Moseley, & Hodges, 2010; Sihvonen et al., 1997) and TrA (Ferreira, Ferreira, & Hodges, 2004), delayed activity of the LM (MacDonald, Moseley, & Hodges, 2009) and TrA (Hodges & Richardson, 1996), and a shift from tonic to phasic activation of the TrA (Saunders, Coppieters, & Hodges, 2004). Whilst

motor control training has previously been shown to correct a number of these dysfunctions, in general it remains a comparatively ineffective treatment modality for LBP (Saragiotto et al., 2016). Therefore, of particular interest and practical importance is the finding that FRED exercise promoted tonic activity of the anterolateral abdominal wall and paraspinal muscles within this chapter. Consequently, incorporating FRED exercise into the rehabilitation strategy of individuals demonstrating atypical recruitment of lumbar spinal stabiliser muscles could have considerable impact on treatment efficacy.

8.3 Limitations & Future Directions

In Chapters III, IV and V where ultrasound imaging was the primary research tool, images were only captured at one specific instant in time through the foot movement cycle (or at heel strike in treadmill walking). Although this is common practice currently in ultra-sonographic assessment of muscle function it is possible that due to the more dynamic nature of this FRED exercise the direct comparisons made may not be entirely reflective of activity throughout the entire movement cycle. Chapter VII did attempt to address this issue, however, the use of surface electromyography rather than intramuscular electromyography limited the assessment to only those muscles that are superficial/accessible.

In each of the studies within this thesis a healthy asymptomatic cohort was used which limits the generalisability of the results to people with low back pain. At this stage of the investigations concerning FRED exercise, this was unavoidable. Prior to the studies contained within this thesis, only one study had previously investigated LM and TrA activity during FRED exercise (Debuse et al., 2013). Although the authors

did find the LM and TrA to be recruited automatically in a healthy population, it was felt that this knowledge alone was not sufficient to warrant direct investigation in a clinical population. However, as a result of concentrating on investigating muscle activity during a one-off exposure to FRED exercise in a non-symptomatic population, a more substantial body of evidence is now available to justify further investigations in clinical populations. This should now be used to form the basis of investigations of FRED exercise device in individuals with low back pain.

Immediate areas for address include:

1) investigating the true nature of differential activation of the paraspinal and lateral abdominal wall musculature during FRED exercise using the gold-standard technique of intramuscular electromyography;
2) examining the usefulness of FRED exercise in the assessment of dynamic muscle function in people with low back pain to identify whether or not distinctions between groups exist;
3) determining the effectiveness of incorporating FRED exercise in a rehabilitation programme for people with low back pain.

8.4 Contribution to Knowledge

This thesis aimed to expand our understanding of FRED exercise in the context of its potential relevance to LBP. In doing so, this thesis has provided:

1) the first dataset available demonstrating the reliability and precision of measurement using ultrasound imaging in the assessment of LM and TrA activity during a cyclical dynamic exercise on three separate occasions;

2) the first indirect estimation of LM and TrA contraction intensity during FRED exercise;
3) the first evidence of preferential activation of the TrA over IO and EO during FRED exercise;
4) the first demonstration of phasic-to-tonic shifts in muscle activity and the preferential activity of trunk and knee extensors over flexors during FRED exercise.

Taken together, these findings support the potential for use of FRED exercise in the rehabilitation of LM and TrA in populations where known dysfunctions of these muscles exist. Exercise using the Functional Re-adaptive Exercise Device results in a functionally dynamic non-volitional tonic co-activation of LM and TrA of a relatively low magnitude. This activity has the potential to augment Motor Control exercise therapies for individuals with maladaptive motor control strategies, such as those observed in low back pain, and following long-term bed-rest and exposure to microgravity environments. Therefore, the Functional Re-adaptive Exercise Device should be investigated further within these populations.

References

Adams, M., A, Burton, K., & Bogduk, N. (2013). *The biomechanics of back pain* (3 ed.). London: Elsevier Health Sciences.

Adams, M., A, & Hutton, W., C. (1983). The mechanical function of the lumbar apophyseal joints. *Spine, 8*(3), 327-330.

Adams, M. A., & Roughley, P. J. (2006). What is intervertebral disc degeneration, and what causes it? *Spine, 31*(18), 2151-2161.

Ainscough-Potts, A. M., Morrissey, M. C., & Critchley, D. (2006). The response of the transverse abdominis and internal oblique muscles to different postures. *Manual Therapy, 11*(1), 54-60.

Airaksinen, O., Brox, J. I., Cedraschi, C., Hildebrandt, J., Klaber-Moffett, J., Kovacs, F., . . . Ursin, H. (2006). Chapter 4 European guidelines for the management of chronic nonspecific low back pain. *European Spine Journal*, *15*, s192-s300.

Akbari, A., Khorashadizadeh, S., & Abdi, G. (2008). The effect of motor control exercise versus general exercise on lumbar local stabilizing muscles thickness: randomized controlled trial of patients with chronic low back pain. *Journal of Back and Musculoskeletal Rehabilitation, 21*(2), 105-112.

Allison, G. T., Morris, S. L., & Lay, B. (2008). Feedforward responses of transversus abdominis are directionally specific and act asymmetrically: implications for core stability theories. *Journal of Orthopaedic & Sports Physical Therapy, 38*(5), 228-237.

Amankwah, A. A., Banks, A., Derrick, I., Luff, M., Taylor, C., & Wales, P. (2015). *Economic Review, June 2015*. London: Office for National Statistics.

Anders, C., Wagner, H., Puta, C., Grassme, R., Petrovitch, A., & Scholle, H. C. (2007). Trunk muscle activation patterns during walking at different speeds. *Journal of Electromyography and Kinesiology, 17*(2), 245-252.

Andersson, G. B. J., Svensson, H., & Odén, A. (1983). The intensity of work recovery in low back pain. *Spine, 8*(8), 880-884.

Arslan, O. E. (2005). Anatomy of the abdominal wall. In M. A. Shiffman & S. Mirrafati (Eds.), *Aesthetic Surgery of the Abdominal Wall* (pp. 1-28). Berlin: Springer.

Askar, O. M. (1977). Surgical anatomy of the aponeurotic expansions of the anterior abdominal wall. *Annals of the Royal college of Surgeons of England, 59*(4), 313-321.

Aspden, R. M. (1992). Review of the functional anatomy of the spinal ligaments and the lumbar erector spinae muscles. *Clinical Anatomy*, *5*(5), 372-387.

Bakker, E. W. P., Verhagen, A. P., Lucas, C., Koning, H. J. C. M. F., de Haan, R. J., & Koes, B. W. (2007). Daily spinal mechanical loading as a risk factor for acute non-specific low back pain: a case–control study using the 24-Hour Schedule. *European Spine Journal, 16*(1), 107-113.

Balagué, F., Mannion, A. F., Pellisé, F., & Cedraschi, C. (2012). Non-specific low back pain. *The Lancet, 379*(9814), 482-491.

Barker, P. J., Guggenheimer, K. T., Grkovic, I., Briggs, C. A., Jones, D. C., Thomas, C. D. L., & Hodges, P. W. (2006). Effects of tensioning the lumbar fasciae on segmental stiffness during flexion and extension: Young Investigator Award winner. *Spine, 31*(4), 397-405.

Beazell, J. R., Grindstaff, T. L., Hart, J. M., Magrum, E. M., Cullaty, M., & Shen, F. H. (2011). Changes in lateral abdominal muscle thickness during an

abdominal drawing-in maneuver in individuals with and without low back pain. *Research in Sports Medicine, 19*(4), 271-282.

Behrsin, J. F., & Briggs, C. A. (1988). Ligaments of the lumbar spine: a review. *Surgical and Radiologic Anatomy, 10*(3), 211-219.

Belavý, D. L., Armbrecht, G., Richardson, C. A., Felsenberg, D., & Hides, J. A. (2011). Muscle atrophy and changes in spinal morphology: is the lumbar spine vulnerable after prolonged bed-rest? *Spine, 36*(2), 137-145.

Belavý, D. L., Hides, J. A., Wilson, S. J., Stanton, W., Dimeo, F. C., Rittweger, J., . . . Richardson, C. A. (2008). Resistive simulated weightbearing exercise with whole body vibration reduces lumbar spine deconditioning in bed-rest. *Spine, 33*(5), E121-131.

Belavý, D. L., Miokovic, T., Armbrecht, G., Richardson, C. A., Rittweger, J., & Felsenberg, D. (2009). Differential atrophy of the lower-limb musculature during prolonged bed-rest. *European Journal of Applied Physiology, 107*(4), 489-499.

Belavý, D. L., Richardson, C. A., Wilson, S. J., Felsenberg, D., & Rittweger, J. (2007). Tonic-to-phasic shift of lumbo-pelvic muscle activity during 8 weeks of bed rest and 6-months follow up. *Journal of Applied Physiology, 103*(1), 48-54.

Bergmark, A. (1989). Stability of the lumbar spine. A study in mechanical engineering. *Acta Orthopaedica Scandinavica. Supplementum, 230*, 1-54.

Bernhardt, M., & Bridwell, K. H. (1989). Segmental analysis of the sagittal plane alignment of the normal thoracic and lumbar spines and thoracolumbar junction. *Spine, 14*(7), 717-721.

Biedermann, H. J., Shanks, G. L., Forrest, W. J., & Inglis, J. (1991). Power Spectrum Analyses of Electromyographic Activity: Discriminators in the Differential

Assessment of Patients with Chronic Low-Back Pain. *Spine, 16*(10), 1179-1184.

Bogduk, N. (2005). *Clinical anatomy of the lumbar spine and sacrum*. London: Elsevier Churchill Livingstone.

Bogduk, N., & Macintosh, J. E. (1984). The applied anatomy of the thoracolumbar fascia. *Spine, 9*(2), 164-170.

Bogduk, N., Macintosh, J. E., & Pearcy, M. J. (1992). A universal model of the lumbar back muscles in the upright position. *Spine, 17*(8), 897-913.

Bogduk, N., Pearcy, M., & Hadfield, G. (1992). Anatomy and biomechanics of psoas major. *Clinical Biomechanics, 7*(2), 109-119.

Bousema, E. J., Verbunt, J. A., Seelen, H. A. M., Vlaeyen, J. W. S., & Knottnerus, J. A. (2007). Disuse and physical deconditioning in the first year after the onset of back pain. *Pain, 130*(3), 279-286.

Brown, S. H., & McGill, S. M. (2010). A comparison of ultrasound and electromyography measures of force and activation to examine the mechanics of abdominal wall contraction. *Clinical Biomechanics, 25*(2), 115-123.

Brumagne, S., Janssens, L., Janssens, E., & Goddyn, L. (2008). Altered postural control in anticipation of postural instability in persons with recurrent low back pain. *Gait and Posture, 28*(4), 657-662.

Buckey, J. C. (2006). *Space physiology*: Oxford University Press.

Bunce, S. M., Moore, A. P., & Hough, A. D. (2002). M-mode ultrasound: a reliable measure of transversus abdominis thickness? *Clinical Biomechanics, 17*(4), 315-317.

Burton, A. K. (2005). How to prevent low back pain. *Best Practice & Research Clinical Rheumatology, 19*(4), 541-555.

Carrino, J. A., Campbell Jr, P. D., Lin, D. C., Morrison, W. B., Schweitzer, M. E., Flanders, A. E., . . . Vaccaro, A. R. (2011). Effect of spinal segment variants on numbering vertebral levels at lumbar MR imaging. *Radiology, 259*(1), 196-202.

Cavanaugh, J. M., Ozaktay, A. C., Yamashita, H. T., & King, A. I. (1996). Lumbar facet pain: biomechanics, neuroanatomy and neurophysiology. *Journal of Biomechanics, 29*(9), 1117-1129.

Choi, B. C. K., & Pak, A. W. P. (2005). A catalog of biases in questionnaires. *Preventing Chronic Disease, 2*(1), 1-13.

Cholewicki, J., & McGill, S. M. (1996). Mechanical stability of the in vivo lumbar spine: implications for injury and chronic low back pain. *Clinical Biomechanics, 11*(1), 1-15.

Cholewicki, J., McGill, S. M., & Norman, R. W. (1991). Lumbar spine loads during the lifting of extremely heavy weights. *Medicine and Science in Sports and Exercise, 23*(10), 1179-1186.

Cholewicki, J., & Van Vliet , J. J. (2002). Relative contribution of trunk muscles to the stability of the lumbar spine during isometric exertions. *Clinical Biomechanics, 17*(2), 99-105.

Claus, A. P., Hides, J. A., Moseley, G. L., & Hodges, P. W. (2009). Different ways to balance the spine: subtle changes in sagittal spinal curves affect regional muscle activity. *Spine, 34*(6), E208-214.

Coldron, Y., Stokes, M., & Cook, K. (2003). Lumbar multifidus muscle size does not differ whether ultrasound imaging is performed in prone or side lying. *Manual Therapy, 8*(3), 161-165.

Comerford, M. J., & Mottram, S. L. (2001). Movement and stability dysfunction–contemporary developments. *Manual Therapy, 6*(1), 15-26.

Costa, L. O., Maher, C. G., Latimer, J., Hodges, P. W., Herbert, R. D., Refshauge, K. M., . . . Jennings, M. D. (2009a). Motor control exercise for chronic low back pain: a randomized placebo-controlled trial. *Physical Therapy, 89*(12), 1275-1286.

Costa, L. O., Maher, C. G., Latimer, J., & Smeets, R. J. (2009b). Reproducibility of rehabilitative ultrasound imaging for the measurement of abdominal muscle activity: a systematic review. *Physical Therapy, 89*(8), 756-769.

Coventry, M. B., Ghormley, R. K., & Kernohan, J. W. (1945). The intervertebral disc: its microscopic anatomy and pathology. *The Journal of Bone & Joint Surgery, 27*(3), 460-474.

Cresswell, A. G. (1993). Responses of intra-abdominal pressure and abdominal muscle activity during dynamic trunk loading in man. *European Journal of Applied Physiology and Occupational Physiology, 66*(4), 315-320.

Cresswell, A. G., Grundström, H., & Thorstensson, A. (1992). Observations on intra-abdominal pressure and patterns of abdominal intra-muscular activity in man. *Acta Physiologica Scandinavica, 144*(4), 409-418.

Cresswell, A. G., Oddsson, L., & Thorstensson, A. (1994). The influence of sudden perturbations on trunk muscle activity and intra-abdominal pressure while standing. *Experimental Brain Research, 98*(2), 336-341.

Crisco, J. J., Panjabi, M. M., Yamamoto, I., & Oxland, T. R. (1992). Euler Stability of the Human Ligamentous Lumbar Spine Part II: Experiment. *Clinical Biomechanics, 7*(1), 27-32.

Critchley, D. J., & Coutts, F. J. (2002). Abdominal muscle function in chronic low back pain patients: measurement with real-time ultrasound scanning. *Physiotherapy, 88*(6), 322-332.

Dagenais, S., Caro, J., & Haldeman, S. (2008). A systematic review of low back pain cost of illness studies in the United States and internationally. *The Spine Journal, 8*(1), 8-20.

Dagenais, S., Tricco, A. C., & Haldeman, S. (2010). Synthesis of recommendations for the assessment and management of low back pain from recent clinical practice guidelines. *The Spine Journal, 10*, 514-529.

Daggfeldt, K., & Thorstensson, A. (2003). The mechanics of back-extensor torque production about the lumbar spine. *Journal of Biomechanics, 36*(6), 815-825.

Danneels, L. A., Vanderstraeten, G. G., Cambier, D. C., Witvrouw, E. E., & De Cuyper, H. J. (2000). CT imaging of trunk muscles in chronic low back pain patients and healthy control subjects. *European Spine Journal, 9*(4), 266-272.

Debuse, D., Birch, O., St Clair Gibson, A., & Caplan, N. (2013). Low impact weight-bearing exercise in an upright posture increases the activation of two key local muscles of the pumbo-pelvic region. *Physiotherapy Theory and Practice, 29*, 51-60.

Delitto, A., Erhard, R. E., & Bowling, R. W. (1995). A treatment-based classification approach to low back syndrome: identifying and staging patients for conservative treatment. *Physical Therapy, 75*(6), 470-485.

Delp, S. L., Suryanarayanan, S., Murray, W. M., Uhlir, J., & Triolo, R. J. (2001). Architecture of the rectus abdominis, quadratus lumborum, and erector spinae. *Journal of Biomechanics, 34*(3), 371-375.

Demoulin, C., Distree, V., Tomasella, M., Crielaard, J.-M., & Vanderthommen, M. (2007). *Lumbar functional instability: a critical appraisal of the literature.* Paper presented at the Annales de readaptation et de medecine physique.

Deyo, R. A., Rainville, J., & Kent, D. L. (1992). What can the history and physical examination tell us about low back pain? *JAMA: The Journal of the American Medical Association, 268*(6), 760-765.

Dixon, A. S. J. (1973). Progress and problems in back pain research. *Rheumatology, 12*(4), 165-175.

Dofferhof, A., & Vink, P. (1985). The stabilising function of the mm. iliocostales and the mm. multifidi during walking. *Journal of Anatomy, 140*(Pt 2), 329-336.

Donatelli, R. A. (2007). The Anatomy and Pathophysiology of the CORE. *Sports-specific Rehabilitation*, 135.

Drake, J. D. M., Fischer, S. L., Brown, S. H. M., & Callaghan, J. P. (2006). Do Exercise Balls Provide a Training Advantage for Trunk Extensor Exercises? A Biomechanical Evaluation. *Journal of Manipulative and Physiological Therapeutics, 29*(5), 354-362.

Dreyer, N. A., Tunis, S. R., Berger, M., Ollendorf, D., Mattox, P., & Gliklich, R. (2010). Why observational studies should be among the tools used in comparative effectiveness research. *Health Affairs, 29*(10), 1818-1825.

Dumas, G. A., Poulin, M. J., Roy, B., Gagnon, M., & Jovanovic, M. (1991). Orientation and moment arms of some trunk muscles. *Spine, 16*(3), 293-303.

Ebraheim, N. A., Rollins Jr, J. R., Xu, R., & Yesting, R. A. (1996). Projection of the lumbar pedicle and its morphometric analysis. *Spine, 21*(11), 1296-1300.

Elfering, A., & Mannion, A. F. (2008). Epidemiology and risk factors of spinal disorders *Spinal Disorders* (pp. 153-173): Springer.

Eswaran, S. K., Gupta, A., Adams, M. F., & Keaveny, T. M. (2006). Cortical and trabecular load sharing in the human vertebral body. *Journal of Bone and Mineral Research, 21*(2), 307-314.

Evetts, S. N., Caplan, N., Debuse, D., Lambrecht, G., Damann, V., Petersen, N., & Hides, J. (2014). Post Space Mission Lumbo-Pelvic Neuromuscular Reconditioning: A European Perspective. *Aviation, Space, and Environmental Medicine, 85*(7), 764-765.

Faas, A., Chavannes, A. W., Van Eijk, J. T. M., & Gubbels, J. W. (1993). A randomized, placebo-controlled trial of exercise therapy in patients with acute low back pain. *Spine, 18*(11), 1388-1395.

Ferguson, S. J., & Steffen, T. (2003). Biomechanics of the aging spine. *European Spine Journal, 12*(2), S97-S103.

Ferreira, M. L., Ferreira, P. H., Latimer, J., Herbert, R. D., Hodges, P. W., Jennings, M. D., . . . Refshauge, K. M. (2007). Comparison of general exercise, motor control exercise and spinal manipulative therapy for chronic low back pain: a randomized trial. *Pain, 131*(1), 31-37.

Ferreira, P. H., Ferreira, M. L., & Hodges, P. W. (2004). Changes in recruitment of the abdominal muscles in people with low back pain: ultrasound measurement of muscle activity. *Spine, 29*(22), 2560-2566.

Ferreira, P. H., Ferreira, M. L., Maher, C. G., Herbert, R. D., & Refshauge, K. (2006). Specific stabilisation exercise for spinal and pelvic pain: a systematic review. *Australian Journal of Physiotherapy, 52*(2), 79-88.

Floyd, W. F., & Silver, P. H. S. (1950). Electromyographic study of patterns of activity of the anterior abdominal wall muscles in man. *Journal of Anatomy, 84*(Pt 2), 132-145.

Freburger, J. K., Holmes, G. M., Agans, R. P., Jackman, A. M., Darter, J. D., Wallace, A. S., . . . Carey, T. S. (2009). The rising prevalence of chronic low back pain. *Archives of Internal Medicine, 169*(3), 251-258.

Freeman, M. D., Woodham, M. A., & Woodham, A. W. (2010). The role of the lumbar multifidus in chronic low back pain: a review. *PM&R, 2*(2), 142-146.

Fritz, J. M., Erhard, R. E., & Hagen, B. F. (1998). Segmental instability of the lumbar spine. *Physical Therapy, 78*(8), 889-896.

Fujiwara, A., Tamai, K., An, H. S., Shimizu, K., Yoshida, H., & Saotome, K. (2000). The interspinous ligament of the lumbar spine: magnetic resonance images and their clinical significance. *Spine, 25*(3), 358-363.

Fukunaga, T., Kubo, K., Kawakami, Y., Fukashiro, S., Kanehisa, H., & Maganaris, C. N. (2001). In vivo behaviour of human muscle tendon during walking. *Proceedings of the Royal Society of London. Series B: Biological Sciences, 268*(1464), 229-233.

Gardner-Morse, M. G., & Stokes, I. A. F. (1998). The effects of abdominal muscle coactivation on lumbar spine stability. *Spine, 23*(1), 86-91.

Gardner-Morse, M. G., Stokes, I. A. F., & Laible, J. P. (1995). Role of muscles in lumbar spine stability in maximum extension efforts. *Journal of Orthopaedic Research, 13*(5), 802-808.

George, S. Z., Fritz, J. M., Bialosky, J. E., & Donald, D. A. (2003). The effect of a fear-avoidance–based physical therapy intervention for patients with acute low back pain: results of a randomized clinical trial. *Spine, 28*(23), 2551-2560.

Gilchrist, R. V., Frey, M. E., & Nadler, S. F. (2003). Muscular control of the lumbar spine. *Pain Physician, 6*(3), 361-368.

Girden, E. R. (1992). *ANOVA: repeated measures*. Newbury Park, CA: Sage.

Goldman, J. M., Lehr, R. P., Millar, A. B., & Silver, J. R. (1987). An electromyographic study of the abdominal muscles during postural and

respiratory manoeuvres. *Journal of Neurology, Neurosurgery & Psychiatry, 50*(7), 866-869.

Gorbet, N., Selkow, N. M., Hart, J. M., & Saliba, S. (2010). No difference in transverse abdominis activation ratio between healthy and asymptomatic low back pain patients during therapeutic exercise. *Rehabilitation Research and Practice, 2010*(Article ID 459738), 1-6.

Granata, K. P., & Marras, W. S. (2000). Cost–benefit of muscle cocontraction in protecting against spinal instability. *Spine, 25*(11), 1398-1404.

Gregory, D. E., Dunk, N. M., & Callaghan, J. P. (2006). Stability ball versus office chair: comparison of muscle activation and lumbar spine posture during prolonged sitting. *Human Factors, 48*(1), 142-153.

Grenier, S. G., Vera-Garcia, F. J., & McGill, S. M. (2000). Abdominal response during curl-ups on both stable and labile surfaces. *Physical Therapy, 86*, 564-569.

Guzik, D. C., Keller, T. S., Szpalski, M., Park, J. H., & Spengler, D. M. (1996). A biomechanical model of the lumbar spine during upright isometric flexion, extension, and lateral bending. *Spine, 21*(4), 427-433.

Hagen, K. B., Hilde, G., Jamtvedt, G., & Winnem, M. (2004). Bed rest for acute low-back pain and sciatica. *The Cochrane Library*(4), 1-41.

Hamlyn, N., Behm, D. G., & Young, W. B. (2007). Trunk muscle activation during dynamic weight-training exercises and isometric instability activities. *The Journal of Strength & Conditioning Research, 21*(4), 1108-1112.

Hauggaard, A., & Persson, A. L. (2007). Specific spinal stabilisation exercises in patients with low back pain–a systematic review. *Physical Therapy Reviews, 12*(3), 233-248.

Hebert, J. J., Koppenhaver, S. L., Magel, J. S., & Fritz, J. M. (2010). The relationship of transversus abdominis and lumbar multifidus activation and prognostic factors for clinical success with a stabilization exercise program: a cross-sectional study. *Archives of Physical Medicine and Rehabilitation, 91*(1), 78-85.

Hebert, J. J., Koppenhaver, S. L., Parent, E. C., & Fritz, J. M. (2009). A Systematic Review of the Reliability of Rehabilitative Ultrasound Imaging for the Quantitative Assessment of the Abdominal and Lumbar Trunk Muscles. *Spine, 34*(23), E848-E856.

Hendrick, P., Te Wake, A. M., Tikkisetty, A. S., Wulff, L., Yap, C., & Milosavljevic, S. (2010). The effectiveness of walking as an intervention for low back pain: a systematic review. *European Spine Journal, 19*(10), 1613-1620.

Henry, S. M., Hitt, J. R., Jones, S. L., & Bunn, J. Y. (2006). Decreased limits of stability in response to postural perturbations in subjects with low back pain. *Clinical Biomechanics, 21*(9), 881-892.

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology, 10*(5), 361-374.

Hestbaek, L., Leboeuf-Yde, C., Engberg, M., Lauritzen, T., Bruun, N. H., & Manniche, C. (2003). The course of low back pain in a general population. Results from a 5-year prospective study. *Journal of Manipulative and Physiological Therapeutics, 26*(4), 213-219.

Hestbaek, L., Leboeuf-Yde, C., & Manniche, C. (2003). Low back pain: what is the long-term course? A review of studies of general patient populations. *European Spine Journal, 12*(2), 149-165.

Hides, J., Stanton, W., Wilson, S., Freke, M., McMahon, S., & Sims, K. (2010). Retraining motor control of abdominal muscles among elite cricketers with low back pain. *Scandinavian Journal of Medicine & Science in Sports, 20*(6), 834-842.

Hides, J. A., Belavy, D. L., Stanton, W., Wilson, S. J., Rittweger, J., Felsenberg, D., & Richardson, C. A. (2007a). Magnetic resonance imaging assessment of trunk muscles during prolonged bed rest. *Spine, 32*(15), 1687-1692.

Hides, J. A., Cooper, D. H., & Stokes, M. J. (1992). Diagnostic Ultrasound Imaging for Measurement of the Lumbar Multifidus Muscle in Normal Young Adults. *Physiotherapy Theory and Practice, 8*(1), 19-26.

Hides, J. A., Jull, A., & Richardson, C. A. (2001). Long-term effects of specific stabilizing exercises for first-episode low back pain. *Spine, 26*(11), e243-e248.

Hides, J. A., Lambrecht, G., Richardson, C. A., Stanton, W. R., Armbrecht, G., Pruett, C., . . . Belavý, D. L. (2011a). The effects of rehabilitation on the muscles of the trunk following prolonged bed rest. *European Spine Journal, 20*(5), 808-818.

Hides, J. A., Miokovic, T., Belavy, D. L., Stanton, W. R., & Richardson, C. A. (2007b). Ultrasound imaging assessment of abdominal muscle function during drawing-in of the abdominal wall: an intrarater reliability study. *Journal of Orthopaedic & Sports Physical Therapy, 37*(8), 480-486.

Hides, J. A., Richardson, C. A., & Jull, G. A. (1996). Multifidus muscle recovery is not automatic after resolution of acute, first-episode low back pain. *Spine, 21*(23), 2763-2769.

Hides, J. A., Scott, Q., Jull, G., & Richardson, C. (2000). A clinical palpation test to check the activation of the deep stabilizing muscles of the lumbar spine. *International SportMed Journal, 1*(4), 1-4.

Hides, J. A., Stanton, W., Dilani Mendis, M., & Sexton, M. (2011b). The relationship of transversus abdominis and lumbar multifidus clinical muscle tests in patients with chronic low back pain. *Manual Therapy, 16*(6), 573-577.

Hides, J. A., & Stanton, W. R. (2014). Can motor control training lower the risk of injury for professional football players. *Medicine and Science in Sports and Exercise, 46*(4), 762-768.

Hides, J. A., Stanton, W. R., McMahon, S., Sims, K., & Richardson, C. A. (2008). Effect of stabilization training on multifidus muscle cross-sectional area among young elite cricketers with low back pain. *Journal of Orthopaedic & Sports Physical Therapy, 38*(3), 101-108.

Hides, J. A., Stanton, W. R., Mendis, M. D., Gildea, J., & Sexton, M. J. (2012). Effect of motor control training on muscle size and football games missed from injury. *Medicine and Science in Sports and Exercise, 44*(6), 1141-1149.

Hides, J. A., Stokes, M. J., Saide, M., Jull, G. A., & Cooper, D. H. (1994). Evidence of lumbar multifidus muscle wasting ipsilateral to symptoms in patients with acute/subacute low back pain. *Spine, 19*(2), 165-172.

Hodges, P., & Cholewicki, J. (2007). Functional control of the spine. In A. Vleeming, V. Mooney, & R. Stoeckart (Eds.), *Movement, Stability and Lumbopelvic Pain*. Endinburgh: Churchill Livingstone.

Hodges, P., Ferreira, P. H., & Ferreira, M. L. (2009). Lumbar spine: treatment of instability and disorders of movement control. In D. Magee, J. Zachazewski, & W. Quillen (Eds.), *Pathology and Intervention in Musculoskeletal Rehabilitation* (1st Edition ed., pp. 389-425). United States: Elsevier

Hodges, P., Holm, A. K., Hansson, T., & Holm, S. (2006). Rapid atrophy of the lumbar multifidus follows experimental disc or nerve root injury. *Spine, 31*(25), 2926-2933.

Hodges, P., Richardson, C., & Jull, G. (1996). Evaluation of the relationship between laboratory and clinical tests of transversus abdominis function. *Physiotherapy Research International, 1*(1), 30-40.

Hodges, P. W. (1999). Is there a role for transversus abdominis in lumbo-pelvic stability? *Manual Therapy, 4*(2), 74-86.

Hodges, P. W. (2003). Core stability exercise in chronic low back pain. *Orthopedic Clinics of North America, 34*(2), 245-254.

Hodges, P. W. (2011). Pain and motor control: From the laboratory to rehabilitation. *Journal of Electromyography and Kinesiology, 21*(2), 220-228.

Hodges, P. W., Cholewicki, J., & Van Dieën, J. H. (2013). *Spinal Control: The Rehabilitation of Back Pain: State of the Art and Science*: Elsevier Health Sciences.

Hodges, P. W., Cresswell, A. G., Daggfeldt, K., & Thorstensson, A. (2001). In vivo measurement of the effect of intra-abdominal pressure on the human spine. *Journal of Biomechanics, 34*(3), 347-353.

Hodges, P. W., McGill, S. M., & Hides, J. (2013). Motor control of the spine and changes in pain: debate about the extrapolation from research observations of motor control strategies to effective treatments for back pain. In P. W. Hodges, J. Cholewicki, & J. H. Van Dieën (Eds.), *Spinal Control: The Rehabilitation of Back Pain*: Elsevier Health Sciences.

Hodges, P. W., Pengel, L. H., Herbert, R. D., & Gandevia, S. C. (2003). Measurement of muscle contraction with ultrasound imaging. *Muscle and Nerve, 27*(6), 682-692.

Hodges, P. W., & Richardson, C. A. (1996). Inefficient muscular stabilization of the lumbar spine associated with low back pain. A motor control evaluation of transversus abdominis. *Spine, 21*(22), 2640-2650.

Hodges, P. W., & Richardson, C. A. (1997). Feedforward contraction of transversus abdominis is not influenced by the direction of arm movement. *Experimental Brain Research, 114*(2), 362-370.

Hodges, P. W., & Richardson, C. A. (1999). Altered trunk muscle recruitment in people with low back pain with upper limb movement at different speeds. *Archives of Physical Medicine and Rehabilitation, 80*(9), 1005-1012.

Hodges, P. W., van Dillen, L., McGill, S. M., Brumagne, S., Hides, J., & Moseley, G. L. (2013). Integrated clinical approach to motor control interventions in low back and pelvic pain. In P. W. Hodges, J. Cholewicki, & J. H. Van Dieën (Eds.), *Spinal Control: The Rehabilitation of Back Pain*: Elsevier Health Sciences.

Horn, S. D., & Gassaway, J. (2010). Practice based evidence: incorporating clinical heterogeneity and patient-reported outcomes for comparative effectiveness research. *Medical Care, 48*(6), S17-S22.

Howard, D., Best, W., & Nickels, L. (2015). Optimising the design of intervention studies: Critiques and ways forward. *Aphasiology, 29*(5), 526-562.

Hoy, D., Bain, C., Williams, G., March, L., Brooks, P., Blyth, F., . . . Buchbinder, R. (2012a). A systematic review of the global prevalence of low back pain. *Arthritis & Rheumatism, 64*(6), 2028-2037.

Hoy, D., Brooks, P., Woolf, A., Blyth, F., March, L., Bain, C., . . . Buchbinder, R. (2012b). Assessing risk of bias in prevalence studies: modification of an existing tool and evidence of interrater agreement. *Journal of Clinical Epidemiology, 65*(9), 934-939.

Hu, H., Meijer, O. G., Hodges, P. W., Bruijn, S. M., Strijers, R. L., Nanayakkara, P. W., . . . van Dieën, J. H. (2012). Understanding the Active Straight Leg Raise (ASLR): An electromyographic study in healthy subjects. *Manual Therapy, 17*(6), 531-537.

Humzah, M., & Soames, R. (1988). Human intervertebral disc: structure and function. *The Anatomical Record, 220*(4), 337-356.

Itz, C., Geurts, J., Kleef, M. v., & Nelemans, P. (2013). Clinical course of non-specific low back pain: A systematic review of prospective cohort studies set in primary care. *European Journal of Pain, 17*(1), 5-15.

Jensen, G. M. (1980). Biomechanics of the lumbar intervertebral disk: a review. *Physical Therapy, 60*(6), 765-773.

Joffe, D., Watkins, M., Steiner, L., & Pfeifer, B. A. (2002). Treadmill ambulation with partial body weight support for the treatment of low back and leg pain. *Journal of Orthopaedic & Sports Physical Therapy, 32*(5), 202-213.

John, E., & Beith, I. (2007). Can activity within the external abdominal oblique be measured using real-time ultrasound imaging? *Clinical Biomechanics, 22*(9), 972-979.

Johnston, S. L., Campbell, M. R., Scheuring, R., & Feiveson, A. H. (2010). Risk of herniated nucleus pulposus among US astronauts. *Aviation, Space, and Environmental Medicine, 81*(6), 566-574.

Jonsson, B. (1970). The functions of individual muscles in the lumbar part of the spinae muscle. *Electromyography, 10*(1), 5-21.

Kader, D. F., Wardlaw, D., & Smith, F. W. (2000). Correlation between the MRI changes in the lumbar multifidus muscles and leg pain. *Clinical Radiology, 55*(2), 145-149.

Kavcic, N., Grenier, S., & McGill, S. M. (2004). Determining the stabilizing role of individual torso muscles during rehabilitation exercises. *Spine, 29*(11), 1254-1265.

Kay, A. G. (2000). An extensive literature review of the lumbar multifidus: anatomy. *Journal of Manual & Manipulative Therapy, 8*(3), 102-114.

Kidd, A. W., Magee, S., & Richardson, C. A. (2002). Reliability of real-time ultrasound for the assessment of transversus abdominis function. *Journal of Gravitational Physiology, 9*(1), 131-132.

Kiesel, K. B., Uhl, T., Underwood, F. B., & Nitz, A. J. (2008). Rehabilitative ultrasound measurement of select trunk muscle activation during induced pain. *Manual Therapy, 13*(2), 132-138.

Kiesel, K. B., Uhl, T., Underwood, F. B., Rodd, D. W., & Nitz, A. J. (2007a). Measurement of lumbar multifidus muscle contraction with rehabilitative ultrasound imaging. *Manual Therapy, 12*(2), 161-166.

Kiesel, K. B., Underwood, F. B., Mattacola, C. G., Nitz, A. J., & Malone, T. R. (2007b). A comparison of select trunk muscle thickness change between subjects with low back pain classified in the treatment-based classification system and asymptomatic controls. *Journal of Orthopaedic & Sports Physical Therapy, 37*(10), 596-607.

Kim, C., Gottschalk, L., Eng, C., Ward, S., & Lieber, R. (2007). The multifidus muscle is the strongest stabilizer of the lumbar spine. *The Spine Journal, 7*, 76S.

Kim, M., Kim, S., Kim, S., Park, J., & Han, D. (2012). Muscle activations of the paraspinal muscles in different types of shoe during walking. *Journal of Physical Therapy Science, 24*(9), 905-907.

Kingma, I., & van Dieën, J. H. (2009). Static and dynamic postural loadings during computer work in females: Sitting on an office chair versus sitting on an exercise ball. *Applied Ergonomics, 40*(2), 199-205.

Kirkaldy-Willis, W., & Farfan, H. (1982). Instability of the lumbar spine. *Clinical Orthopaedics and Related Research, 165*, 110-123.

Kjaer, P., Bendix, T., Sorensen, J. S., Korsholm, L., & Leboeuf-Yde, C. (2007). Are MRI-defined fat infiltrations in the multifidus muscles associated with low back pain? *BMC Medicine, 5*(1), 1-10.

Koes, B. W., & Van Den Hoogen, H. M. M. (1994). Efficacy of bed rest and orthoses of low-back pain: A review of randomized clinical trials. *European Journal of Physical Medicine & Rehabilitation, 4*(3), 86-93.

Koes, B. W., van Tulder, M., Lin, C.-W. C., Macedo, L. G., McAuley, J., & Maher, C. (2010). An updated overview of clinical guidelines for the management of non-specific low back pain in primary care. *European Spine Journal, 19*(12), 2075-2094.

Koes, B. W., van Tulder, M. W., Ostelo, R., Burton, A. K., & Waddell, G. (2001). Clinical guidelines for the management of low back pain in primary care: an international comparison. *Spine, 26*(22), 2504-2513.

Koes, B. W., Van Tulder, M. W., & Thomas, S. (2006). Diagnosis and treatment of low back pain. *British Medical Journal, 332*(7555), 1430-1434.

Koppenhaver, S. L., Fritz, J. M., Hebert, J. J., Kawchuk, G. N., Childs, J. D., Parent, E. C., . . . Teyhen, D. S. (2011). Association between changes in abdominal and lumbar multifidus muscle thickness and clinical improvement after spinal manipulation. *Journal of Orthopaedic & Sports Physical Therapy, 41*(6), 389-399.

Koppenhaver, S. L., Hebert, J. J., Fritz, J. M., Parent, E. C., Teyhen, D. S., & Magel, J. S. (2009a). Reliability of rehabilitative ultrasound imaging of the transversus abdominis and lumbar multifidus muscles. *Archives of Physical Medicine and Rehabilitation, 90*(1), 87-94.

Koppenhaver, S. L., Parent, E. C., Teyhen, D. S., Hebert, J. J., & Fritz, J. M. (2009b). The effect of averaging multiple trials on measurement error during ultrasound imaging of transversus abdominis and lumbar multifidus muscles in individuals with low back pain. *Journal of Orthopaedic & Sports Physical Therapy, 39*(8), 604-611.

Kordi, R., Rostami, M., Noormohammadpour, P., & Mansournia, M. A. (2011). The effect of food consumption on the thickness of abdominal muscles, employing ultrasound measurements. *European Spine Journal, 20*(8), 1312-1317.

Korfmacher, G., Debuse, D., & Pinotti, M. (2006). STIMULON--specific training of the musculoskeletal system as nature intended. *Journal of Biomechanics, 39*(S1), S488-489.

Koumantakis, G. A., Watson, P. J., & Oldham, J. A. (2005). Supplementation of general endurance exercise with stabilisation training versus general exercise only: physiological and functional outcomes of a randomised controlled trial of patients with recurrent low back pain. *Clinical Biomechanics, 20*(5), 474-482.

Kriese, M., Clijsen, R., Taeymans, J., & Cabri, J. (2010). [Segmental stabilization in low back pain: a systematic review]. *Sportverletzung Sportschaden: Organ der Gesellschaft fur Orthopadisch-Traumatologische Sportmedizin, 24*(1), 17-25.

Larivière, C., Gagnon, D., De Oliveira Jr, E., Henry, S. M., Mecheri, H., & Dumas, J.-P. (2013). Ultrasound Measures of the Lumbar Multifidus: Effect of Task and Transducer Position on Reliability. *PM&R, 5*(8), 678-687.

Leboeuf-Yde, C., & Lauritsen, J. M. (1995). The Prevalence of Low Back Pain in the Literature A Structured Review of 26 Nordic Studies From 1954 to 1993. *Spine, 20*(19), 2112-2118.

Lee, J.-H., Ooi, Y., & Nakamura, K. (1995). Measurement of muscle strength of the trunk and the lower extremities in subjects with history of low back pain. *Spine, 20*(18), 1994-1996.

Levangie, P. K., & Norkin, C. C. (Eds.). (2011). *Joint structure and function: a comprehensive analysis* (5 ed.). Philadelphia: FA Davis.

Lidgren, L. (2003). The bone and joint decade 2000-2010. *Bulletin of the World Health Organization, 81*(9), 629-629.

Lin, C.-W. C., McAuley, J. H., Macedo, L., Barnett, D. C., Smeets, R. J., & Verbunt, J. A. (2011). Relationship between physical activity and disability in low back pain: a systematic review and meta-analysis. *Pain, 152*(3), 607-613.

Lotz, J., Fields, A., & Liebenberg, E. (2013). The role of the vertebral end plate in low back pain. *Global Spine Journal, 3*(3), 153-164.

Lund, J. P., Donga, R., Widmer, C. G., & Stohler, C. S. (1991). The pain-adaptation model: a discussion of the relationship between chronic musculoskeletal pain and motor activity. *Canadian Journal of Physiology and Pharmacology, 69*(5), 683-694.

Luoto, S., Heliövaara, M., Hurri, H., & Alaranta, H. (1995). Static back endurance and the risk of low-back pain. *Clinical Biomechanics, 10*(6), 323-324.

MacDonald, D., Moseley, G. L., & Hodges, P. W. (2009). Why do some patients keep hurting their back? Evidence of ongoing back muscle dysfunction during remission from recurrent back pain. *Pain, 142*(3), 183-138.

MacDonald, D., Moseley, G. L., & Hodges, P. W. (2010). People with recurrent low back pain respond differently to trunk loading despite remission from symptoms. *Spine, 35*(7), 818-824.

MacDonald, D. A., Moseley, G. L., & Hodges, P. W. (2006). The lumbar multifidus: does the evidence support clinical beliefs? *Manual Therapy, 11*(4), 254-263.

Macedo, L. G., Maher, C. G., Latimer, J., & McAuley, J. H. (2009). Motor control exercise for persistent, nonspecific low back pain: a systematic review. *Physical Therapy, 89*(1), 9-25.

Macintosh, J. E., & Bogduk, N. (1986). The biomechanics of the lumbar multifidus. *Clinical Biomechanics, 1*(4), 205-213.

Macintosh, J. E., & Bogduk, N. (1987). 1987 Volvo Award in Basic Science: The Morphology of the Lumbar Erector Spinae. *Spine, 12*(7), 658-668.

Macintosh, J. E., Valencia, F., Bogduk, N., & Munro, R. R. (1986). The morphology of the human lumbar multifidus. *Clinical Biomechanics, 1*(4), 196-204.

Malmivaara, A., Häkkinen, U., Aro, T., Heinrichs, M.-L., Koskenniemi, L., Kuosma, E., . . . Vaaranen, V. (1995). The treatment of acute low back pain—bed rest, exercises, or ordinary activity? *New England Journal of Medicine, 332*(6), 351-355.

Mangum, L. C., Sutherlin, M. A., Saliba, S. A., & Hart, J. M. (2015). Reliability of Ultrasound Imaging Measures of Transverse Abdominis and Lumbar Multifidus in Various Positions. *PM&R, 8*(4), 340-347.

Maniadakis, N., & Gray, A. (2000). The economic burden of back pain in the UK. *Pain, 84*(1), 95-103.

Manniche, C., Bentzen, L., Hesselsoe, G., Christensen, I., & Lundberg, E. (1988). Clinical trial of intensive muscle training for chronic low back pain. *The Lancet, 332*(8626), 1473-1476.

Manniche, C., Lundberg, E., Christensen, I., Bentzen, L., & Hesselsøe, G. (1991). Intensive dynamic back exercises for chronic low back pain: a clinical trial. *Pain, 47*(1), 53-63.

Mannion, A. F., Pulkovski, N., Gubler, D., Gorelick, M., O'Riordan, D., Loupas, T., . . . Sprott, H. (2008). Muscle thickness changes during abdominal hollowing: an assessment of between-day measurement error in controls and patients with chronic low back pain. *European Spine Journal, 17*(4), 494-501.

Manshadi, F. D., Parnianpour, M., Sarrafzadeh, J., reza Azghani, M., & Kazemnejad, A. (2011). Abdominal hollowing and lateral abdominal wall muscles' activity in both healthy men & women: An ultrasonic assessment in supine and standing positions. *Journal of Bodywork and Movement Therapies, 15*(1), 108-113.

Marshall, P. W., & Murphy, B. A. (2005). Core stability exercises on and off a Swiss ball. *Archives of Physical Medicine and Rehabilitation, 86*(2), 242-249.

Masharawi, Y., Salame, K., Mirovsky, Y., Peleg, S., Dar, G., Steinberg, N., & Hershkovitz, I. (2008). Vertebral body shape variation in the thoracic and lumbar spine: characterization of its asymmetry and wedging. *Clinical Anatomy, 21*(1), 46-54.

May, S., & Johnson, R. (2008). Stabilisation exercises for low back pain: a systematic review. *Physiotherapy, 94*(3), 179-189.

Mayer, T., Kondraske, G., Mooney, V., Carmichael, T. W., & Butsch, R. (1989). Lumbar Myoelectric Spectral Analysis for Endurance Assessment: A Comparison of Normals with Deconditioned Patients. *Spine, 14*(9), 986-991.

Mayer, T. G., Smith, S. S., Keeley, J., & Mooney, V. (1985). Quantification of Lumbar Function: Part 2: Sagittal Plane Trunk Strength in Chronic Low-Back Pain Patients. *Spine, 10*(8), 765-772.

McArdle, W. D., Katch, F. I., & Katch, V. L. (1991). *Exercise physiology : energy, nutrition, and human performance* (3rd ed.). Philadelphia ; London: Lea & Febiger.

McCook, D. T., Vicenzino, B., & Hodges, P. W. (2009). Activity of deep abdominal muscles increases during submaximal flexion and extension efforts but antagonist co-contraction remains unchanged. *Journal of Electromyography and Kinesiology, 19*(5), 754-762.

McGill, S. M. (2007). *Low back disorders: evidence-based prevention and rehabilitation*: Human Kinetics.

McGill, S. M. (2013). Opinions on the links between back pain and motor control: the disconnect between clinical practice and research. In P. Hodges, J. Cholewicki, & J. H. Van Dieën (Eds.), *Spinal Control: The Rehabilitation of Back Pain*: Elsevier Health Sciences.

McGill, S. M. (2015). *Low Back Disorders, 3E*: Human Kinetics.

McGill, S. M., Kavcic, N. S., & Harvey, E. (2006). Sitting on a chair or an exercise ball: various perspectives to guide decision making. *Clinical Biomechanics, 21*(4), 353-360.

McGill, S. M., & Norman, R. W. (1986). Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine, 11*(7), 666-678.

McGill, S. M., Santaguida, L., & Stevens, J. (1993). Measurement of the trunk musculature from T 5 to L 5 using MRI scans of 15 young males corrected for muscle fibre orientation. *Clinical Biomechanics, 8*(4), 171-178.

McMeeken, J. M., Beith, I. D., Newham, D. J., Milligan, P., & Critchley, D. J. (2004). The relationship between EMG and change in thickness of transversus abdominis. *Clinical Biomechanics, 19*(4), 337-342.

McNeill, T., Warwick, D., Andersson, G., & Schultz, A. (1980). Trunk strengths in attempted flexion, extension, and lateral bending in healthy subjects and patients with low-back disorders. *Spine, 5*(6), 529-538.

Mens, J., Vleeming, A., Snijders, C. J., Stam, H. J., & Ginai, A. Z. (1999). The active straight leg raising test and mobility of the pelvic joints. *European Spine Journal, 8*(6), 468-473.

Miedema, H. S., Chorus, A. M. J., Wevers, C. W. J., & van der Linden, S. (1998). Chronicity of back problems during working life. *Spine, 23*(18), 2021-2028.

Miller, E. R., Schenk, R. J., Karnes, J. L., & Rousselle, J. G. (2005). A comparison of the McKenzie approach to a specific spine stabilization program for chronic low back pain. *Journal of Manual & Manipulative Therapy, 13*(2), 103-112.

Mirovsky, Y., Grober, A., Blankstein, A., & Stabholz, L. (2006). The effect of ambulatory lumbar traction combined with treadmill on patients with chronic low back pain. *Journal of Back and Musculoskeletal Rehabilitation, 19*(2), 73-78.

Morey-Klapsing, G., Arampatzis, A., & Bruggemann, G. P. (2004). Choosing EMG parameters: comparison of different onset determination algorithms and EMG integrals in a joint stability study. *Clinical Biomechanics, 19*(2), 196-201.

Moseley, G. L., Hodges, P. W., & Gandevia, S. C. (2002). Deep and superficial fibers of the lumbar multifidus muscle are differentially active during voluntary arm movements. *Spine, 27*(2), E29-36.

Moseley, G. L., Hodges, P. W., & Gandevia, S. C. (2003). External perturbation of the trunk in standing humans differentially activates components of the medial back muscles. *Journal of Physiology, 547*(2), 581-587.

Murray, C. J., Richards, M. A., Newton, J. N., Fenton, K. A., Anderson, H. R., Atkinson, C., . . . Bourne, R. (2013). UK health performance: findings of the Global Burden of Disease Study 2010. *The Lancet, 381*(9871), 997-1020.

Myklebust, J. B., Pintar, F., Yoganandan, N., Cusick, J. F., Maiman, D., Myers, T. J., & Sances Jr, A. (1988). Tensile strength of spinal ligaments. *Spine, 13*(5), 528-531.

Nachemson, A. L., & Evans, J. H. (1968). Some mechanical properties of the third human lumbar interlaminar ligament (ligamentum flavum). *Journal of Biomechanics, 1*(3), 211-220.

Nelson, B. W., O'Reilly, E., Miller, M., Hogan, M., Wegner, J., & Kelly, C. (1995). The clinical effects of intensive, specific exercise on chronic low back pain: a controlled study of 895 consecutive patients with 1-year follow up. *Orthopedics, 18*(10), 971-981.

Ng, J. K., Kippers, V., & Richardson, C. A. (1998). Muscle fibre orientation of abdominal muscles and suggested surface EMG electrode positions. *Electromyography and Clinical Neurophysiology, 38*(1), 51-58.

Nicolaisen, T., & Jørgensen, K. (1984). Trunk strength, back muscle endurance and low-back trouble. *Scandinavian Journal of Rehabilitation Medicine, 17*(3), 121-127.

Nitz, A. J., & Peck, D. (1986). Comparison of muscle spindle concentrations in large and small human epaxial muscles acting in parallel combinations. *The American Surgeon, 52*(5), 273-277.

Nordin, M., & Campello, M. (1999). Physical therapy: Exercises and the modalities: When, what and why? *Neurologic Clinics, 17*(1), 75-89.

Norris, C. M. (2008). *Back stability: integrating science and therapy*: Human Kinetics.

O'Sullivan, K., O'Sullivan, P., O'Keeffe, M., O'Sullivan, L., & Dankaerts, W. (2013). The effect of dynamic sitting on trunk muscle activation: A systematic review. *Applied Ergonomics, 44*(4), 628-635.

O'Sullivan, P. B., Dankaerts, W., Burnett, A., Straker, L., Bargon, G., Moloney, N., . . . Tsang, S. (2006a). Lumbopelvic kinematics and trunk muscle activity during sitting on stable and unstable surfaces. *Journal of Orthopaedic & Sports Physical Therapy, 36*(1), 19-25.

O'Sullivan, P. B., Dankaerts, W., Burnett, A. F., Farrell, G. T., Jefford, E., Naylor, C. S., & O'Sullivan, K. J. (2006b). Effect of different upright sitting postures on spinal-pelvic curvature and trunk muscle activation in a pain-free population. *Spine, 31*(19), E707-712.

O'Sullivan, P. B., Grahamslaw, K. M., Kendell, M., Lapenskie, S. C., Möller, N. E., & Richards, K. V. (2002). The effect of different standing and sitting postures on trunk muscle activity in a pain-free population. *Spine, 27*(11), 1238-1244.

O'Sullivan, P. B., Twomey, L., & Allison, G. T. (1998). Altered abdominal muscle recruitment in patients with chronic back pain following a specific exercise intervention. *Journal of Orthopaedic & Sports Physical Therapy, 27*(2), 114-124.

O'Sullivan, P. B., Twomey, L. T., & Allison, G. T. (1997). Evaluation of specific stabilizing exercise in the treatment of chronic low back pain with radiologic diagnosis of spondylolysis or spondylolisthesis. *Spine, 22*(24), 2959-2967.

Panjabi, M. M. (1992a). The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement. *Journal of Spinal Disorders, 5*(4), 383-389.

Panjabi, M. M. (1992b). The stabilizing system of the spine. Part II. Neutral zone and instability hypothesis. *Journal of Spinal Disorders & Techniques, 5*(4), 390-397.

Panjabi, M. M., Abumi, K., Duranceau, J., & Oxland, T. (1989). Spinal stability and intersegmental muscle forces. A biomechanical model. *Spine, 14*(2), 194-200.

Panjabi, M. M., Cholewicki, J., Nibu, K., Grauer, J., Babat, L. B., & Dvorak, J. (1998). Critical load of the human cervical spine: an in vitro experimental study. *Clinical Biomechanics, 13*(1), 11-17.

Panjabi, M. M., Crisco, J. J., Vasavada, A., Oda, T., Cholewicki, J., Nibu, K., & Shin, E. (2001). Mechanical properties of the human cervical spine as shown by three-dimensional load–displacement curves. *Spine, 26*(24), 2692-2700.

Panjabi, M. M., & White III, A. A. (1980). Basic biomechanics of the spine. *Neurosurgery, 7*(1), 76-93.

Pengel, L. H. M., Herbert, R. D., Maher, C. G., & Refshauge, K. M. (2003). Acute low back pain: systematic review of its prognosis. *British Medical Journal, 327*(7410), 323-326.

Pengel, L. H. M., Maher, C. G., & Refshauge, K. M. (2002). Systematic review of conservative interventions for subacute low back pain. *Clinical Rehabilitation, 16*(8), 811-820.

Phillips, S., Mercer, S., & Bogduk, N. (2008). Anatomy and biomechanics of quadratus lumborum. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 222*(2), 151-159.

Pietrek, M., Sheikhzadeh, A., Hagins, M., & Nordin, M. (2000). Evaluation of abdominal muscles by ultrasound imaging: reliability, and comparison to electromyography. *European Spine Journal, 9*(4), 309.

Pintar, F. A., Yoganandan, N., Myers, T., Elhagediab, A., & Sances, A. (1992). Biomechanical properties of human lumbar spine ligaments. *Journal of Biomechanics, 25*(11), 1351-1356.

Polcyn, A. F., Lipsitz, L. A., Kerrigan, D. C., & Collins, J. J. (1998). Age-related changes in the initiation of gait: degradation of central mechanisms for momentum generation. *Archives of Physical Medicine and Rehabilitation, 79*(12), 1582-1589.

Portney, L. G., & Watkins, M. P. (2008). *Foundations of clinical research: applications to practice* (3 ed.). Upper Saddle River, NJ: Prentice Hall.

Powell, P. L., Roy, R. R., Kanim, P., Bello, M. A., & Edgerton, V. R. (1984). Predictability of skeletal muscle tension from architectural determinations in guinea pig hindlimbs. *Journal of Applied Physiology, 57*(6), 1715-1721.

Putz, R. (1992). The detailed functional anatomy of the ligaments of the vertebral column. *Annals of Anatomy-Anatomischer Anzeiger, 174*(1), 40-47.

Rackwitz, B., De Bie, R., Limm, H., Von Garnier, K., Ewert, T., & Stucki, G. (2006). Segmental stabilizing exercises and low back pain. What is the evidence? A systematic review of randomized controlled trials. *Clinical Rehabilitation, 20*(7), 553-567.

Radebold, A., Cholewicki, J., Panjabi, M. M., & Patel, T. C. (2000). Muscle response pattern to sudden trunk loading in healthy individuals and in patients with chronic low back pain. *Spine, 25*(8), 947-954.

Rainoldi, A., Melchiorri, G., & Caruso, I. (2004). A method for positioning electrodes during surface EMG recordings in lower limb muscles. *Journal of Neuroscience Methods, 134*(1), 37-43.

Rankin, G., Stokes, M., & Newham, D. J. (2006). Abdominal muscle size and symmetry in normal subjects. *Muscle and Nerve, 34*(3), 320-326.

Rasmussen-Barr, E., Nilsson-Wikmar, L., & Arvidsson, I. (2003). Stabilizing training compared with manual treatment in sub-acute and chronic low-back pain. *Manual Therapy*, *8*(4), 233-241.

Rasouli, O., Arab, A. M., Amiri, M., & Jaberzadeh, S. (2011). Ultrasound measurement of deep abdominal muscle activity in sitting positions with different stability levels in subjects with and without chronic low back pain. *Manual Therapy, 16*(4), 388-393.

Reeve, A., & Dilley, A. (2009). Effects of posture on the thickness of transversus abdominis in pain-free subjects. *Manual Therapy, 14*(6), 679-684.

Reeves, N. P., Narendra, K. S., & Cholewicki, J. (2007). Spine stability: the six blind men and the elephant. *Clinical Biomechanics*, *22*(3), 266-274.

Richardson, C. A., & Hides, J. A. (2004). Closed chain segmental control. In C. A. Richardson, P. Hodges, & J. A. Hides (Eds.), *Therapeutic Exercise for Lumbopelvic Stabilization*. Edinburgh: Churchill Livingstone.

Richardson, C. A., Hodges, P. W., & Hides, J. A. (2004). *Therapeutic Exercise for Lumbo-Pelvic Stabilization: a motor control approach for the treatment of low back pain*. Edinburgh, UK: Churchill Livingstone.

Richardson, C. A., & Jull, G. A. (1995). Muscle control-pain control. What exercises would you prescribe? *Manual Therapy, 1*(1), 2-10.

Richardson, C. A., Jull, G. A., Hodges, P. W., & Hides, J. (1999). *Therapeutic Exercise for Spinal Segmental Stabilization in Low Back Pain*. Edinburgh: Churchill Livingstone.

Richardson, C. A., Snijders, C. J., Hides, J. A., Damen, L., Pas, M. S., & Storm, J. (2002). The relation between the transversus abdominis muscles, sacroiliac joint mechanics, and low back pain. *Spine, 27*(4), 399-405.

Roberts, S., McCall, I., Menage, J., Haddaway, M., & Eisenstein, S. (1997). Does the thickness of the vertebral subchondral bone reflect the composition of the intervertebral disc? *European Spine Journal, 6*(6), 385-389.

Roberts, S., Menage, J., & Urban, J. (1989). Biochemical and structural properties of the cartilage end-plate and its relation to the intervertebral disc. *Spine, 14*(2), 166-174.

Rowlands, O., Singleton, N., Maher, J., & Higgins, V. (1997). *Living in Britain*.

Roy, S. H., De Luca, C. J., & Csavant, D. A. (1989). Lumbar muscle fatigue and chronic lower back pain. *Spine, 14*(9), 992-1001.

Saunders, S. W., Coppieters, M., & Hodges, P. (2004). *Reduced Tonic Activity of the Deep Trunk Muscle During Locomotion in People with Low Back Pain*. Paper presented at the Fifth Interdisciplinary World Congress on Low Back and Pelvic Pain, Melbourne.

Saunders, S. W., Rath, D., & Hodges, P. W. (2004). Postural and respiratory activation of the trunk muscles changes with mode and speed of locomotion. *Gait and Posture, 20*(3), 280-290.

Saunders, S. W., Schache, A., Rath, D., & Hodges, P. W. (2005). Changes in three dimensional lumbo-pelvic kinematics and trunk muscle activity with speed and mode of locomotion. *Clinical Biomechanics, 20*(8), 784-793.

Savigny, P., Kuntze, S., Watson, P., Underwood, M., Ritchie, G., Cotterell, M., . . . Coffey, P. (2009). Low back pain: early management of persistent non-specific low back pain. *London: National Collaborating Centre for Primary Care and Royal College of General Practitioners, 14*.

Sayson, J. V., Lotz, J., Parazynski, S., & Hargens, A. R. (2013). Back pain in space and post-flight spine injury: Mechanisms and countermeasure development. *Acta Astronautica, 86*, 24-38.

Schollum, M. L., Robertson, P. A., & Broom, N. D. (2009). A microstructural investigation of intervertebral disc lamellar connectivity: detailed analysis of the translamellar bridges. *Journal of Anatomy, 214*(6), 805-816.

Seay, J. F., Van Emmerik, R. E. A., & Hamill, J. (2011). Influence of low back pain status on pelvis-trunk coordination during walking and running. *Spine, 36*(16), E1070-1079.

Shaughnessy, M., & Caulfield, B. (2004). A pilot study to investigate the effect of lumbar stabilisation exercise training on functional ability and quality of life in patients with chronic low back pain. *International Journal of Rehabilitation Research, 27*(4), 297-301.

Shrout, P. E., & Fleiss, J. L. (1979). Intraclass correlations: Uses in assessing rater reliability. *Psychological Bulletin, 86*, 420-428.

Sihvonen, T., Lindgren, K.-A., Airaksinen, O., & Manninen, H. (1997). Movement disturbances of the lumbar spine and abnormal back muscle electromyographic findings in recurrent low back pain. *Spine, 22*(3), 289-295.

Sions, J. M., Velasco, T. O., Teyhen, D. S., & Hicks, G. E. (2014). Ultrasound Imaging: Intraexaminer and Interexaminer Reliability for Multifidus Muscle Thickness Assessment in Adults Aged 60 to 85 Years Versus Younger Adults. *Journal of Orthopaedic & Sports Physical Therapy, 44*(6), 425-434.

Skinner, T. (Ed.) (1997). *National Health Survey. Summary of Results.* (Vol. ABS catalogue number 4364.0). Canberra: Australian Bureau of Statistics.

Smeets, R. J. E. M., Wittink, H., Hidding, A., & Knottnerus, J. A. (2006). Do patients with chronic low back pain have a lower level of aerobic fitness than healthy controls?: are pain, disability, fear of injury, working status, or level of leisure time activity associated with the difference in aerobic fitness level? *Spine, 31*(1), 90-97.

Smith, B. E., Littlewood, C., & May, S. (2014). An update of stabilisation exercises for low back pain: a systematic review with meta-analysis. *BMC Musculoskeletal Disorders, 15*(1), 416.

Solomonow, M., Zhou, B.-H., Harris, M., Lu, Y., & Baratta, R. V. (1998). The ligamento-muscular stabilizing system of the spine. *Spine, 23*(23), 2552-2562.

Stackhouse, C., Shewokis, P. A., Pierce, S. R., Smith, B., McCarthy, J., & Tucker, C. (2007). Gait initiation in children with cerebral palsy. *Gait and Posture, 26*(2), 301-308.

Stanton, T. R., Henschke, N., Maher, C. G., Refshauge, K. M., Latimer, J., & McAuley, J. H. (2008). After an episode of acute low back pain, recurrence is unpredictable and not as common as previously thought. *Spine, 33*(26), 2923-2928.

Stokes, I. A., Gardner-Morse, M. G., & Henry, S. M. (2010). Intra-abdominal pressure and abdominal wall muscular function: Spinal unloading mechanism. *Clinical Biomechanics, 25*(9), 859-866.

Stokes, I. A., Gardner-Morse, M. G., & Henry, S. M. (2011). Abdominal muscle activation increases lumbar spinal stability: analysis of contributions of different muscle groups. *Clinical Biomechanics, 26*(8), 797-803.

Stokes, I. A., Henry, S. M., & Single, R. M. (2003). Surface EMG electrodes do not accurately record from lumbar multifidus muscles. *Clinical Biomechanics, 18*(1), 9-13.

Stokes, M., Hides, J., Elliott, J., Kiesel, K., & Hodges, P. (2007). Rehabilitative ultrasound imaging of the posterior paraspinal muscles. *Journal of Orthopaedic & Sports Physical Therapy, 37*(10), 581-595.

Stokes, M., Rankin, G., & Newham, D. J. (2005). Ultrasound imaging of lumbar multifidus muscle: normal reference ranges for measurements and practical guidance on the technique. *Manual Therapy, 10*(2), 116-126.

Streicher, H., Mätzold, F., Hamilton, C., & Wagner, P. (2014). Comparison of group motor control training versus individual training for people suffering from back pain. *Journal of Bodywork and Movement Therapies, 18*(3), 489-496.

Strohl, K. P., Mead, J., Banzett, R. B., Loring, S. H., & Kosch, P. C. (1981). Regional differences in abdominal muscle activity during various maneuvers in humans. *Journal of Applied Physiology, 51*(6), 1471-1476.

Sweeney, N., O'Sullivan, C., & Kelly, G. (2014). Multifidus Muscle Size and Percentage Thickness changes among Patients with Unilateral Chronic LBP and Healthy Controls in Prone and Standing. *Manual Therapy, 19*(5), 433-439.

Szabo, T. L. (2004). *Diagnostic ultrasound imaging: inside out*. Boston: Elsevier Science.

Taimela, S., & Härkäpää, K. (1996). Strength, mobility, their changes, and pain reduction in active functional restoration for chronic low back disorders. *Journal of Spinal Disorders & Techniques, 9*(4), 306-312.

Taylor, N. F., Evans, O. M., & Goldie, P. A. (2003). The effect of walking faster on people with acute low back pain. *European Spine Journal, 12*(2), 166-172.

Teyhen, D. S. (2006). Rehabilitative Ultrasound Imaging Symposium *Journal of Orthopaedic & Sports Physical Therapy, 36*(8), A1-A3.

Teyhen, D. S., Bluemle, L. N., Dolbeer, J. A., Baker, S. E., Molloy, J. M., Whittaker, J., & Childs, J. D. (2009a). Changes in lateral abdominal muscle thickness during the abdominal drawing-in maneuver in those with lumbopelvic pain. *Journal of Orthopaedic & Sports Physical Therapy, 39*(11), 791-798.

Teyhen, D. S., Childs, J. D., Stokes, M. J., Wright, A. C., Dugan, J. L., & George, S. Z. (2012). Abdominal and Lumbar Multifidus Muscle Size and Symmetry at Rest and During Contracted States Normative Reference Ranges. *Journal of Ultrasound in Medicine, 31*(7), 1099-1110.

Teyhen, D. S., George, S. Z., Dugan, J. L., Williamson, J., Neilson, B. D., & Childs, J. D. (2011). Inter-rater reliability of ultrasound imaging of the trunk musculature among novice raters. *Journal of Ultrasound in Medicine, 30*(3), 347-356.

Teyhen, D. S., Gill, N. W., Whittaker, J. L., Henry, S. M., Hides, J. A., & Hodges, P. (2007). Rehabilitative ultrasound imaging of the abdominal muscles. *Journal of Orthopaedic & Sports Physical Therapy, 37*(8), 450-466.

Teyhen, D. S., Miltenberger, C. E., Deiters, H. M., Del Toro, Y. M., Pulliam, J. N., Childs, J. D., . . . Flynn, T. W. (2005). The use of ultrasound imaging of the abdominal drawing-in maneuver in subjects with low back pain. *Journal of Orthopaedic & Sports Physical Therapy, 35*(6), 346-355.

Teyhen, D. S., Rieger, J. L., Westrick, R. B., Miller, A. C., Molloy, J. M., & Childs, J. D. (2008). Changes in deep abdominal muscle thickness during common trunk-strengthening exercises using ultrasound imaging. *Journal of Orthopaedic & Sports Physical Therapy, 38*(10), 596-605.

Teyhen, D. S., Williamson, J. N., Carlson, N. H., Suttles, S. T., O'Laughlin, S. J., Whittaker, J. L., . . . Childs, J. D. (2009b). Ultrasound characteristics of the deep abdominal muscles during the active straight leg raise test. *Archives of Physical Medicine and Rehabilitation, 90*(5), 761-767.

Thorstensson, A., Carlson, H., Zomlefer, M. R., & Nilsson, J. (1982). Lumbar back muscle activity in relation to trunk movements during locomotion in man. *Acta Physiologica Scandinavica, 116*(1), 13-20.

Torstensen, T. A., Ljunggren, A. E., Meen, H. D., Odland, E., Mowinckel, P., & Geijerstam, S. (1998). Efficiency and costs of medical exercise therapy, conventional physiotherapy, and self-exercise in patients with chronic low back pain. A pragmatic, randomized, single-blinded, controlled trial with 1-year follow-up. *Spine, 23*(23), 2616-2624.

Tsao, H., Danneels, L. A., & Hodges, P. W. (2011). ISSLS prize winner: smudging the motor brain in young adults with recurrent low back pain. *Spine, 36*(21), 1721-1727.

Tsao, H., & Hodges, P. W. (2007). Immediate changes in feedforward postural adjustments following voluntary motor training. *Experimental Brain Research, 181*(4), 537-546.

Tsao, H., & Hodges, P. W. (2008). Persistence of improvements in postural strategies following motor control training in people with recurrent low back pain. *Journal of Electromyography and Kinesiology, 18*(4), 559-567.

Unsgaard-Tøndel, M., Fladmark, A. M., Salvesen, Ø., & Vasseljen, O. (2010). Motor control exercises, sling exercises, and general exercises for patients with chronic low back pain: a randomized controlled trial with 1-year follow-up. *Physical Therapy, 90*(10), 1426-1440.

Urwin, M., Symmons, D., Allison, T., Brammah, T., Busby, H., Roxby, M., . . . Williams, G. (1998). Estimating the burden of musculoskeletal disorders in the community: the comparative prevalence of symptoms at different anatomical sites, and the relation to social deprivation. *Annals of the Rheumatic Diseases, 57*(11), 649-655.

van den Hoogen, H. J. M., Koes, B. W., van Eijk, J. T. M., Bouter, L. M., & Devillé, W. (1998). On the course of low back pain in general practice: a one year follow up study. *Annals of the Rheumatic Diseases, 57*(1), 13-19.

van der Hulst, M., Vollenbroek-Hutten, M. M., Rietman, J. S., Schaake, L., Groothuis-Oudshoorn, K. G., & Hermens, H. J. (2010). Back muscle activation patterns in chronic low back pain during walking: a "guarding" hypothesis. *The Clinical Journal of Pain, 26*(1), 30-37.

van Dieën, J. H., Selen, L. P. J., & Cholewicki, J. (2003). Trunk muscle activation in low-back pain patients, an analysis of the literature. *Journal of Electromyography and Kinesiology, 13*(4), 333-351.

Van, K., Hides, J. A., & Richardson, C. A. (2006). The use of real-time ultrasound imaging for biofeedback of lumbar multifidus muscle contraction in healthy subjects. *Journal of Orthopaedic & Sports Physical Therapy, 36*(12), 920-925.

Van Tulder, M., Becker, A., Bekkering, T., Breen, A., Gil del Real, M. T., Hutchinson, A., . . . Malmivaara, A. (2006). Chapter 3 European guidelines for the management of acute nonspecific low back pain in primary care. *European Spine Journal, 15*, s169-s191.

van Tulder, M., Malmivaara, A., Esmail, R., & Koes, B. (2000). Exercise therapy for low back pain: a systematic review within the framework of the cochrane collaboration back review group. *Spine, 25*(21), 2784-2796.

Van Tulder, M. W., Koes, B. W., & Bouter, L. M. (1997). Conservative treatment of acute and chronic nonspecific low back pain: a systematic review of randomized controlled trials of the most common interventions. *Spine, 22*(18), 2128-2156.

Varlotta, G. P., Lefkowitz, T. R., Schweitzer, M., Errico, T. J., Spivak, J., Bendo, J. A., & Rybak, L. (2011). The lumbar facet joint: a review of current knowledge: part 1: anatomy, biomechanics, and grading. *Skeletal Radiology, 40*(1), 13-23.

Vera-Garcia, F. J., Elvira, J. L., Brown, S. H., & McGill, S. M. (2007). Effects of abdominal stabilization maneuvers on the control of spine motion and stability against sudden trunk perturbations. *Journal of Electromyography and Kinesiology, 17*(5), 556-567.

Verbunt, J. A., Seelen, H. A., Vlaeyen, J. W., Heijden, G. J., Heuts, P. H., Pons, K., & Knottnerus, J. A. (2003). Disuse and deconditioning in chronic low back pain: concepts and hypotheses on contributing mechanisms. *European Journal of Pain, 7*(1), 9-21.

Vink, P., Van Der Velde, E., & Verbout, A. (1987). A functional subdivision of the lumbar extensor musculature. Recruitment patterns and force-RA-EMG relationships under isometric conditions. *Electromyography and Clinical Neurophysiology, 27*(8), 517-525.

Vlaeyen, J. W. S., & Linton, S. J. (2000). Fear-avoidance and its consequences in chronic musculoskeletal pain: a state of the art. *Pain, 85*(3), 317-332.

Von Korff, M., & Saunders, K. (1996). The course of back pain in primary care. *Spine, 21*(24), 2833-2837.

Waddell, G. (1987). 1987 Volvo Award in Clinical Sciences: a new clinical model for the treatment of low-back pain. *Spine, 12*(7), 632-644.

Walker, B. F. (2000). The prevalence of low back pain: a systematic review of the literature from 1966 to 1998. *Journal of Spinal Disorders & Techniques, 13*(3), 205-217.

Wallwork, T. L., Stanton, W. R., Freke, M., & Hides, J. A. (2009). The effect of chronic low back pain on size and contraction of the lumbar multifidus muscle. *Manual Therapy, 14*(5), 496-500.

Wand, B. M., & O'Connell, N. E. (2008). Chronic non-specific low back pain–sub-groups or a single mechanism? *BMC Musculoskeletal Disorders, 9*(11), 1-15.

Wang, S., & McGill, S. M. (2008). Links between the mechanics of ventilation and spine stability. *Journal of Applied Biomechanics, 24*(2), 166-174.

Ward, S. R., Kim, C. W., Eng, C. M., Gottschalk, L. J. t., Tomiya, A., Garfin, S. R., & Lieber, R. L. (2009). Architectural analysis and intraoperative measurements demonstrate the unique design of the multifidus muscle for lumbar spine stability. *Journal of Bone & Joint Surgery, American Volume, 91*(1), 176-185.

Warwick, R., & Williams, P. L. (1973). *Gray's Anatomy* (35 ed.). London: Longmans.

Weinstein, S. L. (2000). 2000–2010: the bone and joint decade. *The Journal of Bone & Joint Surgery, 82*(1), 1-3.

White III, A. A., Johnson, R. M., Panjabi, M. M., & Southwick, W. O. (1975). Biomechanical analysis of clinical stability in the cervical spine. *Clinical Orthopaedics and Related Research, 109*, 85-96.

Whittaker, J. L., Teyhen, D. S., Elliott, J. M., Cook, K., Langevin, H. M., Dahl, H. H., & Stokes, M. (2007). Rehabilitative ultrasound imaging: understanding the technology and its applications. *Journal of Orthopaedic & Sports Physical Therapy, 37*(8), 434-449.

Whittaker, J. L., Warner, M. B., & Stokes, M. J. (2009). Induced transducer orientation during ultrasound imaging: effects on abdominal muscle thickness and bladder position. *Ultrasound in Medicine & Biology, 35*(11), 1803-1811.

Wild, J. J., & Neal, D. (1951). Use of high frequency ultrasonic waves for detecting changes of texture in living tissues. *The Lancet, 257*(6656), 655-657.

Wilke, H.-J., Wolf, S., Claes, L. E., Arand, M., & Wiesend, A. (1995). Stability Increase of the Lumbar Spine With Different Muscle Groups: A Biomechanical In Vitro Study. *Spine, 20*(2), 192-197.

Wong, A. Y., Parent, E. C., Funabashi, M., Stanton, T. R., & Kawchuk, G. N. (2013). Do various baseline characteristics of transversus abdominis and lumbar multifidus predict clinical outcomes in nonspecific low back pain? A systematic review. *PAIN®, 154*(12), 2589-2602.

Wood, P. H. N., & Badley, E. M. (1980). Epidemiology of Back Pain. In M. I. V. Jayson (Ed.), *The Lumbar Spine and Back Pain* (2 ed., pp. 29-55). Tunbridge Wells: Pitman Medical.

Woolf, A. D., & Pfleger, B. (2003). Burden of major musculoskeletal conditions. *Bulletin of the World Health Organization, 81*(9), 646-656.

Yahia, L. H., Garzon, S., Strykowski, H., & Rivard, C. H. (1990). Ultrastructure of the Human Interspinous Ligament and Ligamentum Flavum A Preliminary Study. *Spine, 15*(4), 262-268.

Yong-Hing, K., Reilly, J., & Kirkaldy-Willis, W. (1976). The ligamentum flavum. *Spine, 1*(4), 226-234.

Young, A., Hughes, I., Russell, P., Parkers, M. J., & Nichols, P. J. (1980). Measurement of quadriceps muscle wasting by ultrasonography. *Rheumatology and Rehabilitation, 19*(3), 141-148.

Zielinski, K. A., Henry, S. M., Ouellette-Morton, R. H., & DeSarno, M. J. (2013). Lumbar multifidus muscle thickness does not predict patients with low back pain who improve with trunk stabilization exercises. *Archives of Physical Medicine and Rehabilitation, 94*(6), 1132-1138.

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Appendix H – Participant Information Sheet (Chapters III and IV)

PARTICIPANT INFORMATION

Project Title: Assessing the Reliability of Freehand Muscle Thickness Measurements Using Rehabilitative Ultrasound Imaging During Dynamic Movements

Participant ID Number: _____

Principal Investigator: Karl Christian Gibbon

Investigator contact details: Telephone: 0191 243 7018 Email: karl.gibbon@unn.ac.uk

This project is funded by: Northumbria University

INFORMATION TO POTENTIAL PARTICIPANTS

1. What is the purpose of the project?

The purpose of this study is to investigate how reliable and accurate ultrasound imaging of the abdomen and lower back muscles is during movement.

This information is being collected in preparation for future investigations concerning comparisons between treadmill walking and using a new exercise device. The new exercise device is similar to an elliptical trainer; however, the activity is much less demanding at a slower speed.

2. Why have I been selected to take part?

You have been asked to take part because you are a healthy individual, 18 years of age or older, with no history of muscle, skeletal or neuromuscular diseases/injuries, and abdominal or spinal surgery. You also have not experienced low back pain within the past six months (requiring medical consultation/treatment) and are not currently pregnant.

3. What will I have to do?

You will be asked to visit the laboratory for approximately ***one and a half hours*** on ***three*** occasions within a ***one week*** period. Each visit will be separated by ***two*** days (e.g. Tuesday, Friday and Monday).

During each visit we will be using ultrasound imaging to investigate two different muscles, one in the lower back (*lumbar multifidus*) and one in the abdomen (*transversus abdominis*), during both resting and contracted states on both sides of the body.

Ultrasound imaging uses sound waves to produce an image of the tissues beneath the skin. This is a safe technique and used on a daily basis worldwide. The technology is identical to that which is used for checking on the development of babies within the womb during pregnancy.

Each visit will contain the following assessments;

Lower Back

1) Rest – Images will be taken with you lying on your front on a medical bed with pillows placed under your abdomen.

2) Unloaded Arm Rise – As rest condition, however you will be required raise one arm off the bed by approximately 5 centimetres. This will be repeated for the second arm.

3) Loaded Arm Rise – As unloaded arm rise, however you will raise your arm whilst holding a weight of 0.68kg or 0.9kg, depending on your body mass.

4) Treadmill Walking – You will be asked to walk on a treadmill at a self-selected comfortable and maintainable speed. Images will be collected during this activity.

5) Elliptical Trainer – As treadmill walking, however participants will be asked to use the exercise device at a comfortable and maintainable speed.

Abdomen

1. Rest – You will lie on your back on a medical bed with your knees slightly bent.

2. Straight Leg Raise – As rest condition, however you will be asked to raise one leg off the bed by approximately 20cm. This will be repeated for your opposite leg also.

3. Abdominal Drawing-in – As rest condition, however you will be asked to voluntarily contract your transversus abdominis with minimal co-activation of surrounding musculature

4. Treadmill Walking – You will be asked to walk on a treadmill at a self-selected comfortable and maintainable speed. Images will be collected during this activity.

5. Elliptical Trainer – As treadmill walking, however participants will be asked to use the exercise device at a comfortable and maintainable speed.

Please also refrain from consuming food within the three hours prior to your scheduled testing session. Water consumption is permitted.

4. What are the exclusion criteria (i.e. are there any reasons why I should not take part)?

- Below 18 years of age
- History musculoskeletal or neuromuscular pathologies/injuries
- History of abdominal or spinal surgery
- Experienced low back pain (requiring medical consultation/treatment) within the previous 6 months
- Current pregnancy

5. Will my participation involve any physical discomfort?

Ultrasound transmission gel will be applied to the abdomen and lower back at numerous points throughout each testing session. This can have a cold feeling initially, however body temperature quickly counteracts this feeling. Surplus gel will also be removed when no longer needed.

Treadmill walking and using the exercise device will cause an increase in breathing and heart rates; however the level of activity will not be anything above what is likely to take place during everyday activities.

6. Will my participation involve any psychological discomfort or embarrassment?

For the duration of each assessment your abdomen and lower back will be exposed and this may cause you embarrassment if you are conscious of those areas of your body. We will ensure that the testing environment is as private as possible and not overlooked.

7. Will I have to provide any bodily samples (i.e. blood, saliva)?

No.

8. How will confidentiality be assured?

You will be allocated a numeric participant code and all referrals to data relating to yourself shall be noted with that code. All data will be stored within a locked filing cabinet or on a password protected desktop computer in accordance with the Data Protection Act (1998).

9. Who will have access to the information that I provide?

Information you provide and the data we collect will be seen only by the principal investigator (Karl Gibbon) and their supervisors (Dr Nicholas Caplan & Dr Dorothée Debuse). All records will be kept confidential except for review by Northumbria University Ethics Committee and regulatory authorities.

10. How will my information be stored / used in the future?

Your information will be stored on a password-protected computer or in a locked filling cabinet. All information will be used solely for the purposes of this investigation. Any personal information will be destroyed after 3 years. Data may be published in peer-reviewed journals or presented as posters/abstracts at conferences; however all data will be grouped and any personal information will not be referred to at any time.

11. Has this investigation received appropriate ethical clearance?

The study has received full ethical approval from the school of Life Sciences Ethics Committee. If you require confirmation of this please contact the chair of the Committee, stating the title of the research project and the name of the Principal Investigator:

Nick Neave
Chair of School of Psychology and Sports Sciences
Ethics Committee
Northumberland Building
Northumbria University
Newcastle Upon Tyne
NE1 8ST

12. Will I receive any financial rewards / travel expenses for taking part?

No.

13. How can I withdraw from the project?

You can withdraw your data from this project at anytime, without need for explanation or justification. To do this simply contact the principal investigator by telephone, email or in person.

14. If I require further information who should I contact and how?

Any further information required for this study can be obtain from the principle investigator:

Karl Christian Gibbon
School of Life Sciences
Department of Sport & Exercise Sciences
Northumberland Building (NB431)
Northumbria University
Newcastle-upon-Tyne
NE1 8ST
Tel: 0191 243 7018
E-mail: karl.gibbon@unn.ac.uk

If you would like to discuss the study, withdraw your data or register a complaint please contact the chair of the ethics committee on the address listed in section 11.

Appendix I – Blank Consent Form (Chapters III and IV)

INFORMED CONSENT FORM

Project Title: Assessing the Reliability of Freehand Muscle Thickness Measurements Using Rehabilitative Ultrasound Imaging During Dynamic Movements

Principal Investigator: Karl Christian Gibbon

Participant Number: _____

	please tick where applicable
I have read and understood the Participant Information Sheet.	☐
I have had an opportunity to ask questions and discuss this study and I have received satisfactory answers.	☐
I understand I am free to withdraw from the study at any time, without having to give a reason for withdrawing, and without prejudice.	☐
I agree to take part in this study.	☐
I would like to receive feedback on the overall results of the study at the email address given below. I understand that I will not receive individual feedback on my own performance. Email address……	☐

Signature of participant..... Date.....……..

(NAME IN BLOCK LETTERS).....……

Signature of Parent / Guardian in the case of a minor

.....

Signature of researcher..... Date.....……..

(NAME IN BLOCK LETTERS).....……

Appendix J – Summary Raw Data Table for Lumbar Multifidus Ultrasound Thickness Measurements (Chapters III and IV)

	Rest									***CAL***									***LCAL***								
	Visit 1			***Visit 2***			***Visit 3***			***Visit 1***			***Visit 2***			***Visit 3***			***Visit 1***			***Visit 2***			***Visit 3***		
Participant #	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
001	28.0	30.3	30.8	27.6	30.6	30.1	28.7	28.2	30.6	34.1	31.9	32.3	34.2	36.5	34.1	33.2	32.6	34.1	42.7	39.6	41.6	38.6	42.1	39.3	39.5	41.1	41.0
002	34.8	37.3	38.9	35.3	35.0	32.9	36.9	35.0	37.5	43.3	42.0	43.5	43.2	39.9	42.6	42.4	43.5	42.6	46.8	46.3	48.9	47.7	46.9	49.5	46.5	51.3	50.8
003	21.3	23.5	23.5	23.5	23.4	23.0	24.3	22.8	24.6	28.4	26.1	27.5	27.0	31.4	29.6	26.3	26.1	24.3	27.0	30.4	28.4	28.7	28.5	30.3	29.6	30.7	29.9
004	27.4	28.8	26.4	30.3	28.4	29.4	24.4	25.6	24.5	33.4	36.2	33.6	36.9	35.1	31.3	27.4	33.4	31.3	35.5	38.6	33.7	36.8	36.4	35.3	38.9	37.5	36.8
005	31.9	33.9	31.7	31.7	34.8	33.4	34.9	32.5	31.3	44.5	43.0	45.9	43.9	43.2	46.6	44.7	46.0	48.3	47.7	49.3	46.5	45.1	48.2	48.4	48.8	50.3	49.7
006	35.8	37.8	37.1	34.2	34.3	33.5	35.3	36.4	33.3	40.1	39.5	38.2	40.4	40.0	39.8	42.5	37.6	37.7	51.5	47.4	51.3	47.3	47.0	49.2	50.3	48.9	48.6
007	39.3	39.8	39.4	39.7	37.3	36.9	39.8	40.4	37.6	45.3	45.0	42.2	45.0	42.4	42.6	45.8	43.3	44.1	50.6	52.3	49.2	54.0	50.1	51.5	50.6	51.2	53.3
008	23.0	21.7	21.7	22.6	23.9	22.6	21.4	23.0	23.4	27.3	26.9	25.0	25.9	24.6	28.3	22.6	25.1	27.5	29.2	33.1	29.1	29.8	31.4	31.1	29.4	31.0	29.9
009	35.0	32.2	33.5	34.2	36.0	34.3	36.5	36.2	35.7	42.2	47.0	42.5	43.0	43.3	45.2	41.6	41.5	43.2	48.9	52.4	50.7	50.2	50.4	48.2	52.6	46.2	50.8
010	27.3	25.6	25.7	21.5	23.7	24.4	23.8	22.0	24.9	26.0	26.5	26.1	29.0	25.5	25.7	27.3	29.6	28.9	32.6	30.4	32.7	30.1	32.7	31.9	33.1	31.7	31.6
011	22.6	23.0	21.7	23.1	23.8	22.8	23.7	22.5	25.9	30.1	25.6	29.7	26.3	25.6	27.8	24.4	26.3	25.7	30.5	33.3	31.0	32.2	30.1	32.6	31.6	33.3	32.0
012	19.2	21.4	19.4	21.8	22.3	21.5	20.8	20.5	22.1	26.5	24.4	25.4	26.5	29.4	26.7	25.5	25.3	27.3	29.7	31.3	31.2	30.7	31.5	31.8	30.2	33.3	34.2
013	25.6	25.2	27.5	22.8	26.5	27.6	26.5	29.2	27.7	34.4	34.1	33.9	36.2	34.2	35.5	32.6	33.2	37.9	38.4	39.7	42.6	38.6	41.2	40.3	44.1	40.3	39.5
014	24.4	25.1	24.3	23.4	23.7	24.5	24.7	24.9	24.8	26.2	28.9	26.4	27.7	27.7	28.1	27.8	27.2	26.1	40.1	40.1	43.9	43.4	39.4	43.1	44.3	41.8	39.8
015	39.3	40.4	37.8	34.7	36.1	38.4	38.9	40.2	37.5	48.9	48.7	45.6	47.1	49.1	47.4	45.9	49.5	45.6	58.4	55.7	54.1	56.1	56.3	55.4	57.7	57.8	55.0

Treadmill Walk									*FRED*								
Visit 1			*Visit 2*			*Visit 3*			*Visit 1*			*Visit 2*			*Visit 3*		
1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
46.9	49.4	49.5	45.8	40.5	48.1	49.0	48.1	49.7	37.5	45.4	41.8	40.80	38.60	37.80	37.80	43.10	43.30
56.4	54.4	56.4	49.7	44.3	49.3	49.1	51.4	46.0	39.7	45.5	50.2	50.70	47.30	43.50	52.80	51.90	45.20
34.7	39.2	35.5	37.5	35.8	38.1	30.9	35.1	38.6	27.9	30.6	29.8	34.40	35.90	32.30	28.60	30.10	31.90
39.5	44.1	41.1	43.3	38.5	40.9	38.7	37.5	42.3	31.7	30.5	39.8	33.40	36.40	35.00	35.80	33.10	38.70
59.5	48.8	57.0	44.8	53.8	55.8	49.5	47.7	41.3	43.1	49.9	45.7	49.60	47.70	45.70	53.00	53.50	47.30
59.3	57.8	52.6	47.8	57.1	57.0	46.7	43.2	50.2	38.2	56.1	52.6	42.40	53.10	51.50	58.10	51.00	51.30
57.9	58.8	50.3	50.0	57.8	53.2	57.3	56.1	53.9	40.7	58.1	50.5	51.70	48.50	61.80	58.00	49.40	52.70
31.0	35.3	31.9	33.7	32.7	34.2	29.0	29.3	30.7	21.8	32.6	31.5	32.80	31.40	33.10	32.70	29.40	32.10
52.1	49.7	51.1	46.5	47.6	43.4	48.3	48.6	47.5	47.0	42.4	49.8	52.60	44.50	57.70	47.50	53.90	56.10
37.3	37.8	40.3	38.4	37.3	38.0	33.7	37.3	32.4	25.9	28.7	32.8	31.70	28.50	35.20	33.10	30.40	31.60
34.8	29.3	38.4	34.6	28.8	33.5	28.2	30.4	31.0	26.7	29.3	27.0	33.70	32.40	34.10	29.00	28.90	33.40
28.9	30.7	26.1	32.8	32.3	28.9	31.9	32.7	28.4	24.9	29.8	30.5	28.10	30.80	29.20	31.00	34.50	30.90
47.2	48.1	49.8	47.6	46.7	41.2	37.8	41.2	42.2	37.7	38.4	37.8	43.60	39.70	40.80	36.40	40.70	43.80
48.0	40.8	43.7	48.2	46.8	50.2	41.5	42.2	45.2	28.8	41.7	46.5	46.10	45.80	43.50	41.00	48.50	42.30
53.4	51.2	50.6	50.7	46.1	50.9	47.9	47.7	47.8	49.2	61.7	54.0	55.80	51.30	49.30	61.20	53.00	55.60

Appendix K – Summary Raw Data Table for Transversus Abdominis Ultrasound Thickness Measurements (Chapters III and IV)

	Rest									***ADIM***									***ASLR***								
	Visit 1			***Visit 2***			***Visit 3***			***Visit 1***			***Visit 2***			***Visit 3***			***Visit 1***			***Visit 2***			***Visit 3***		
Participant #	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
001	3.5	3.2	3.3	3.7	3.2	3.8	3.0	4.1	3.7	4.0	4.7	4.9	5.7	5.9	4.4	5.3	5.7	5.8	3.8	3.9	4.2	4.1	4.0	4.6	4.3	4.1	4.7
002	4.8	3.5	3.8	3.6	3.6	3.8	3.8	3.2	3.2	5.1	5.9	5.4	6.0	6.9	5.2	5.3	5.9	5.6	4.4	5.3	4.9	4.2	4.9	4.3	4.5	4.9	4.2
003	6.5	5.6	6.2	6.2	5.4	6.5	6.4	5.0	6.4	8.7	9.5	9.0	7.5	7.9	7.8	8.5	7.9	7.5	7.9	7.4	8.5	7.3	7.1	6.1	7.0	7.3	6.7
004	5.2	5.0	5.6	5.7	6.4	5.7	5.5	5.4	5.8	6.1	7.7	7.9	8.2	8.5	8.5	8.4	8.7	8.1	6.5	6.7	6.3	6.5	6.2	6.4	6.0	6.7	6.5
005	7.6	7.0	8.1	7.4	7.8	7.5	7.9	7.8	7.6	8.6	8.6	9.5	9.2	8.0	8.4	9.6	9.5	10.4	8.2	8.9	8.4	8.3	7.5	8.3	7.9	8.9	8.4
006	3.3	3.8	3.7	3.8	3.4	3.2	3.7	3.8	3.4	4.9	5.6	5.0	4.6	4.9	4.6	5.2	5.5	5.7	5.1	4.7	4.9	4.7	4.8	4.4	4.8	4.6	4.7
007	3.4	3.8	3.6	3.7	3.7	3.9	3.6	3.0	3.5	4.9	4.9	4.3	5.5	5.8	6.0	4.2	4.6	6.9	4.3	3.9	4.0	3.9	4.5	4.2	4.2	4.0	4.3
008	2.9	3.5	3.0	3.2	3.3	3.7	3.2	3.4	3.8	4.3	4.4	4.9	4.8	4.7	4.2	4.3	4.7	4.2	4.5	4.7	4.7	4.7	4.9	5.1	5.5	4.9	4.5
009	8.8	8.4	8.5	9.1	8.6	8.9	7.6	7.9	8.0	10.4	10.8	10.1	10.6	11.4	11.2	10.6	10.7	10.1	12.4	11.7	13.0	11.1	11.9	11.8	12.0	11.4	12.4
010	3.1	3.9	3.2	3.2	3.8	4.0	3.7	3.7	3.2	4.7	4.9	5.1	4.7	4.6	4.4	5.3	5.9	4.5	4.2	4.7	4.7	4.6	5.0	4.2	4.3	5.1	4.2
011	2.1	2.5	2.7	2.6	3.6	2.4	2.7	2.5	2.9	3.4	4.3	3.2	3.6	4.3	3.7	4.0	3.5	3.8	4.6	4.2	3.8	4.5	4.8	4.3	4.5	5.0	4.6
012	2.9	3.7	3.2	3.3	2.7	3.2	3.2	3.3	3.2	6.0	6.4	6.3	5.3	5.9	5.5	5.7	5.3	5.7	6.3	5.0	6.0	6.2	6.8	6.7	6.5	6.3	6.9
013	3.8	3.8	3.6	4.0	3.6	3.8	3.8	4.1	3.5	5.7	4.5	5.9	7.0	4.5	4.8	5.7	5.3	5.8	5.5	5.8	4.2	5.0	4.4	4.7	5.1	5.5	5.2
014	2.9	3.4	3.3	3.2	3.0	3.3	3.1	3.1	3.0	4.5	4.9	5.4	5.2	4.9	5.3	4.7	4.3	4.7	3.9	4.2	4.9	3.7	4.5	4.4	4.1	4.9	4.5
015	3.3	3.6	3.2	3.4	3.8	3.6	3.6	3.2	3.7	6.8	3.9	5.7	6.3	6.5	5.0	5.4	5.8	5.2	3.7	3.9	4.0	4.2	3.9	4.2	3.8	4.6	4.0

Treadmill Walk									*FRED*								
Visit 1			*Visit 2*			*Visit 3*			*Visit 1*			*Visit 2*			*Visit 3*		
1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
4.3	5.0	4.1	5.2	5.9	4.3	4.9	4.7	4.1	4.6	4.9	4.1	5.1	4.4	4.6	5.0	4.5	4.5
3.7	4.6	4.0	4.2	3.8	5.0	3.7	4.2	4.2	5.1	6.7	6.0	6.9	6.1	6.2	5.6	5.8	6.5
8.2	8.9	9.1	7.6	8.2	7.1	7.3	8.5	8.1	7.4	7.9	7.8	8.1	7.6	8.2	8.5	8.6	8.1
5.8	5.8	6.5	7.1	6.2	6.8	7.5	7.2	8.3	7.6	7.2	8.2	7.3	7.8	7.2	7.6	7.9	8.5
9.2	9.0	7.8	9.3	9.5	10.3	10.3	8.8	9.0	11.5	11.9	12.4	12.7	12.9	11.5	11.7	11.2	12.3
6.5	6.1	5.7	6.4	6.3	6.1	5.1	5.8	6.1	5.4	5.9	5.1	5.7	5.3	6.1	6.1	6.8	6.1
4.5	4.7	5.7	3.9	5.7	4.9	4.0	4.5	4.7	4.2	4.5	4.3	5.0	4.2	4.5	5.0	5.7	4.3
3.0	5.6	3.5	4.1	3.2	3.9	3.1	4.7	4.3	4.2	4.6	3.9	4.3	4.9	4.1	5.1	4.2	4.2
9.3	10.4	10.5	8.9	9.6	9.4	9.9	10.8	9.5	10.3	11.2	10.7	9.5	10.5	10.1	10.3	10.2	8.7
6.3	6.2	4.3	4.0	5.6	5.2	4.3	4.9	3.9	5.1	4.2	5.2	4.5	4.7	4.5	4.3	4.7	4.6
3.1	3.9	3.0	4.1	3.6	3.8	4.1	3.5	4.2	3.3	3.6	4.1	3.8	4.7	3.6	4.1	3.6	4.5
4.6	4.2	5.9	3.8	3.7	3.5	4.6	3.9	3.7	4.7	4.5	4.4	3.5	4.1	3.2	3.8	3.9	4.5
6.1	6.3	5.9	5.6	5.2	6.0	5.0	5.5	6.1	6.2	5.8	6.4	6.3	6.6	5.4	6.4	5.5	6.4
4.1	4.9	5.2	3.7	3.6	2.9	3.5	3.7	4.4	5.0	4.6	4.5	4.3	4.7	4.1	4.8	4.3	4.1
4.8	4.2	4.0	5.1	5.6	5.0	4.9	4.0	5.1	4.1	3.6	3.8	4.5	4.2	4.7	5.3	4.4	4.6

Appendix L - Participant Information Sheet (Chapter V)

PARTICIPANT INFORMATION

Project Title: Evaluation of the Effectiveness of a Range of Rehabilitative Exercises in Recruiting Local Lumbopelvic Muscles.

Participant ID Number: _____

Principal Investigator: Karl Christian Gibbon

Investigator contact details: Telephone: 0191 243 7018 Email: karl.gibbon@unn.ac.uk

This project is funded by: Northumbria University

INFORMATION TO POTENTIAL PARTICIPANTS

1. What is the purpose of the project?

The purpose of the study is to assess the effectiveness of a number of rehabilitative exercises in recruiting key muscles that are known to be dysfunctional in groups of people such as those with low back pain.

2. Why have I been selected to take part?

You have been asked to take part because you are a healthy individual, 18 years of age or older, with no history of muscle, skeletal or neuromuscular diseases/injuries, and abdominal or spinal surgery. You also have not experienced low back pain within the past six months (requiring medical consultation/treatment).

3. What will I have to do?

You will be asked to visit the laboratory for approximately **90 minutes** on **one** occasion. During this visit you will be asked to perform a number of tasks which are detailed below:

During the session we will be using ultrasound imaging to investigate two different muscles, one in the lower back (lumbar multifidus) and one in the abdomen (transversus abdominis), during both resting and exercising conditions. In total there are seven conditions in which we will assess both transversus abdominis lumbar multifidus. The session will contain the following exercises (although not necessarily in the order presented);

1. Rest – You will be asked to lie on a standard physiotherapy examination bench in two different positions (on your back and on your front).
2. Sitting - You will be required to sit on a stool, and will be instructed on how to achieve a neutral spinal posture.
3. Standing – You will be required to stand on the ground. Again, you will be instructed on achieving a neutral spinal posture.
4. Gym ball - You will be asked to sit on a gym ball, and will be instructed on how to achieve the appropriate spinal posture.
5. Wobble board – You will be asked to stand in an upright posture on a wobble board.
6. Exercise device in sitting – you will be asked to exercise on an elliptical-type trainer while sitting. The device differs to normal elliptical trainers in that there is no resistance to motion. You should aim to keep your upper body as still as possible during the exercise.

7. Exercise device in standing – you will be asked to exercise on an elliptical-type trainer in standing. The device differs to normal elliptical trainers in that there is no resistance to motion. You should aim to keep your upper body as still as possible during the exercise.

Total anticipated exercise duration is expected to be less than 60 minutes at a very low intensity, with individual exercise conditions lasting no more than 10 minutes.

Ultrasound imaging uses sound waves to produce an image of the tissues beneath the skin. This is a safe technique and used on a daily basis worldwide. The technology is identical to that which is used for checking on the development of foetuses within the womb during pregnancy.

4. What are the exclusion criteria (i.e. are there any reasons why I should not take part)?

- Below 18 years of age
- History of musculoskeletal or neuromuscular pathologies/injuries affecting natural walking gait
- History of abdominal or spinal surgery
- Experienced low back pain (requiring medical consultation/treatment) within the previous 6 months

5. Will my participation involve any physical discomfort?

Ultrasound transmission gel will be applied to the abdomen and lower back at numerous points throughout each testing session. This can have a cold feeling initially, however body temperature quickly counteracts this feeling. Surplus gel will also be removed when no longer needed with the use of alcohol free hypoallergenic skin wipes.

Treadmill walking and using the exercise device will cause an increase in breathing and heart rates; however the level of activity will not be anything above what is likely to take place during everyday activities.

6. Will my participation involve any psychological discomfort or embarrassment?

For the duration of this study, you will be required to expose your abdomen and lower back, and this could cause some embarrassment. Embarrassment will be reduced in the testing environment, with making it private and not overlooked.

7. Will I have to provide any bodily samples (i.e. blood, saliva)?

No.

8. How will confidentiality be assured?

You will be allocated a participant code that will always be used to identify any data that you provide. Your name or other personal details will not be associated with your data, for example the consent form that you sign will be kept separate from your data.

Only the research team will have access to any identifiable information; paper records will be stored in a locked filing cabinet and electronic information will be stored on a password-protected computer. This will be kept separate from any data and will be treated in accordance with the Data Protection Act.

9. Who will have access to the information that I provide?

Information you provide and the data we collect will be seen only by the principal investigators (Karl Gibbon, Andrew Winnard, Sarah Audsley and Eileen Baron) and their supervisors (Dr Nicholas Caplan, Dr Dorothée Debuse and Dr Angela Hibbs). All records will be kept confidential except for potential auditioning by Northumbria University Ethics Committee and regulatory authorities.

10. How will my information be stored / used in the future?

Your information will be stored on a password-protected computer or in a locked filling cabinet. All information will be used solely for the purposes of this investigation. Any personal information will be destroyed after 3 years. Data may be published in peer-reviewed journals or presented as posters/abstracts at conferences; however all data will be grouped and any personal information will not be referred to at any time.

11. Has this investigation received appropriate ethical clearance?

The study has received full ethical approval from the School of Life Sciences Ethics Committee. If you require confirmation of this please contact the chair of the Committee, stating the title of the research project and the name of the Principal Investigator:

Nick Neave
Chair of School of Psychology and Sports Sciences
Ethics Committee
Northumberland Building
Northumbria University
Newcastle Upon Tyne
NE1 8ST

12. Will I receive any financial rewards / travel expenses for taking part?

No.

13. How can I withdraw from the project?

You can withdraw your data from this project at anytime, without need for explanation or justification. To do this simply contact the principal investigator by telephone, email or in person.

14. If I require further information who should I contact and how?

Any further information required for this study can be obtain from the principle investigator:

Karl Christian Gibbon
Department of Sport & Exercise Sciences
Northumberland Building (NB431)
Northumbria University
Newcastle-upon-Tyne
NE1 8ST
E-mail: karl.gibbon@unn.ac.uk

If you would like to discuss the study, withdraw your data or register a complaint please contact the chair of the ethics committee on the address listed in section 11.

Appendix M - Blank Consent Form (Chapter V)

INFORMED CONSENT FORM

Project Title: Evaluation of the Effectiveness of a Range of Rehabilitative Exercises in Recruiting Local Lumbopelvic Muscles.

Principal Investigator: Karl Christian Gibbon

Participant Number: _____

	please tick where applicable
I have read and understood the Participant Information Sheet.	☐
I have had an opportunity to ask questions and discuss this study and I have received satisfactory answers.	☐
I understand I am free to withdraw from the study at any time, without having to give a reason for withdrawing, and without prejudice.	☐
I agree to take part in this study.	☐
I would like to receive feedback on the overall results of the study at the email address given below. I understand that I will not receive individual feedback on my own performance. Email address……	☐

Signature of participant..... Date.....……..

(NAME IN BLOCK LETTERS).....…….

Signature of Parent / Guardian in the case of a minor

.....

Signature of researcher..... Date.....……..

(NAME IN BLOCK LETTERS).....…….

Appendix N - Summary Raw Data Table for Lumbar Multifidus Ultrasound Thickness Measurements (Chapter V)

	Rest			Sitting			Standing			Gym Ball			Wobble Board			FRED Sitting			FRED Standing		
Participant #	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
001	37.7	29.9	31.7	33.5	33.1	33.3	38.6	38.8	38.8	35.4	34.8	35.1	45.3	39.3	35.8	32.7	34.0	33.4	36.5	37.3	36.9
002	34.4	36.2	35.3	37.9	37.1	37.4	41.4	41.1	42.0	42.0	41.0	41.4	42.0	43.1	42.5	39.4	39.4	39.4	43.4	43.7	44.7
003	33.1	32.8	32.1	37.7	32.4	33.5	42.9	42.9	44.7	39.3	39.4	40.0	43.6	44.3	43.6	38.1	37.0	38.5	39.6	39.6	43.0
004	32.9	33.8	33.0	33.2	34.2	35.1	36.7	37.8	36.7	34.8	34.8	34.2	35.5	36.5	37.1	31.8	31.7	30.6	45.3	40.0	44.3
005	33.8	33.4	33.3	35.1	35.1	34.6	41.2	41.5	41.3	37.1	37.4	37.4	42.5	41.7	41.3	36.8	37.1	37.3	41.5	41.7	43.0
006	26.6	25.7	26.4	26.3	26.9	27.1	27.1	27.1	27.5	34.0	33.9	32.9	32.5	31.9	31.5	28.7	28.1	29.5	29.8	30.5	30.8
007	36.7	36.8	37.0	36.5	37.0	37.2	42.4	43.3	41.5	38.3	39.8	38.7	44.2	44.0	43.0	39.2	39.0	38.9	41.6	41.5	40.1
008	31.5	31.7	32.0	33.1	33.6	32.8	44.5	42.4	43.0	34.8	35.5	33.1	44.5	43.5	44.1	34.1	35.8	36.5	42.2	41.6	41.3
009	32.6	33.1	33.3	32.3	34.2	34.2	44.4	44.9	44.6	34.4	34.7	34.8	43.5	41.5	41.6	39.1	39.0	37.4	42.1	43.2	44.9
010	37.9	36.6	36.5	41.1	40.0	41.0	44.5	48.0	44.6	42.2	42.7	43.8	47.0	47.4	47.1	43.9	46.4	47.0	48.0	49.8	49.2
011	37.8	39.6	37.5	41.5	44.1	36.5	47.3	47.6	47.3	37.4	38.7	38.6	46.3	44.4	44.6	44.0	44.2	43.8	45.6	47.5	48.5
012	30.5	31.5	33.5	33.1	33.1	33.0	43.4	42.7	43.5	35.5	36.2	36.0	43.0	42.8	43.7	33.3	33.1	33.6	42.4	43.5	43.5

Appendix O – Summary Raw Data Table for Transversus Abdominis, Internal Oblique and External Oblique Ultrasound Thickness Measurements (Chapter V)

	A - Rest									***B - Sitting***									***C - Standing***								
	TrA			***IO***			***EO***			***TrA***			***IO***			***EO***			***TrA***			***IO***			***EO***		
Participant #	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
001	5.3	4.4	4.5	13.9	13.4	13.1	6.2	5.6	5.5	4.9	4.6	4.9	15.4	15.6	14.7	7.9	6.5	6.4	5.4	5.0	5.3	13.5	13.0	13.4	6.0	5.6	5.3
002	4.8	4.3	4.0	11.1	9.9	10.5	6.0	7.4	7.8	5.7	5.5	4.9	11.7	10.0	10.6	6.0	6.0	6.5	7.1	5.0	6.1	11.3	10.5	10.9	7.0	6.0	6.6
003	5.3	5.2	4.6	12.8	12.6	11.8	7.5	7.1	6.6	7.1	6.8	5.9	13.1	12.8	12.1	8.6	9.6	8.2	6.0	5.5	6.8	13.3	13.5	14.5	8.1	8.5	8.5
004	5.7	5.6	6.2	14.4	14.8	16.7	7.5	7.4	7.2	7.9	7.7	8.3	16.2	17.6	18.6	8.3	9.1	9.7	7.6	6.7	7.8	15.1	14.0	17.1	6.3	7.3	7.6
005	5.0	4.8	4.5	11.7	11.5	11.8	7.0	7.1	7.3	6.5	7.0	6.4	13.7	13.9	13.6	7.9	8.1	8.1	5.7	5.6	5.8	13.0	12.7	13.0	7.5	7.4	6.8
006	4.3	4.3	2.4	8.1	6.7	8.6	6.0	7.3	8.0	5.2	5.2	5.4	10.3	9.2	9.8	7.4	6.9	6.8	4.8	5.0	5.0	8.8	9.3	9.6	9.6	9.9	1.5
007	6.6	6.6	6.6	12.5	12.2	12.3	8.7	8.8	10.0	7.4	7.9	6.9	13.7	13.3	13.0	7.8	8.0	8.6	6.8	6.6	6.4	15.7	15.9	16.8	11.8	12.0	12.1
008	4.1	3.4	3.2	12.9	13.7	13.3	8.6	7.5	8.8	5.0	5.0	5.3	12.2	14.1	13.6	10.2	10.8	9.7	5.1	5.5	5.4	12.6	12.7	13.2	6.0	6.1	6.1
009	3.3	3.9	3.7	10.9	10.9	11.1	6.0	5.4	5.3	3.5	3.4	3.6	12.8	12.2	13.0	6.2	6.0	6.2	4.5	4.6	4.3	12.5	12.5	10.9	6.1	6.3	5.9
010	5.7	5.2	4.7	8.0	8.9	9.2	6.6	7.0	7.0	6.1	5.5	5.4	8.5	7.3	7.2	7.4	7.2	8.4	5.7	6.2	5.5	10.3	8.9	9.6	6.9	5.9	6.4
011	3.9	3.6	3.1	12.5	12.5	13.4	7.0	7.0	7.3	3.9	4.6	4.1	22.1	24.1	21.7	9.1	10.7	9.9	6.7	6.0	6.0	17.8	16.7	14.5	8.4	7.2	7.8
012	4.2	3.8	4.1	11.8	11.0	10.1	6.8	7.3	7.0	4.7	5.2	6.1	14.3	16.8	15.4	7.8	8.8	8.6	5.1	5.0	5.2	11.9	12.6	12.6	6.3	7.1	7.0

D - Gym Ball									*E - Wobble Board*									*F - FRED Sitting*								
TrA			***IO***			***EO***			***TrA***			***IO***			***EO***			***TrA***			***IO***			***EO***		
1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
4.7	5.2	4.2	16.3	17.3	15.3	5.1	4.9	5.6	5.7	5.2	4.1	14.9	14.8	13.8	4.7	6.0	5.5	6.4	5.4	4.9	10.9	13.1	12.1	6.3	6.8	6.4
5.4	5.1	5.5	11.7	11.9	11.4	8.2	6.3	7.3	6.5	7.0	6.7	13.9	14.5	14.4	5.7	5.7	5.1	6.1	5.7	5.1	11.1	10.7	12.0	5.5	5.1	6.1
6.1	7.4	6.4	10.2	12.1	13.5	7.1	8.0	9.1	5.7	6.8	5.8	14.7	14.1	14.9	7.9	8.5	7.6	6.1	6.8	6.2	13.6	13.5	18.0	5.8	6.5	7.1
6.3	6.1	7.6	17.5	18.5	19.4	8.2	8.3	9.0	6.1	6.0	7.7	16.0	14.7	15.7	7.2	7.4	6.7	7.5	7.0	7.9	18.5	17.1	16.8	12.2	11.5	10.3
6.5	6.4	6.5	14.0	14.3	14.2	7.7	7.6	8.0	6.0	6.3	5.6	13.8	13.9	14.2	6.4	6.9	7.0	5.9	5.8	6.1	15.0	14.3	15.3	7.5	7.3	7.5
5.2	5.8	5.7	9.3	10.0	10.1	5.3	5.6	5.7	5.3	4.6	5.2	10.4	10.4	10.5	5.0	5.4	5.9	5.9	6.1	6.8	8.2	8.0	9.3	6.3	5.7	5.6
7.7	7.1	8.2	14.7	14.9	13.9	8.2	9.6	8.2	11.9	10.5	6.5	16.8	18.5	20.2	8.0	7.9	11.4	11.7	11.8	11.7	19.3	18.1	19.9	9.5	10.1	10.5
5.9	5.5	6.1	14.7	13.9	14.9	11.1	10.5	11.7	5.0	5.6	5.4	12.0	13.4	13.0	6.0	6.8	7.0	5.7	6.2	6.4	21.0	14.0	15.5	10.2	7.6	7.4
4.1	4.4	4.9	13.1	12.4	12.4	6.9	6.5	6.9	4.0	4.7	4.4	11.8	11.7	12.6	5.6	6.3	6.6	5.5	4.9	5.7	11.7	11.5	12.8	5.4	4.9	5.2
7.4	6.7	8.0	11.6	10.4	11.2	9.1	9.4	9.0	6.5	7.5	5.9	11.9	12.8	12.0	7.1	7.2	6.7	6.7	5.8	6.3	10.8	6.9	10.0	5.5	6.0	7.0
4.9	4.9	4.5	20.6	20.7	20.7	8.0	7.2	8.9	6.1	6.4	6.6	18.1	16.5	16.7	7.4	8.0	7.9	6.1	5.5	5.2	21.2	22.8	20.3	8.5	8.6	8.3
6.1	4.8	5.6	14.3	14.9	13.5	7.4	7.5	6.9	4.9	5.9	5.1	11.2	12.0	12.2	5.6	6.5	6.8	6.8	6.8	7.6	18.7	21.1	21.2	7.0	7.8	8.2

G - FRED Standing								
TrA			***IO***			***EO***		
1	2	3	1	2	3	1	2	3
7.2	5.7	5.9	14.6	14.7	15.6	5.1	5.5	5.3
6.1	6.4	6.3	12.4	10.8	11.2	4.5	4.6	4.9
5.6	5.0	6.4	17.0	16.1	15.4	7.4	7.7	6.0
8.7	8.9	8.6	15.5	16.8	16.3	7.0	7.1	7.5
6.2	6.2	6.8	14.7	14.9	15.0	6.7	6.9	6.9
6.8	6.9	6.7	9.2	7.5	8.5	6.6	6.2	5.9
12.1	11.4	12.5	15.7	18.9	17.4	8.8	10.3	10.7
6.4	5.6	7.1	17.8	18.9	18.5	9.0	9.0	7.9
7.2	7.4	7.9	15.1	16.2	14.2	6.2	6.3	6.1
6.3	7.0	6.9	12.5	11.8	12.4	5.9	5.8	6.4
7.1	6.4	6.9	14.9	14.1	16.6	5.8	5.8	5.2
6.5	6.9	7.1	16.8	18.4	15.0	6.9	7.8	8.3

Appendix P – Participant Information Sheet (Chapters VI and VII)

PARTICIPANT INFORMATION

Project Title: Kinematic, Kinetic, and Electromyographical Comparison of Overground Walking and Exercise Using a Newly Developed Exercise Device.

Participant ID Number: _____

Principal Investigator: Karl Christian Gibbon

Investigator contact details: Telephone: 0191 243 7018 Email: karl.gibbon@unn.ac.uk

This project is funded by: Northumbria University

INFORMATION TO POTENTIAL PARTICIPANTS

1. What is the purpose of the project?

The purpose of this study is to compare the movement and muscular activation characteristics of a person using a newly developed exercise device with those observed during natural overground walking.

This comparison is being drawn in an attempt to quantify the functionality of the newly developed device.

2. Why have I been selected to take part?

You have been asked to take part because you are a healthy individual, 18 years of age or older, with no history of muscle, skeletal or neuromuscular diseases/injuries, and abdominal or spinal surgery. You also have not experienced low back pain within the past six months (requiring medical consultation/treatment) and are not currently pregnant.

3. What will I have to do?

You will be asked to visit the laboratory for approximately ***one and a half hours*** on a ***single*** occasion.

During this visit we will be using 3-dimensional motion capture, integrated with force and muscular activation analysis equipment to capture movement data during overground walking and whilst using a prototype elliptical exercise device.

To accurately capture the movements of your body during walking and exercise we will attach a temporary marker set (individual lightweight reflective spheres) to 41 prominent anatomical landmarks on your upper and lower extremities, torso, and head. Due to the location of several of these markers you will be required to wear only shorts (with the exception of female participants who can also wear a sports/running vest). Changing rooms are available if required.

Various anthropometrical measurements will also be taken including height, mass, leg lengths, ankle widths, knees widths, wrist widths, elbow widths, and hand thickness.

Muscle activity in 14 muscles of interest will be recorded simultaneously alongside movement data using a process called surface electromyography (sEMG). This involves

the attachment of hypoallergenic self-adhesive pads and lightweight telemetric transmitter units to the surface of the skin. To attain a true reflection of underlying muscular activation the outermost layers of skin need to be free from body hair, dead skin cells, and any bodily secretions such as perspiration and residue. Therefore prior to EMG attachment, small areas of skin may require shaving, exfoliating and cleansing.

Walking trials will consist of a walking on a flat 10m walkway with your natural gait style and speed. Both motion data and muscle activity data will be captured simultaneously. This will be repeated until we have three complete trials where both your left foot and right foot make direct contact with a force measurement device. This device is embedded in the floor, flush with the floors surface and will not pose a trip hazard.

4. What are the exclusion criteria (i.e. are there any reasons why I should not take part)?

- Below 18 years of age
- History of musculoskeletal or neuromuscular pathologies/injuries affecting natural walking gait
- History of abdominal or spinal surgery
- Experienced low back pain (requiring medical consultation/treatment) within the previous 6 months
- Current pregnancy
- History of allergic reactions to tape adhesive and/or EMG electrode gel

5. Will my participation involve any physical discomfort?

Removal of the reflective markers and EMG electrode pads can cause mild discomfort caused by the adhesive pulling on the skin and any body hair.

Although all tape adhesives and electrode conductivity gels are hypoallergenic the small possibility of minor skin reactions remains.

EMG preparation will involve the removal of some body hair (wet shave removal) where required, skin exfoliation and alcohol cleansing. This procedure can cause mild skin soreness; however, this sensation will be short lasting (approximately 20-30 seconds).

6. Will my participation involve any psychological discomfort or embarrassment?

Male participants will be required to wear only a pair of shorts (preferably above knee length). Female participants are required to wear shorts and a tight-fitting vest, sports bra or equivalent. This may cause you embarrassment if you are conscious of your body in any way, however the testing environment is completely private with opaque blinds and curtains and a lockable door.

One area assessed for muscular activation is the gluteus maximus, which may cause embarrassment for some participants. Every effort will be made to make the process as efficient as possible. A female can also be present if requested to assist with preparation and placement.

7. Will I have to provide any bodily samples (i.e. blood, saliva)?

No.

8. How will confidentiality be assured?

You will be allocated a numeric participant code and all referrals to data pertaining to yourself shall be noted with that code. All data will be stored within a locked filing cabinet or on a password protected desktop computer in accordance with the Data Protection Act (1998).

9. Who will have access to the information that I provide?

Information you provide and the data we collect will be seen only by the principal investigators (Karl Gibbon & Sam Brennan) and their supervisors (Dr Nicholas Caplan & Dr Dorothée Debuse). All records will be kept confidential except for review by Northumbria University Ethics Committee and regulatory authorities.

10. How will my information be stored / used in the future?

Your information will be stored on a password-protected computer or in a locked filling cabinet. All information will be used solely for the purposes of this investigation. Any personal information will be destroyed after 3 years. Data may be published in peer-reviewed journals or presented as posters/abstracts at conferences; however all data will be grouped and any personal information will not be referred to at any time.

11. Has this investigation received appropriate ethical clearance?

The study has received full ethical approval from the School of Life Sciences Ethics Committee. If you require confirmation of this please contact the chair of the Committee, stating the title of the research project and the name of the Principal Investigator:

Nick Neave
Chair of School of Psychology and Sports Sciences
Ethics Committee
Northumberland Building
Northumbria University
Newcastle Upon Tyne
NE1 8ST

12. Will I receive any financial rewards / travel expenses for taking part?

No.

13. How can I withdraw from the project?

You can withdraw your data from this project at anytime, without need for explanation or justification. To do this simply contact the principal investigator by telephone, email or in person.

14. If I require further information who should I contact and how?

Any further information required for this study can be obtain from the principle investigator:

Karl Christian Gibbon
Department of Sport & Exercise Sciences
Northumberland Building (NB431)
Northumbria University
Newcastle-upon-Tyne
NE1 8ST
E-mail: karl.gibbon@unn.ac.uk

If you would like to discuss the study, withdraw your data or register a complaint please contact the chair of the ethics committee on the address listed in section 11.

Appendix Q – Blank Consent Form (Chapters VI and VII)

INFORMED CONSENT FORM

Project Title: Kinematic, Kinetic, and Electromyographical Comparison of Overground Walking and Exercise Using a Newly Developed Exercise Device.

Principal Investigator: Karl Christian Gibbon

Participant Number: _____

	please tick where applicable
I have read and understood the Participant Information Sheet.	☐
I have had an opportunity to ask questions and discuss this study and I have received satisfactory answers.	☐
I understand I am free to withdraw from the study at any time, without having to give a reason for withdrawing, and without prejudice.	☐
I agree to take part in this study.	☐
I would like to receive feedback on the overall results of the study at the email address given below. I understand that I will not receive individual feedback on my own performance. Email address……	☐

Signature of participant..... Date.....……. (NAME IN BLOCK LETTERS).....…….
Signature of Parent / Guardian in the case of a minor .....

Signature of researcher..... Date.....……. (NAME IN BLOCK LETTERS).....…….