TRUNK MUSCLE ACTIVATION PATTERNS DURING THE BARBELL BACK SQUAT WITH PROGRESSIVE LOAD AND INSTABILITY

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Ian Stewart Rowan-Legg, candidate for the degree of Master of Science in Kinesiology & Health Studies, has presented a thesis titled, *Trunk Muscle Activation Patterns During the Barbell Back Squat with Progressive Load and Instability*, in an oral examination held on December 9, 2019. The following committee members have found the thesis acceptable in form and content, and that the candidate demonstrated satisfactory knowledge of the subject material.

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Abstract

Low back injuries are common in sports that involve repetitive/high spinal loading. The muscular system is largely responsible for maintaining proper spinal posture and efficiently transferring load during movement to protect the spine from excessive strain. Atrophy/reduced recruitment of specific trunk muscles are associated with low back pain, suggesting changes in recruitment patterns and compensations that may lead to additional stress on spinal structures. Athletes involved in high loading sports (e.g., football) engage in strength training programs that typically involve high load compound movements to increase strength (e.g., back squat) and stability training. It may prove effective to increase the loading specificity of these movements by performing stability work under moderate loads, to increase total trunk muscle activity and induce more sport-specific adaptations. The purpose of this study was to examine the effect of progressive load and support surface stability on trunk muscle activity and lumbar spine ranges of motion during the back squat. Twenty-six participants recruited from the University of Regina Rams football team performed a series of progressive load (30/50/70%) back squats on a stable and unstable surface. Peak and average activity of the superficial trunk muscles were measured using surface electromyography. Lumbar ranges of motion were measured using motion capture. Results show that surface instability had no significant effects on trunk muscle activity or lumbar ranges of motion. Conversely, increases in load were associated with increased trunk muscle activity, with some differences being specific to the phase of movement. Increases in load were also associated with decreased upper lumbar flexion and increased lower lumbar extension. Findings suggest that the level of support surface instability required to elicit significant increases in trunk muscle activity needs to be greater than what was used in the present study. However, the potential performance benefit from incorporating further instability is dubious.
since the regular back squat would seem to be an effective exercise to train trunk muscle activation and stability.

*Keywords: Spine, Trunk, Strength Training, Stability Training, Football*
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List of Abbreviations

1RM – 1-repetition maximum
ANOVA – Analysis of variance
AF – Annulus fibrosis
AHM – Abdominal hollowing maneuver
ASC – Ascending phase of the squat
ASIS – Anterior superior iliac spine
CEP – Cartilaginous endplate
COM – Center of mass
DSC – Descending phase of the squat
EMG – Electromyography
EMGpeak – Peak EMG amplitude
EMGavg – Average EMG amplitude
EO – External oblique
ES – Erector spinae
GM – Gluteus maximus
GMd – Gluteus medius
IAP – Intra-abdominal pressure
IAR – Instantaneous axis of rotation
IO – Internal oblique
IVD – Intervertebral disc
LBP – Low back pain
MFD – Multifidus
MIVC – Maximal isometric voluntary contraction
MVC – Maximal voluntary contraction
NALBP – Non-acute low back pain
NP – Nucleus pulposis
PS – Paraspinal muscles
PSIS – Posterior superior iliac spine
QL – Quadratus lumborum
RA – Rectus abdominis
RMS – Root mean square
STA – Soft tissue artifact
TF – Technical frame
TrA – Transversus abdominis
1. Introduction

Low back pain (LBP) and injury are common occurrences in sports that involve high or repetitive loading (i.e., compression force, flexion-extension cycles) (Baranto et al., 2009), with one study reporting 90% of athletes involved in high loading sports or training showing accelerated signs of intervertebral disc (IVD) degeneration and 78% reporting previous or current LBP (Baranto et al., 2009). Stress can be further increased with movements that combine high loading with repetitive flexion-extension cycles (Foss et al., 2012).

Acute back injuries can be identified by overload and failure of a given tissue that attaches to the spine, while non-acute LBP (NALBP) typically evolves from repetitive strain on spinal tissues that could be associated with poor posture or faulty motor control patterns (Panjabi, 1992). In individuals with LBP, improper neuromuscular recruitment and atrophy of deep trunk muscles such as the transversus abdominis (TrA) and multifidus (MFD) have been reported (Hides et al., 1994; Hodges, 2001; Hodges & Richardson, 1996; Walsh et al., 2007). Appropriate rest must be taken following low back injury in order to allow tissues to recover during the initial stages of healing and limit additional damage, with appropriate rehabilitation following to allow for re-strengthening of the associated musculature and training of proper motor patterns. If the body continues to compensate by loading certain tissues in excess or engaging in faulty movement patterns, the risk of developing NALBP or experiencing another acute injury is potentially increased. This suggests that compensatory and potentially detrimental activation patterns may allow for transient stability, but lead to long term damage of spinal structures or musculature.

While sometimes viewed as compensatory, the activation of superficial trunk musculature is also important in maintaining spinal stability in athletic settings. Trunk
musculature can be separated into local and global functional groups, where the local group contributes to sagittal and lateral stiffness to maintain mechanical stability at the spinal level, while the global muscles balance and transfer external load between the trunk and extremities (Bergmark, 1989). Muscles such as the TrA and MFD, being part of the local functional group, are recognized for their importance in maintaining spinal stability from a postural standpoint during daily activity (Haff & Triplett, 2015; Ward et al., 2009). When attempting to reinforce trunk stiffness during exercise and athletic movements, a common technique used is the abdominal bracing technique. This involves the conscious contraction of superficial trunk muscles surrounding the spine (anteriorly, laterally, and dorsally) to increase pressure within the abdominal cavity. While this has been shown to slightly increase spinal compression, the increase in intra-abdominal pressure (IAP) combined with increasing muscular activity of the internal oblique (60% maximum voluntary contraction (MVC)), rectus abdominus (18% MVC), external oblique (27% MVC), and erector spinae (19% MVC) increases spinal stability (Sticklet et al., 2015). Similarly, Grenier and McGill (2007) noted a 32% increase in spinal stability and 15% increase in spinal compression during abdominal bracing. This suggests that activation patterns of trunk muscles are influenced by voluntary stabilizing techniques. Through these findings, it could be further implied that the magnitude of the load and subsequent lumbopelvic stability requirements increase global muscle activity during resistance training, as abdominal bracing is very often used during moderate to high-load lifting.

Football players experience repetitive bouts of high impact hits in unstable conditions, with some positions experiencing spinal compression forces of up to 10 000N (Gatt et al., 1997). Football training must include moderate-high load and high power multi-joint exercises as well as stability training in order to elicit adaptations that prepare players for the conditions of their
sport. A common exercise used in football strength training is the back squat, as it has been shown to be effective at increasing maximal strength in the lower limb and trunk, improving distance sprint times, and increasing jump height (McBride et al., 2009; Wisløff et al., 2004). Unstable variations of the back squat under little to no load have been observed to increase activation of the trunk musculature, though reduced muscular activity and force output is observed in the lower limbs relative to stable conditions (Saeterbakken et al., 2013; Behm et al., 2005; Lawrence et al., 2015). While this occurs as a response to the increased stability requirements, most notably in the lumbopelvic complex, it is also recognized that the back squat has its own significant spinal stability and trunk strength requirements. Furthermore, trunk muscle activity may change during certain portions of the movement, depending on specific torso, hip, and knee joint angles (Garcia-Vaquero et al., 2012; Caterisano et al., 2002; Dionisio et al., 2008). While increases in lumbar extension have been noted with increasing loads during the back squat (Adams et al., 2000; Walsh et al., 2007), spinal motion during this exercise has not been examined in unstable conditions.

In order to maximize sport specific adaptation, training must progress to a point that closely mimics sport specific movements. While football players often train their lower limbs with moderate to heavy loads, stability work is often performed under light or no load to minimize risk of injury, even though these two attributes of the sport (high load, high instability) are often experienced simultaneously. Minimal research has examined the effect of increasing load (moderate) on trunk stabilizer muscle activity during an unstable back squat (Anderson et al., 2005; Behm et al., 2010), with none being performed in a trained athletic population (Willardson et al., 2009; Clark et al., 2017). It could be proposed that as load increases, trunk muscle activation increases to a greater extent to both withstand the spinal loading associated
with the movement and stabilize the lumbopelvic complex through the unstable surface condition. Activity of these muscles may further increase during portions of the movement where lumbopelvic stability and trunk musculature strength requirements are greater, specifically when reaching end range hip flexion. By having athletes perform unstable squats under various loads, a balance may be found that both promotes sport-specific adaptations and activation of trunk musculature, while still offering sufficient load to the lower limbs to improve strength. This could subsequently reduce the risk of sport-related back injuries, while also challenging athletes without the need for high loads.

Surface electromyography (EMG) is an effective method for measuring neuromuscular activity of superficial trunk musculature (Ng et al., 2002; Vera-Garcia et al., 2010; Roth et al., 2016). This is assuming electrode placement is appropriate to record neuromuscular activity and reduce crosstalk, and appropriate settings are applied to filter additional noise (De Luca, 1997; Miathiassen et al., 1995; Ekstrom et al., 2005). Motion capture technology can be used to analyze relative segment motion and determine joint kinematics by placing markers on the skin over various anatomical landmarks and tracking their movement (Cappozzo et al., 2005). This can be synchronized with EMG to examine muscular activity relative to joint movement (Berme et al., 1990; Ekstrom et al., 2005), giving us greater insight into human movement.

The purpose of this study was to examine the effect of progressive load and support surface stability on trunk muscle activity and lumbar spine ranges of motion during the back squat. It was hypothesized that an increase in trunk muscle activity would occur as load increases and when squatting on an unstable surface during all loads. It was also hypothesized that an increase in peak lumbar extension angle would occur as load increases as this is a compensation observed in moderate to high loads during the back squat, and that an increase in peak trunk
lateral flexion angle on an unstable surface would occur due to the introduction of frontal plane instability.
2. Literature Review

2.1. Back Pain and Injury in Sport

Acute back injury may be characterized by overload and potentially failure of a given spinal tissue. This may include damage to the muscle tissue, ligaments, IVD’s, vertebrae, and even the nerves and spinal cord. If the trunk musculature does not sufficiently stabilize the spine during sudden or high loading, it leaves the passive tissues to absorb a portion of the load and increases risk of injury (Maduri et al., 2008). While small moments exerted on the spine keep the instantaneous axis of rotation (IAR) within the vertebral body, increases in axial loading and moment magnitude shift this axis posteriorly towards the facet joints, causing additional facet stress during certain movements, most notably axial rotation (Schmidt et al., 2008). This also suggests that the facets play an important structural role in spinal stability. When removing all posterior (to the vertebral body) bony structures and ligaments in vitro, IVD strain increased 2.5 fold with axial loading in a neutral position, and nearly 4 fold during combination movements (Heuer et al., 2008).

Risk of damage to the IVD is associated with both the magnitude of spinal loading and the number of loading cycles performed in bovine tissues (Iatridis et al., 2005). It has been shown that the elastin content of the annulus fibrosis (AF) plays a mechanical role in enhancing the integrity of the AF fiber bundles during transverse shear (Michalek et al., 2009). The presence of changes in this elastic fiber network of herniated and degenerated IVDs suggests increased risk of micro failure during motions that generate high tissue shear (Michalek et al., 2009). This also implies that different motions may have differing negative effects on the AF given its anisotropic structure.
While IVD degeneration is a normal aspect of aging, this can be accelerated through exposure to repetitive compression and flexion moments, as flexing the lumbar spine has been shown to increase compressive force on the IVD (Adams et al., 2000) and increase IVD strain by 266% \textit{in vitro} (Schmidt et al., 2008). While the nucleus pulposus (NP) and AF of the IVD can withstand high compression and shear force, flexion movements have a tendency to “push” the IVD posteriolaterally towards the nerve roots and spinal cord, increasing the risk of nerve compression and subsequent neurological damage (Goel et al., 1995). While lumbar extension has been shown to increase spinal compression as well, the IVD does not have the same anterior movement capability as it does posteriorly, given the placement of the anterior longitudinal ligament. However, hyperextension of the lumbar spine does expose the vertebra to risk of localized damage to the vertebral processes or facet joints (McGill, 2015).

It has been observed that NALBP is associated with altered neuromuscular recruitment and coordination (e.g., onset times) of various trunk muscles that are responsible for maintaining mechanical stability of the spine during movement (Hodges, 2001). If these protective muscles do not activate sufficiently or in a coordinated fashion as spinal loading increases, the resultant shear, compressive, or tensile forces associated with the given movement are placed on the passive structures of the spine rather than the surrounding musculature (Panjabi, 1992). It has also been reported that NALBP is associated with delayed activation and poor endurance of the hip extensor and abductor muscles (e.g., gluteus maximus, gluteus medius) and poor balance, highlighting the hip musculature’s contribution to stabilizing the pelvis and transferring forces between the lower limbs and trunk (Hart et al., 2007; Kankaanpää et al., 1998).

One study found that 90% of athletes involved in high loading sports (i.e., compression force, flexion-extension cycles) or training show signs of IVD degeneration, with 88% of these
athletes also experiencing accelerated long term IVD deterioration (Baranto et al., 2009). Of all athletes observed, 78% had reported having previous or current LBP, with the highest frequency of visible IVD damage and LBP being observed in those athletes involved in higher loading sports (e.g., weightlifting, ice hockey) (Baranto et al., 2009). While the control group (non-athletes) only showed a 38% lifetime prevalence of LBP at the initial assessment (vs. 78% in athletes), LBP was almost identical after a 15-year follow up (71% vs. 75% in control vs. athletes, respectively) (Baranto et al, 2009). This may suggest that chronic repetitive stress or trauma leads to the development of NALBP or accelerates its onset. In a 10-year cohort study by Foss et al. (2012), athletes of various sports reported point prevalence of LBP within a 12-month span. When compared to non-athletes, a significantly higher point prevalence of LBP was observed in athletes participating in sports involving high spinal loading, high volumes of flexion/extension cycles, or both (e.g., wrestling, rowing, hockey). This further supports the notion that increased spinal damage and accelerated IVD degeneration are consequences of high loading sports, especially when end range movements are involved.

2.2. The Lumbar Spine

2.2.1. Spinal Stability

The spinal column (i.e., “spine”) is a complex structure and is responsible for protecting some of the most important tissues and organs in the entire human body. While being an insertion point for several muscle groups, the spine houses various blood vessels as well as the spinal cord and branching nerve roots, which supply sensory and motor innervation to the body (Axler et al., 1997). The lumbar spine, in combination with the pelvis, contributes to trunk
stability and acts as a medium for force transfer between the trunk and lower extremities (Izzo et al., 2013).

The spine is susceptible to injury in the sense that it is exposed to almost constant compression or shear forces, and can withstand minimal loading in the absence of supporting skeletal muscle (McGill, 2015). The various surrounding muscles of the spine are what allow for proper movement patterns and stable positioning (McCook et al., 2007). However, when these particular muscles are injured or weakened, it can promote poor posture and compromise stability.

Joint stability is defined as the “ability to maintain or control joint movement or position, with stability being achieved by coordinating actions of surrounding tissues and the neuromuscular system” (Panjabi, 1992). Spinal stability has been defined as the spine’s proficiency under various loads to limit movement as a means of preventing damage or irritation of the spine or any of the structures it protects (White et al., 1975). There is a close relationship between the bones and ligaments (passive subsystem) and surrounding musculature (active subsystem) of the spine, as well as the nervous system (control unit), and interruption in any of these systems may lead to excessive stress on or dysfunction of the remaining systems (Panjabi, 1992). While the joint capsules and ligaments provide direct joint support, they also contain mechanoreceptors that send a constant stream of proprioceptive information to the brain and spinal cord regarding load and movement, along with the muscle spindles and golgi tendon organs. This information is received by the central nervous system and then used to organize and produce motor responses (Panjabi, 1992). Silfies et al. (2015) proposes that trunk (or core) stability is a dynamic process requiring optimal muscle strength, endurance, and power, as well as neuromuscular control (accurate joint and muscle receptors and neural pathways) that can
quickly integrate sensory information and alter motor responses relative to internal and external information.

2.2.2. Anatomy and Biomechanics

The spinal column is made up of vertebrae, with adjacent vertebrae being separated by an IVD that collectively allow the vertebral column to bend or twist while distributing compressive load to the adjacent vertebral bodies (Adams et al., 2006). The IVDs are mostly avascular and have three main components: the NP, AF, and a cartilaginous endplate (CEP) which bonds the IVD to the vertebral endplate (Newell et al., 2017). The NP is a gelatinous structure that accounts for 40-50% of the IVD volume, and is composed primarily of proteoglycan, collagen, and water (Antoniou et al., 1996; Bayliss et al., 1992; Farfan et al., 1970). Due to its high water content, the NP displays high hydrostatic pressure in response to compressive loading (McNally et al., 1992). The AF is comprised of 15-25 layers (lamellae) of collagen fibre bundles that are bridged together by a network of collagen in order to provide shear resistance between lamellae (Derby et al., 2015; Marchand et al., 1990). The inner AF is composed mainly of type II collagen, while the outer AF is composed mainly of type I collagen. The mechanical properties of these collagen types allow for structural integrity of the AF, where the type I collagen provides tensile strength while the type II collagen forms a network that binds proteoglycans and water in order to withstand high compressive force (Schollmeier et al., 2000). Elastin within the AF is situated between adjacent lamellae to improve recoil following deformation (Yu et al., 2007). Due to the viscoelastic properties of the IVD, the quicker the IVD is compressed, the stiffer it becomes in response (Costi et al., 2008).
The lumbar spine consists of five vertebrae which are larger and stronger than other moveable vertebrae, and increase in size from L1-L5. Each vertebra is composed of a vertebral body, a neural arch which protects the spinal cord, and bony processes (2 transverse, 1 spinous) that increase specific muscle action efficiency. Four articular processes (two superior and two inferior) lie between the transverse and spinous processes. Lumbar vertebral bodies can be most easily distinguished by their lack of rib facets and general shape compared to other vertebrae (Marieb et al., 2007).

Given the function of the lumbar spine in maintaining trunk stability, its range of motion (ROM) is less than that of the thoracic spine (Geelhoed et al., 2006). Flexion-extension, lateral flexion, and rotational capability of the lumbar spine typically ranges from 12-20°, 24°, and 2°, respectively (Nordin et al., 2001). Axial rotation is much less than other movements because of the interlocking articular processes. In a study by Xia et al. (2010), participants moved into different postural positions to examine lumbar vertebral motion in the sagittal and transverse planes at the L2/3 and L3/4 segments using combined dual fluoroscopic and magnetic resonance imaging (MRI) techniques. ROM and IAR of each vertebral segment were analyzed during flexion-extension and left-right twisting motions. For sagittal plane movements (i.e., flexion-extension), the anterior portion of the vertebrae had a smaller ROM than the posterior portion and the vertebral IAR was located at the posterior third of the vertebral body. For transverse plane movements (i.e., axial rotation), the vertebrae rotated with respect to points approximately 30mm in front of the vertebrae, indicating different IARs during different planar movements. Using dual fluoroscopic imaging and MRI images as reference, Liu et al. (2016) examined the ROM and IAR of intervertebral segments of the lower lumbar spine (L4/L5, L5/S1) during sagittal plane lifting (extension). Average flexion angles were 6.6° and 5.3°, with extension
angles reaching 1.8° and 3.5°, for L4/L5 and L5/S1, respectively. Movement direction was shown to influence IAR location, with the L4/L5 IAR located 75% posterior to the anterior edge of the IVD during flexion, and moving an additional 17% posterior during extension. Conversely, the L5/S1 IAR was located 95% posterior to the anterior edge of the IVD during flexion, and moved 12% outside the posterior edge of the IVD during extension. This provides evidence that extension moves the IAR posterior, and may vary based on how much vertebral movement is present during dynamic lifting (Wattananon et al., 2017). It has been suggested that this posterior shift may cause increased load on the facet joints (Schmidt et al., 2008). Notably, individuals with LBP have been observed to experience less variability in IAR with sagittal movements (Wattananon et al., 2017), suggesting the presence of compensations to stiffen the lumbar spine and reduce LBP during functional tasks.

2.2.3. Musculature

Various trunk and hip muscles are involved in maintaining spinal and lumbopelvic stability with some having synergistic interactions during specific stability techniques. Some of these muscles are outlined below, along with research supporting their importance for spinal stability and strength training exercises. A summary of each muscle’s origin, insertion, and action can be seen in Table 1.

2.2.3.1. Transversus Abdominis (TrA)

The TrA’s main functions are to compress the abdomen and provide lumbopelvic stability. The TrA has been observed to work in tandem with the MFD to maintain spinal stability, with reduced recruitment ability and strength in one being linked to poor recruitment
and loss of strength in the other (Hides et al., 2011). Research suggests a relationship between LBP and altered TrA recruitment, where individuals with LBP demonstrate a reduced ability to anticipatorily contract their TrA during upper and lower limb movements (Hodges et al., 1996; Hodges, 2001). Its role in postural control is further supported by its high relative composition of Type-I muscle fibers (Häggmark et al., 1979), and the TrA has been identified as a key area of focus for rehabilitation of LBP (Ferreira et al., 2010).

2.2.3.2. **Internal Oblique (IO)**

The IO laterally flexes and ipsilaterally rotates the trunk (Fan et al., 2014), with research also suggesting that its role in compressing the abdomen to create stability is fairly substantial. Maeo et al. (2013) observed activation of the IO to reach 60% MVC when performing abdominal bracing in the absence of any external resistance, which was 2-3x higher than the other trunk muscles examined. Westad et al. (2010) demonstrated that the IO was the first trunk muscle active 65% of the time during rapid arm movements, with an average delay in the onset of TrA contraction of 48ms compared to the IO. While this may suggest the IO’s importance, relative to the TrA, in stabilizing the trunk prior to potentially destabilizing movements (i.e., anticipatory motor control), the magnitude of such a difference in muscle onsets may have little clinical significance. Large increases in IO activation are also seen during unilateral exercises in order to limit pelvic rotation and increase spinal stability (García-Vaquero et al., 2012). Additionally, significant increases in IO activation have been observed during combination movements, indicating increased activation with increased lumbopelvic stability demands (García-Vaquero et al., 2012).
2.2.3.3. External Oblique (EO)

The EO, whose fibers are perpendicular to the IO, laterally flexes and contralaterally rotates the trunk (Fan et al., 2014). Activation of the EO reaches 27% MVC during abdominal bracing (Maeo et al., 2013), supporting its role in stabilizing the spine. Relative to the IO, the EO has been observed to activate later and only to half the extent of the IO during abdominal bracing (Maeo et al., 2013), suggesting a more supportive role in maintaining IAP and spinal stability.

2.2.3.4. Rectus Abdominis (RA)

The RA, which runs vertically down the anterior wall of the abdomen, flexes the trunk and also plays a role in providing multisegmental stiffness during dynamic movements (Behm et al., 2009). The RA shows a minimal level of activation (18% MVC) during abdominal bracing, suggesting a minor role in contributing to IAP. In regard to stability, it appears its function as a stabilizer is more effective during co-contraction with the surrounding trunk musculature.

2.2.3.5. Multifidus (MFD)

The MFD laterally flexes and contralaterally rotates the trunk (Rosatelli et al., 2008). Though the MFD fibres are short, the muscle’s large cross-sectional area creates a high potential for force generation over a short range of movement, making it well suited for stability (Ward et al., 2009; Rosatelli et al., 2008). The MFD is also recognized for its importance in stabilizing the spine and maintaining a neutral spinal posture, and may account for more than 66% of spinal stiffness in this position (Wilke et al., 1995). The MFD is divided into deep and superficial fibers, which serve different functions. The deep fibers span two vertebral segments and contract tonically, while the superficial fibers span 3-5 segments and contract phasically. This makes the
deeper fibers better suited for stability and posture, while the superficial fibers may function more actively against resistance (MacDonald, 2006; MacIntosh, 1986). In addition, localized atrophy and delayed onset of contraction during upper and lower limb movements has been observed in individuals with LBP (Hides et al., 1994), making the MFD an important area of focus for injury prevention and performance.

2.2.3.6. Quadratus Lumborum (QL)

The QL laterally flexes the lumbar spine (Marieh et al., 2007). Its attachments on the lumbar transverse processes create a large lateral moment arm and allow for high lateral shear resistance, making it an important stabilizer of the lumbopelvic region (McGill, 2009). The QL experiences little change in length during most spinal motions, suggesting that it typically contracts isometrically to allow optimal fibre overlap to be maintained (McGill, 2015). Previous research by McGill et al. (1996) suggests that QL activation increases synergistically with the IO as stability demands increase. Bergmark (1989) states that the QL also has a role in contributing to IAP during abdominal bracing. In football players, a strong relationship has been observed between the cross-sectional area of the QL and the prevalence of back injury during and out of the competitive season in football players, suggesting the importance of QL strength in stabilizing the spine in high impact and unstable sports (Hides et al., 2017).

2.2.3.7. Erector Spinae (ES)

The ES consists of the iliocostalis (lumborum, thoracis, cervicis), longissimus (thoracis, cervicis, capitis), and spinalis (thoracis, cervicis, capitis) muscle groups. For simplicity purposes when studying the low back, the ES can be divided into thoracic and lumbar regions. The ES
muscles extend longitudinally along the spine in their respective regions and provide some lateral movement and stability (Bergmark, 1989). During sagittal lifting exercises (e.g. back squat, deadlift), the ES typically contracts isometrically to support a neutral spine posture. During abdominal bracing, ES activity reaches 19% MVC, similar to that of the RA, suggesting a similar role in providing dorsal stability during this technique. Behm et al. (2005) found activity of the ES to increase when performing exercises on unstable vs. stable surfaces. In contrast, Lawrence et al., (2015) did not observe any differences in ES activity during back squats with an unstable load. Hamlyn et al., (2007) found that back squats elicited more lumbar ES activity, whereas more thoracic ES activity was observed during the deadlift.

**2.2.3.8. Gluteus Medius (GMd)**

The GMd is an abductor of the hip, and is an important pelvic stabilizer (along with the gluteus minimus) during single leg stance by controlling pelvic rotation in the frontal plane. While the anterior fibres flex and internally rotate the hip, the posterior fibres extend and externally rotate the hip (Marieb et al., 2007). In addition, because of its role in limiting knee valgus, it is suspected to have additional importance in attenuating forces from the lower limb, hip, and spine, with altered recruitment patterns potentially increasing the risk of knee and back injuries (Hart et al., 2007; Nadler et al., 2000). Individuals with patellofemoral pain syndrome were observed to have an average of 18% less hip abduction strength and 17% less hip external rotation strength during a single leg squat, further indicating the importance of GMd strength on lower limb kinematics (Souza et al., 2009). In the context of both general and athletic populations, it has been suggested that core strengthening exercise regimes focus on strengthening and improving the coordination between abdominal, paraspinal, and gluteal
muscles (Nadler et al., 2000). Research has also shown increased lumbar extensor strength following pelvic stabilization training (Pollock et al., 1989) and that hip abduction strength is highly correlated with trunk lateral flexion strength (Stickler et al., 2015).

2.2.3.9. Gluteus Maximus (GMx)

The GMx extends and laterally rotates the hip and also plays a major role in stabilizing the pelvis during trunk rotation and when a shift in the center of mass is experienced (Nadler et al., 2000). This suggests its importance for sport, where movements are generally more reactive and involve high power or force production. The importance of the GMx for both pelvic stability and power production during sprinting (Bartlett et al., 2014) makes it a primary focus of training for sports involving running and jumping. Very high levels of GMx activation can be observed during the back squat (Caterisano, 2002), with increases in hip extension strength and sprint speed being observed after periods of squat training (Wisløff et al., 2004).

2.2.3.10. Latissimus Dorsi (LD)

The LD has been observed to contribute to trunk motion, most notably extension and rotation, as well as general trunk control during movement (Vera-Garcia et al., 2011). It is for this reason that some authors suggest training of the LD is important for neuromuscular control of the trunk and rehabilitation programs for individuals with LBP (Vera-Garcia et al., 2011). Given it’s origin, the LD also influences anterior pelvic tilt. However, little research has been directed towards analyzing the activity of the LD during stability or core control exercises.
Table 1
*Origins, insertions, and actions of important trunk and hip musculature (Marieb et al., 2007).*

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Origin</th>
<th>Insertion</th>
<th>Actions</th>
<th>Unilateral</th>
<th>Bilateral</th>
</tr>
</thead>
<tbody>
<tr>
<td>Transversus Abdominis</td>
<td>Inguinal ligament, iliac crest, thoracolumbar fascia, costal cartilage (7th-12th ribs)</td>
<td>Xiphoid process, linea alba, pubic crest</td>
<td>n/a</td>
<td>Abdominal compression</td>
<td></td>
</tr>
<tr>
<td>Internal Oblique</td>
<td>Inguinal ligament, iliac crest, and lumbodorsal fascia</td>
<td>Linea alba, 10th-12th ribs</td>
<td>Trunk lateral flexion,</td>
<td>Abdominal compression</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Ipsilateral trunk rotation</td>
<td>Abdominal compression</td>
<td></td>
</tr>
<tr>
<td>External Oblique</td>
<td>5th – 12th rib</td>
<td>Iliac crest, pubic tubercle, linea alba</td>
<td>Trunk lateral flexion,</td>
<td>Abdominal compression</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Contralateral trunk rotation</td>
<td>Abdominal compression</td>
<td></td>
</tr>
<tr>
<td>Rectus Abdominis</td>
<td>Crest of pubis</td>
<td>Costal cartilage of 5th-7th rib, xiphoid process</td>
<td>n/a</td>
<td>Trunk flexion</td>
<td></td>
</tr>
<tr>
<td>Multifidus</td>
<td>Sacrum, iliac crest</td>
<td>Spinous processes of vertebrae</td>
<td>Trunk lateral flexion,</td>
<td>Trunk extension</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Contralateral trunk rotation</td>
<td>Trunk extension</td>
<td></td>
</tr>
<tr>
<td>Quadratus Lumborum</td>
<td>Iliac crest, iliolumbar ligament</td>
<td>12th rib, transverse processes of the lumbar vertebrae</td>
<td>Lumbar lateral flexion</td>
<td>Lumbar extension</td>
<td></td>
</tr>
</tbody>
</table>
| Erector Spinae (lumbar & thoracic) | Sacrum, iliac crest, spinous processes of lumbar vertebrae and T11-12 | *Iliocostalis: 1st-12th ribs*
<p>|                               |                                                                        | <em>Longissimus: transverse processes of T1-12</em>| <em>Spinalis: spinous process of T1-8</em> | n/a         | Trunk extension |
|                               |                                                                        |                                                |                               |             |                             |</p>
<table>
<thead>
<tr>
<th>Muscle</th>
<th>Origin</th>
<th>Insertion</th>
<th>Action</th>
<th>Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gluteus Medius</td>
<td>Surface of ilium</td>
<td>Greater trochanter of femur</td>
<td>Abducts hip, flexes &amp; internally rotates hip (ant. fibres), extends &amp; externally rotates hip (post. fibres)</td>
<td>n/a</td>
</tr>
<tr>
<td>Gluteus Maximus</td>
<td>Surface of ilium, lumbar fascia, sacrum, sacrotuberous ligament</td>
<td>Gluteal tuberosity of femur, iliotibial band</td>
<td>Extends &amp; externally rotates hip</td>
<td>n/a</td>
</tr>
<tr>
<td>Latissimus Dorsi</td>
<td>Spinous processes of T7-12, thoracolumbar fascia, iliac crest, 9th-12th ribs, inferior angle of scapula</td>
<td>Intertubercular groove of humerus</td>
<td>Adducts, extends, &amp; internally rotates shoulder, Ipsilateral trunk rotation</td>
<td>Trunk extension</td>
</tr>
</tbody>
</table>
2.2.4. Global vs. Local Functional Groups

From a functional standpoint, the musculature of the trunk can be grouped together based on mechanical roles. One common method of categorization within the literature separates the muscles of the trunk into “local” and “global” functional groups (Bergmark, 1989). The local group is characterised by muscles that originate or insert directly on the vertebrae. This generally includes deep muscles, being the TrA, MFD, ES (lumbar), diaphragm, and pelvic floor muscles. These are the muscles that play the greatest role in controlling spinal curvature and producing sagittal and lateral stiffness in order to maintain mechanical stability and posture (Bergmark, 1989). Notably, the TrA and MFD are commonly the focus of LBP rehabilitation as localized atrophy and altered neuromuscular activation and control is common in these muscles in this population (Bergmark 1989; Hides, 1994; Hides et al., 2011).

The global functional group is made up of muscles that transfer load directly to the thoracic cage and pelvic girdle. These are the ES (thoracic), RA, IO, EO, QL, GMx, and LD. The main function of the global group is to balance external load so that the resulting transference of force can be controlled by the local system; it is for this reason that large variations in the distribution of external load may only give rise to small variations in resultant spinal load (Bergmark, 1989). The LD doesn’t play as substantial a mechanical role in spinal stability and force transfer in the lumbar spine, though it does play a role in thoracic stability (Bergmark, 1989).

The global group contains the main muscles involved in the development of IAP during abdominal bracing. IAP itself plays a global and local mechanical role as well. Globally, it acts directly on the thoracic cage and curved global muscles, while locally, it transfers force posteriorly through the lumbar spine to support flexion and potentially increase extensor activity
While the global system responds to changes in line of action and the local system responds to postural changes, both respond directly to the magnitude of change (Bergmark, 1989). Given what we know, both systems play separate but important roles in generating spinal mechanical stability.

### 2.2.5. Abdominal Hollowing

The abdominal hollowing manoeuvre (AHM), whereby the individual draws the abdominal wall inward (i.e., the individual “sucks in” their belly), is a technique that is used to promote or selectively recruit the deep abdominal and pelvic floor muscles, most notably the TrA, IO, and diaphragm (Critchley, 2002; Hides et al., 2006; Lee et al., 2016;). Beith et al. (2001) found that by performing the AHM, the IO could be activated to a significantly higher degree than the EO, which is important for spinal stability given its early onset of contraction in response to increased stability demands (Westad et al., 2010). Several studies have observed an increase in TrA thickness when performing the AHM (Critchley, 2002; Hides et al., 2006;), which has been associated with activation of the TrA bilaterally (Hides 2006; McMeeken et al., 2004). Kim et al. (2015) found that activation of the TrA and IO was greater when consciously performing the AHM during basic isometric exercises when compared to performing the exercises in conditions with surface instability. While some research has suggested a relationship between TrA activation and increased spinal stability (Hodges, 1997; Southwell et al., 2016), a number of researchers have shown that the AHM is more useful for re-education in controlled environments and has less practicality in providing spinal stability during activity. Research by Vera-Garcia et al. (2007) found that performing the AHM in response to sudden trunk perturbations was not effective in maintaining spinal stability (measured via spinal kinematic
changes). Southwell et al. (2016) found that while IO and TrA activation increased during the AHM, no changes occurred in the ES, which may suggest that the AHM is not effective in increasing superficial muscle activation to promote stability. While increases in both TrA and IO were observed, peak TrA activation was higher. However, increases following AHM biofeedback training were greater for the IO than the TrA (87.9% vs. 72.9%) (Southwell et al., 2016). Increases in spinal flexion were also observed during the AHM, which was attributed to the posterior pelvic tilt created during the use of this technique (Vera-Garcia et al., 2007). While an increase in TrA activation and/or thickness is observed during and following training of the AHM, it may have limited use with dynamic movement in combination with perturbations (i.e. sport performance). The AHM also reduces the moment arm for the RA, EO, and IO, thereby reducing their potential for spinal stability (Grenier & McGill, 2007). While the AHM is an effective method of training posterior pelvic tilt and strengthening the TrA, it does not appear to be an effective method of consciously stabilizing the spine.

2.2.6. Abdominal Bracing

Abdominal bracing is another technique used to increase spinal stability. Bracing is performed by voluntarily co-contracting various dorsal, ventral, and pelvic floor muscles to increase IAP. This creates a high pressure air pocket within the abdominal cavity that increases stability, and may be enhanced by first drawing in air to further increase IAP (McGill, 2001). Stokes et al. (2011) found that a two-fold increase in IAP increased spinal stability on average by a factor of 1.8. Abdominal bracing, while associated with a slight increase in lumbar compression (Vera-Garcia et al., 2006; Vera-Garcia et al., 2007; Grenier et al., 2007), has been shown to greatly increase spinal stability during sudden trunk perturbations and lifting tasks
compared to the AHM (Vera-Garcia et al., 2007; Grenier & McGill, 2007). Grenier et al. (2007) found that the AHM resulted in 32% less stability than bracing. Maeo et al. (2013) found abdominal bracing to be highly effective in activating the IO (60% of MVC), and also moderately effective in activating the EO, RA, and ES (27%, 18%, 19% MVC, respectively). Research seems to suggest different recruitment strategies and muscular contributions when performing an abdominal brace vs. the AHM. Using both EMG and motion capture, along with computerized simulations, Grenier et al. (2007) studied the impact of abdominal bracing on spinal stability along with the influence of specific muscles on spinal stability during bracing. It was observed that the bracing technique improved stability by 32%, with a 15% increase in lumbar compression. However, the TrA’s contribution to stability was less than 0.14%. This raises the question of whether the TrA has significant importance in the context of heavy resistance training, where bracing is a common practice to maintain a neutral spine and proper positioning. Both the IO and TrA have recognized importance for spinal stability, but the IO appears to have a greater role in promoting overall spinal stability. In terms of application to compound movements or exercises, isolated abdominal brace training (separate from exercise) has been shown to increase hip and trunk extension strength, which would have implications for its importance during lower body exercises involving either or both of these actions (Tayashiki et al., 2016).

2.3. Strength Training

Resistance training has become a staple for competitive athletes of various sports. It can be easily manipulated to elicit specific adaptations using different training strategies (Haff et al., 2015). Maximal or absolute strength is defined as the maximal force that a muscle or muscle
group can generate at a specific velocity (Knuttgen et al., 1987), while power is calculated as the product of force on an object and the object’s velocity in the direction in which the force is exerted (Baechle et al., 2008). In the context of strength and power sports, resistance training is crucial for increasing performance variables, as resistance training has been shown to be extremely effective in promoting increases in maximal strength, muscular hypertrophy, neural drive, and rate of force development (Haff et al., 2015; Aagard et al., 2002; Schoenfeld et al., 2015). Some research even supports the implementation of multifaceted resistance training in youth athletes as a means of improving performance and preventing injury, assuming proper coaching is reinforced and technical deficiencies are considered (Faigenbaum et al., 2010). It should be noted that optimal gains are ensured by both the magnitude of effort from the athlete and systematic structuring of the training stimulus (e.g., exercise, intensity, volume, etc.) that will ultimately induce the desired adaptation (Kraemer et al., 2004). Relative to the individual, exercises must be performed at above regular levels of exertion to elicit adaptation, so that the systems involved (e.g., neuromuscular, phosphagen, etc.) are overloaded and subsequently become more efficient in the given environment. Knowing that adaptation is specific to the type and level of stress, exercise selection should be done with the purpose of provoking sport-specific adaptations.

2.3.1. Overload

Overload denotes the assignment of a protocol of greater training intensity or volume than the athlete is accustomed to (Haff et al., 2015). Typically, overload is achieved by an increase in intensity (i.e., external load), which has been shown to increase both neural activation and mechanical overload of the muscle fibers. High load resistance training that causes greater
neuromuscular activity has been shown to increase neural drive (Aagaard et al., 2002), increase fiber recruitment velocity (Duclay et al., 2005), increase neural excitability and firing frequency (Duclay et al., 2005; Linnamo et al., 2003), increase synaptic conduction velocity of the motor axons (Aagard et al., 2002; Bawa, 2002), and improve neuromuscular coordination and efficiency (Carroll et al., 2001). These adaptations have been shown to lead to increases in maximal force production, rate of force production, and movement quality (Aagard et al., 2002; Haff et al., 2015). Greater overload also requires greater recovery time, meaning that appropriate rest should be taken between sessions in order to allow for fibril and neural recovery but still maintain the stress stimulus required to induce adaptation (Fleck et al., 2014).

Overload can be achieved by manipulating other aspects of training as well, such as increasing training volume and/or frequency, or changing exercise type (Haff et al., 2015). Different means of overload will elicit different adaptations, some of which may limit others. High load or high volume exercise can also promote mechanical changes to the muscle fibers by causing micro trauma, which is then repaired, resulting in a stronger muscle fiber capable of withstanding greater loads (hypertrophy). However, research shows that this mechanical breakdown occurs primarily during the eccentric portion of the movement, where relative neural activity is actually lower (Blazevich et al., 2007). Isolated eccentric training has been shown to increase muscular hypertrophy (Norrbrand et al., 2008), though increases in strength may only occur eccentrically as well (Vikne et al., 2006). Hypertrophy of muscle fibers can also occur using lower loads for higher repetitions, assuming sets are performed to failure (Schoenfeld et al., 2015). It has been suggested that lower load, higher repetition sets actually increase rates of muscle protein synthesis for longer periods than higher loads when performed to failure (Mitchell et al., 2012). In saying this, high repetition exercise is associated with endurance
adaptations, which may be less favourable for many strength and power athletes (e.g., conversion of type-II to type-I fiber type, etc.) (Haff et al., 2015). Specific increases in training volume may be purposely incorporated into certain training phases to improve capacity and recoverability, which influences training quality. Given these discrepancies, specific requirements of the sport must always be taken into account when programming exercise.

2.3.2. Specificity

The SAID (specific adaptation to imposed demand) principle, also known as specificity, refers to the concept whereby an athlete is trained in a manner to produce a specific adaptation or training outcome (DeLorme, 1945). The underlying principle is that the type of demand placed on the body dictates the type of adaptation that will occur. In strength and conditioning, the goal of training an athlete is to produce adaptations that will offer the greatest carryover to sport performance by effectively improving sport-specific performance variables (e.g., speed, agility, power, etc.) (Haff et al., 2015). Training should also take into account risk of injury and employ methods to reduce this risk as much as possible. When training a distance runner, the focus should be placed on improving musculoskeletal endurance (increase type-I fiber composition, increase mitochondrial quality and quantity) and aerobic capacity (increase VO2), utilizing high volume, low-moderate intensity aerobic and resistance exercise, as well as some specific power training (McArdle et al., 2007; Gollnick et al., 1972; Edge et al., 2006). Additional specific exercises may be selected to reduce risk of knee or back injury (common areas of injury for distance runners) (van Gent et al., 2007; Schache et al., 1999). Given that distance running does not involve a high level of absolute strength, it would be counterproductive to train the athlete in
this manner. Improvements in running economy (e.g., improved movement pattern quality and efficiency) are more effectively trained through direct practice.

The use of specific resistance training exercises may actually become counterproductive depending on the athlete’s phase of training and the type of adaptation the phase is designed to produce, as overly specialized movements may reduce the extent to which general adaptations are made (Moore et al., 2007; Wahl et al., 2008). For a football player, absolute strength and power can be improved utilizing compound movements at a high load, low velocity (e.g., back squat) (McBride et al., 2009) or high velocity, low-moderate load (e.g., jump squat, power clean) (Baker et al., 2001). While these have been shown to increase football-specific performance variables, these exercises are less comparable to football specific movement patterns (depending on position). Likewise, while football practice may improve skill aspects of the game, the external load on the agonist muscles is incomparable to that experienced during high load resistance training. A balance must be obtained in order to promote general adaptation, while allowing for enough specificity to improve strength and power during sport specific movements, and recovery time to prevent injury.

In addition, too much focus on one performance variable may have negative impacts on other aspects of performance (e.g., endurance and speed training should be performed separately due to the role of excessive fatigue on neuromuscular recruitment and power) (Moore et al., 2007). Exercise programming will progress from general to more specific as an athlete approaches his or her competitive season. This means that exercises may be selected or modified to match specific muscle group utilization, energy system usage, rest times, movement patterns, or stability requirements. While decreasing surface stability (i.e., balance) has a tendency to decrease neuromuscular activity of the primary agonist(s) and potentially limit general strength
gain in the lower limb in trained individuals, it can progressively increase balance and dynamic stability, which may increase sport performance and reduce injury risk (Behm et al., 2010).

2.3.3. The Back Squat

The back squat (see Figure 1) is often recognized for its ability to prominently increase lower body maximal strength and power output in athletes, and increase relevant performance variables (Wisløff et al., 2004). The back squat can be modified to stimulate different adaptations by altering set and rep schemes, tempo, and load. The majority of joint movement during the back squat occurs at the ankle, knee, and hip. However, a high level of muscular activity is required throughout the trunk in order to maintain spinal stability during the movement.

Figure 1

*Visual representation of the barbell back squats. (Retrieved from: www.fitlablv.com)*

Peak moments at the talocrural joint have been observed to reach approximately 50-300Nm during the back squat, which is significantly less than that of the knee and hip (Escamilla
et al., 2001). It is suggested that musculature at the ankle plays a greater role in stabilizing the joint than producing any movement (Hung et al., 1999). However, the center of mass (COM) is extremely important during the back squat. Ideally, the lifter’s COM (which will be affected by the presence of the barbell weight) must be over the mid foot, otherwise anterior-posterior balance is affected. It has been shown that during the back squat, a small internal plantarflexion moment does occur from the initial lift off, which decreases during the descending phase of the squat (DSC) as the hips shift posteriorly and knees move anterolaterally, and increases during the ascending phase of the squat (ASC) (Schoenfeld, 2010). This shift may also be caused by the co-activation between the tibialis anterior and gastrocnemius during the DSC to stabilize the ankle (Dionisio et al., 2008). The soleus plays little role during the back squat, but may be involved more eccentrically during higher degrees of ankle dorsiflexion (Toutouni et al., 2000). The gastrocnemius also shows little activity during the back squat, but tends to increase during deeper knee flexion as the moment arm peaks (Donnelly et al., 2006). Plantarflexor activity may increase with forward lean or if the COM shifts anteriorly during ASC in an attempt to maintain the COM over the mid-foot. Limitations in ankle ROM can be detrimental when trying to develop proper squat movement patterns, as it limits forward knee displacement and increases hip flexion, thereby increasing stress on the spine (List et al., 2013). Hemmerich et al. (2006) found that a dorsiflexion angle of at least 38.5 ± 5.9 degrees was necessary to perform a full squat while maintaining heel-to-floor contact.

During DSC, the thigh is externally rotated as the hip is flexed to allow for proper anterolateral movement of the knee, which is followed by internal rotation during hip extension during ASC. This causes a shift in center of rotation of the knee joint throughout the movement (Donnelly et al., 2006). Increases in patellar tendon tensile force are observed between 30° and
130°, where forces increase from 2000N to 6000N (Nagura et al., 2002). Ultimate tensile strength of the patellar tendon ranges from 10000-15000N, so it is capable of handling such loads (Donnelly et al., 2006). However, tendon creep, whereby slow deformation occurs under the influence of mechanical stress, may occur following long-term training at high volume or intensity (Maganaris et al., 2002), given the visco-elastic nature of the tendon and its non-linear deformation during loading (Pearson et al., 2006). Frequent, repetitive loading through the tendon can cause greater strain at the same level of force, and may reduce overall load tolerance of the tendon (Magnaris et al., 2002).

High knee joint compressive forces have been observed in powerlifters using very high loads (>2.5x body weight), where forces have been observed to increase with deeper flexion, reaching 3500N at 30°, 5500N at 60°, and 8000N at 130° of flexion (Nagura et al., 2002). It is suspected that this compressive force plays a role in initiating the contraction of the quadriceps and hamstrings to stabilize the knee (Hemmerich et al., 2006). The hamstrings are especially important as they prevent anterior translation of the tibia and reduce shear force at the knee joint and stress on the anterior cruciate ligament (ACL) (Hemmerich et al., 2006). There is research to suggest that the anterior shear forces during the squat peak during the first 60° of knee flexion, with ACL peak forces generally occurring at 15-30°, and which decrease with increasing knee flexion (Nagura et al., 2002). While this increases stress on the posterior cruciate ligament (PCL), total strain on the ACL and PCL during the squat is far below the ultimate tensile strength of these ligaments (Sahli et al., 2008), and may actually allow them to become stronger over time as the connective tissue adapts to the repetitive strain (Buchanan et al., 2002). While the vastus medialis and lateralis show equal contribution to knee extension during the squat, the rectus femoris has been shown to produce half the force of each of these muscles (Hwang et al., 2009).
During the back squat, hip flexion, abduction, and external rotation are observed during DSC, and extension, adduction, and internal rotation occur during ASC. During a back squat, 95 ± 27° of hip flexion is observed at the parallel depth (i.e., when a line between the top of the knee and crease of the hip is parallel to the floor), and increases slightly further when squatting to maximal depth (i.e., hip joint falls below knee joint) (Hemmerich et al., 2006). Internal hip joint moments have been observed to increase during DSC, and peak at the bottom of the movement (Nagura et al., 2006). The gluteus maximus is the primary agonist at the hip and is an important pelvic stabilizer, and is thought to play a role in stabilizing the knee through its attachment to the iliotibial band (Rasch et al., 1989). While the moment arm of the gluteus maximus is smallest at 90° of hip flexion, extension force peaks at this point. This is thought to be attributed to an optimal force length relationship, overcoming the mechanical disadvantage by maintaining optimal sarcomere length (Escamilla et al., 2001).

Gluteus maximus activation (where activation is represented as a % of the total neuromuscular activity produced by all muscles analyzed) is highly influenced by squat depth, where average muscle activity has been shown to be significantly higher during a full squat (35.47 ± 1.45%), compared to a partial squat (16.92% ± 8.78%) (Caterisano et al., 2002). Despite increased posterior shear at the bottom of the squat, research suggests that there is little change in hamstring activity throughout the squat movement, with peak activity occurring between 10° and 70° of hip flexion. This can be attributed to the hamstrings’ biarticular structure, where it functions as both a hip extensor and knee flexor and therefore maintains a similar length throughout the movement, resulting in a consistent force output (Escamilla et al., 2001; Walsh et al., 2007).
When performing the back squat, the spine should be kept in a neutral position in order to
decrease shear and compression. Compressive force reaches extremely high levels during a
loaded back squat, with a 0.8-1.6x bodyweight load producing an estimated compressive load of
6-10x bodyweight at the L3/L4 joint during a partial squat (Cappozzo et al., 1985). In vitro
testing has determined ultimate compressive strength of these vertebrae to be around 7800N in
individuals under 40 years of age (Adams et al., 2000). This may suggest either trained
individuals’ spines adapt to this mechanical stress via bone remodelling and increasing BMD in
order to increase compressive tolerance, or that the modelling approaches used to calculate
compressive force and in-vitro methods used to determine ultimate strength are inconsistent.

When squatting with a flexed spine, the moment arm for the lumbar ES is decreased, and
tolerance to compressive load is consequently reduced and transferred from the muscles to
passive tissues (Matsumoto et al., 2001). Increases in shear force are also observed during
flexion, which is thought to be caused by both spinal position and changes in the fiber orientation
of the ES, resulting in a reduced ability to counteract shear force (Noyes et al., 1984).

Hyperextension also increases spinal compressive load, with a 2° increase in extension
increasing posterior AF compression by 16% (Dolan et al., 1999). IAP provides an anti-flexion
moment in the lumbar region which reduces active contraction of the ES and loading associated
with muscular tension (Schoenfeld, 2010). While McGill et al. (1990) found that holding one’s
breath to increase IAP during a weighted back squat significantly reduced lumbar load, recent
research has noted increased compression with increasing IAP (Vera-Garcia et al, 2006; Vera-
Garcia et al., 2007; Grenier et al., 2007). It should be noted that accuracy of IAP calculations are
very dependent on the modelling approach used.
Vakos et al. (1994) found that peak lumbar compression during lifting increased with an increase in movement speed. While lifting at 90-100% 1-repetition maximum (1RM) typically occurs at a low velocity, average and peak compression is extremely high due to the external load (Sahli et al, 2008). In fact, a positive linear correlation was observed between spinal compression and external load (Walsh et al., 2007). Speed or power training utilizes loads normally ranging from 30-70% 1RM to optimize power output (Baker et al., 2001), though load may vary between exercises. It has been observed that when squatting heavier loads (80% vs. 40%), individuals have a tendency to lean forward, which produces compensatory hyperextension of the lumbar spine, thereby increasing compression and placing greater strain on the posterior AF (Adams et al., 2000). This may also suggest that individuals who use high loads and are less well-trained or who have more trouble maintaining proper technique under heavier load may experience additional increases in compression. A direct comparison of relative spinal loading at high velocity – low load vs. low velocity – high load has not been published.

2.4. Stability Training

Stability training can be defined as exercise training designed to improve one’s capacity to maintain dynamic stability at one or more joints by means of improving neuromuscular control and proprioception. It can be incorporated into training regimes for both general and athletic populations. While stability training does not typically elicit increases in hypertrophy, strength, or power, it has been theorized to improve proprioception and joint stability, leading to improvements in balance (Behm et al., 2010; Behm et al., 2013). From a specificity standpoint, stability training is extremely important in the sense that many sports have high stability requirements, and is the reason that many “sport-specific” exercise progressions involve
decreasing the level of stability. Stability exercises diminish involvement of the primary agonist(s), while increasing activity in surrounding muscles that are involved in stabilizing the joint(s) throughout the movement (Saeterbakken et al., 2013). This is especially important with respect to the spine. When an unstable condition is created, particularly for the lower limb, activity of both deep and superficial trunk musculature is greatly increased in order to meet the high stability requirements of the lumbopelvic complex and reduce ligamentous stress (Behm et al., 2010). By increasing the level of instability of the back squat, increases of 20-30% were observed in the thoracic and lumbar ES in order to stabilize and protect the spine (Anderson et al., 2005). Basic instability training (i.e. minimal stability & coordination demands) is also implemented in rehabilitative settings for individuals with LBP, in order to regain neuromuscular control of spinal stabilizers (Macedo et al., 2009).

A distinction should be made between stability training in athletes vs. the general population or individuals with LBP. Functional stability is dependent on both local and global muscle function, with stability training targeting both to some extent. Activation of the local system and maintenance of low force output is vital in increasing local muscle stiffness at the vertebral joints to control excessive translation and maintain a neutral position to decrease ligamentous and capsular strain (Comerford et al., 2001). Global muscle strength is important for power generation and provision of eccentric control during joint motion when controlling and manipulating load. This includes concentric contractions through the full ROM, isometrically holding a position, and eccentrically controlling load against gravity and rotational movement (Comerford et al., 2001). While LBP does have an effect on the global muscular system, many treatment strategies aim to improve recruitment strategies of the local system, as local muscle dysfunction has been associated with the presence of LBP (Hodges, 1996; Hodges, 2001). It is
for this reason that the combination of trunk stability and trunk strength are critical to maximizing sport performance, as athletes must often maintain stability while producing high force.

In the context of a sport such as American football, while the stability requirements are very high, they are paired with high strength requirements concentrically, isometrically, and eccentrically (Hoffman, 2008). Although stability requirements must be progressed in order to see continuous benefit (Wahl et al., 2008), football players require high dynamic stability under high load and thus the production of joint stability in both the trunk and lower limbs must be efficient to meet such demands. From a specificity standpoint, it may be practical to apply external loading to stability exercise to elicit greater specific adaptations and increase the competency of athletes to handle load in unstable conditions. Likewise, if the stability requirements are too high relative to the load or vice versa, it may be counterproductive in the sense that the risk (injury) outweighs the reward. It would be practical to seek a balance between instability and magnitude of external load in order to understand how trunk muscle activity is affected by increasing external load under low-moderate stability conditions.

2.5. Measurement

2.5.1. Electromyography

Electromyography (EMG) is a scientific tool that measures the electrical activity generated within muscles during contraction, indicating the timing and level of neuromuscular activity (Reaz et al., 2006). Through EMG, neuromuscular recruitment strategies and overall
activity can be examined. EMG most often uses surface electrodes (surface EMG), where signals are collected at the skin’s surface superficial to the given muscle.

Muscles produce no detectable electrical signal at rest. The action potentials from one or more motor units during a muscle contraction is what the EMG signal represents. In any given recruitment pattern, a number of motor units are activated. The sum of electrical activity determines the magnitude of the signal, which is detected by electrodes and amplified by the EMG electronics and software. Amplifying the signal (also known as signal “gain”) makes the raw signal larger so that latency and amplitude can be analyzed. Motor units that lie closer to the surface of the muscle contribute most to the EMG signal (with surface EMG), with deeper motor units having to pass through more bodily tissue to reach the surface. This tissue resistance means that less activity of the deeper motor units is represented by the EMG. Additionally, because these tissues have a tendency to absorb high-frequency components of the signal, they act as a low pass filter for the signal. Excess adipose tissue would introduce additional resistance, thereby absorbing part of the signal and reducing the level of activity detected by the electrode.

With more action potentials occurring and more, larger motor units becoming active with increasing contraction intensity, signal emission and EMG signal amplitude increases.

EMG signals acquire electrical noise during collection, which can contaminate the signal. Inherent noise from equipment cannot be eliminated, but is minimized when using higher quality electronics. Motion artifact, caused from movement either between the electrode and skin interface or the electrode and cable can skew data by producing a low frequency signal that is detected by the electrode. Removal of this noise is highly dependent on EMG set up and design. Skin moisture, oil, and even dead skin cells may cause impedance of the signal. To avoid this, the skin is typically abraded and cleaned with alcohol (Lehman et al., 1999). It should be noted
that when using surface EMG to measure deeper muscles, cross-talk from superficial muscles may influence the recorded signal as a form of noise, specifically in the case of the lateral abdominal muscles where multiple muscles lay in close proximity and can be detected by a single electrode. For this reason, appropriate electrode placement is important when recording raw EMG signals to reduce the extent of this cross-talk (McGill et al., 1996). It is worth noting that while the TrA and QL have been described as being important in providing lumbopelvic stability (McGill, 2015), both their depth and proximity to other trunk muscles makes surface EMG an inaccurate means of measuring their activity due to the impedance of the EMG signals between the muscle and electrodes (e.g., skin and other soft tissue) and the amount of cross-talk from other muscles. To obtain accurate measures, invasive fine wire electrodes would be required (McMeeken et al., 2004).

Additional (causative) factors that directly affect the EMG signal itself can be either extrinsic, intrinsic, intermediate or deterministic. Extrinsic factors are things such as electrode structure and placement (e.g., shape of electrode, distance between electrodes, electrode surface area, and surface or location of electrodes with respect to motor points in the muscle). Intrinsic factors are physiological, anatomical, or biomechanical, such as the number of active motor units, fiber composition, blood flow, fiber diameter, location of active fibers, or tissue between the muscle and the electrode. Intermediate factors are affected by either extrinsic or intrinsic factors and impact the resultant EMG signal. These can be things such as electrical potential differences in neighbouring muscles and production of crosstalk, or conduction velocity in motor neurons (De Luca, 1997). Lastly, deterministic factors are those that facilitate the action potential from the CNS and subsequently affect the shape and amplitude of the EMG, such as the number of active motor units, firing frequency, or mechanical interactions between muscle fibers (De
Luca, 1997). Different types of noise must be prevented or removed as much as possible to obtain a high signal-to-noise ratio and a clearer EMG recording of the given muscle (Reaz et al., 2006).

Noise can be greatly reduced by using a pre-amplified electrode that eliminates noise before it travels along the cable and picks up additional noise, which is then amplified by the computer. Once the signal has been amplified, it must be processed. A filter is applied by the EMG instrument itself (analog filter) or the EMG software (digital filter). Typically, this involves using a band pass filter to eliminate additional noise at both the low end (e.g., electrical noise from wire sway or motion artifact) and the high end (e.g., tissue noise from movement at the electrode site) of the frequency spectrum. Greater nervous innervation and frequent activation mean that a higher range must be used to detect higher frequency components in the EMG signal, while still being able to detect lower frequency signal components (especially during muscular fatigue where frequency spectrum decreases) (Criswell, 2010).

The recorded raw EMG signal must be processed in order to determine average and peak activation values. Unprocessed EMG signals typically have an amplitude of 0-6 mV and a frequency of 10-500Hz. Filters are used during collection to eliminate noise. While notch filters may be used to remove specific frequency noise, they also remove EMG data, whereas band pass filters can remove high and low frequency noise. The signal must be rectified, whereby negative values are made positive, though this may serve as an intermediate step before further analysis. One method for processing EMG is to use a linear envelope technique, where raw EMG undergoes full-wave rectification followed by a low-pass filter (4-10Hz). This technique reduces the frequency content of the EMG signal, improving the ease of interpretation of EMG amplitude and the detection of activity onset. One common and effective method used to process EMG is
root-mean-square (RMS), which involves: 1) squaring the value of each raw EMG data point (thereby making all negative values positive in the process), 2) calculating the mean of the resulting squared values using a moving window of specified length over the entire EMG signal, and 3) calculating the square root of the resulting mean values (Basmajian et al., 1985). This method is designed to quantify the intensity of several events of the EMG signal, and is frequently chosen because it reflects the amount of total physiological activity in a given motor unit(s) during contraction (Fukuda et al., 2008).

EMG signals may need to be normalized depending on the study design, whereby the signal is expressed relative to a reference value (percentage). This is done by comparing the signal obtained during the task of interest with the signal obtained from a reference technique or exercise. This may not be required for all types of data analysis (e.g. determining onset of contraction). However, normalization is necessary for any comparative analysis of signal amplitude (e.g., between different individuals, muscles or between different days) (Lehman & McGill, 1999). The use of absolute amplitude values does not consider participant-specific factors, such as subcutaneous fat thickness, which directly affects signal amplitude by impeding the signal from the muscles to the sensor. One must also consider the body’s tendencies to store fat in specific locations, and that certain muscles groups may have more impedance from surrounding subcutaneous fat than others (e.g., abdominal muscles). Additionally, excessive electrical noise may skew the raw EMG data within and between individuals. Likewise, causative factors pertaining to equipment set up and physiological variability may result in differences in the EMG data. Testing between days may introduce significant differences in raw EMG data if procedures are not replicated closely. Normalizing EMG data reduces the impact of
these factors because any noise/error present will be constant during the normalization technique(s) and task(s) of interest for a given participant.

Various normalization techniques have been described in the literature, with no consensus of a true “gold standard” (Burden, 2010). The most common method of normalizing EMG signals from a muscle uses the EMG obtained from the same muscle during a maximal voluntary isometric contraction (MVIC) as the reference value, which can be obtained using a manual resistance exercise or machine. It is recommended that a minimum of three repetitions (3-5 seconds each) be performed separated by at least two minutes of rest to reduce the influence of fatigue (Mathiassen et al., 1995).

Problems arise when selecting the most effective exercise or test to be used to produce maximal neural activity of a given muscle (Ekstrom et al., 2005). Vera-Garcia et al. (2010) found that the highest EO activity with isometric holds were observed during resisted upper trunk lateral flexion (resisted lateral flexion while side-lying on a table), whereas the highest IO activity was observed during resisted lower trunk lateral flexion (resisted downward movement of the pelvis during a side bridge). The highest RA activity was observed during resisted trunk flexion while supine, while ES activity was highest during resisted trunk extension while prone. Similar findings were observed by Ng et al., (2002) who examined normalization exercises for trunk muscles in participants with and without LBP using maximal isometric contractions (flexion, extension, right and left lateral bending, right and left axial rotation) performed while standing with contractions held against physical constraints. It is important to note that while normalization values obtained from maximal isometric contractions may be reliable and generally recommended, such values have been shown to be lower than maximum amplitude values obtained during the performance of certain dynamic tasks (Roth et al. 2016).
2.5.2. Motion Capture

Motion capture is the process of recording human movement by using anatomically-placed markers, which are tracked by a series of cameras and visualized using computer software. Markers must be placed appropriately to define the segment(s) being analyzed and the coordinate system of the segment(s) so that the software is able to identify how and where the marked segment(s) is moving. Through computer software, movements can be analyzed and specific kinematic and kinetic variables (e.g., joint angles between two adjacent segments) can be calculated based on the anatomical model that is being applied.

The anthropometric model used consists of a kinematic chain of links that each represent a portion of the human body (segment). Bony segments are considered non-deformable and are represented as rigid bodies. While soft tissue can be modelled as rigid, and often is without substantial error, some research suggests that soft tissue should be considered deformable. In ignoring deformability, measurement of absolute and relative segment movement is hindered, ultimately influencing the validity and applicability of the results (Cappozzo, 2002). This can appear in cases where inertia causes movement in these tissues, which causes the markers to move in addition to the segments and can be particularly problematic during higher velocity movements (Hatze, 2002), suggesting that slower movements would reduce this error. Low-pass filtering may reduce this error to some extent. Assuming one does not account for deformity of the segment, it may be represented as a rigid body, and a vector quantity need not be applied to the individual markers of the segment. Once the segment is represented with respect to a local frame, it is possible to derive the relative position of the segment and the vector quantities held by each marker or group of markers with respect to the global frame.
Two different types of markers are attached to a segment, namely anatomical markers and technical markers. Anatomical markers are placed at appropriate anatomical landmarks in order to define the segment and its coordinate system, thereby creating an anatomical frame (AF) for the segment (Wu et al., 2002). This allows for appropriate modeling of the segment in the motion capture software. Technical markers may be placed on the segment to assist the motion capture cameras in tracking the segment during a dynamic task, thereby creating a technical frame (TF) for the segment (Wu et al., 2002). The relationship between these two frames is established by the motion capture software using marker data collected during a static calibration trial prior to the performance of the task(s) of interest.

Efforts must be made to reduce measurement errors associated with motion capture. Camera arrangement and the target volume (i.e., the space within which data will be collected) must be appropriately set in order to accurately detect and track markers. The motion capture system must also be calibrated properly prior to data collection to reduce risk of systematic error from things such as optical distortion (Chiari et al., 2005). Post-collection filtering and signal smoothing decreases random error that may occur from anomalous marker movement, while algorithms can be used to fill gaps that occur in marker trajectories (Chiari et al., 2005).

Soft tissue artifact (STA) has been identified as the most substantial source of error during motion capture and can create inaccuracies that impact marker/segment reconstruction and subsequent movement analyses (Andriacchi et al., 2000). Compensation techniques can be used to reduce deformation and the motion of markers being used to identify the segment, which includes solidification, multiple anatomical landmark calibration, pliant surface modeling, dynamic calibration, point cluster, and global optimization. Further detail on different techniques can be explored in Leardini et al. (2005). These techniques are either specific to a single body
segment (Alexander et al., 2003) or globally to the entire limb (Lu et al., 1999) to optimize bone positioning estimation. Accurate placement of anatomical markers at appropriate landmarks and using technical markers that would be expected to undergo a minimum amount of STA have a significant influence on the reliability and accuracy of the resulting kinematic and kinetic calculations.

No gold standard for collecting motion capture data of the spine has been recognized to date. One approach is to use a rigid marker cluster attached to a baseplate, which is attached to the skin over the T12 or L1 spinous process, effectively modelling the lumbar spine as one segment (Schache et al., 2002). This variation may be limiting, as the upper and lower lumbar regions have been shown to have independent contributions to spinal motion, especially during sagittal plane movements (Alqhtani et al., 2016). One variation of this method uses three markers attached to a cross-shaped wand configuration attached to the centre of the baseplate. Schache et al. (2002) reported good within-day and between-day repeatability for the 3D angular time series calculated using this model. A second variation uses one marker attached to the end of a wand extending out from the centre of the baseplate and two markers attached to the lateral borders of the baseplate. Whittle et al. (1998) and Steele et al. (2014) reported on the within-day repeatability for their variations of this method. An alternate approach to modelling lumbar spine motion involves markers that are attached directly to the skin over the lumbar spinous processes and paraspinal muscles. Mason et al. (2016) reported good within-day and between-day repeatability for the 3D angular time series calculated for their variation of this method. Ryan et al. (2017) compared the 3D angular time series of a representative variation of each of these two approaches, and reported very strong cross-correlations and 1-2° root-mean-square differences between the two models for rotations in all three planes during gait.
2.6. Purpose

The primary purpose of this study was to examine the effects of progressive external load and support surface stability on trunk muscle activity and lumbar spine ROM in collegiate football players performing the back squat. The four primary research questions and associated hypotheses addressed by the study were:

1) What effect does progressive external load have on peak and average trunk muscle amplitudes during the back squat in collegiate football players?

   It was hypothesized that trunk muscle amplitudes would increase as load increased, most notably for the IO (due to its proposed importance for lumbopelvic stability) and paraspinal muscles (due to the requirement for a large internal extension moment).

2) What effect does support surface stability have on peak and average trunk muscle amplitudes during the back squat in collegiate football players?

   It was hypothesized that trunk muscle amplitudes would be higher for the IO and EO when squatting on an unstable surface compared to a stable surface due to the increased stability requirements resulting from the introduction of increased frontal plane instability.

3) What effect does progressive external load have on lumbar spine ROMs during the back squat in collegiate football players?

   It was hypothesized that lumbar extension ROM would increase as load increased (due to this being a compensation that has been observed in moderate/high loads during the back squat).

4) What effect does support surface stability have on lumbar spine ROMs during the back squat in collegiate football players?
It was hypothesized that lateral flexion ROM would increase when squatting on an unstable surface compared to a stable surface (due to the increased frontal plane stability requirements in a movement that generally does not challenge frontal plane stability to a great extent compared to the sagittal plane).

In addition to the primary research questions and hypotheses, the study also addressed the following secondary research questions, which were exploratory in nature and therefore did not have specific associated hypotheses:

1) What effect does progressive external load have on the relative timing of peak trunk muscle amplitudes during the back squat in collegiate football players?

2) What effect does support surface stability have on the relative timing of peak trunk muscle amplitudes during the back squat in collegiate football players?
3. Methods

3.1. Participants

An *a priori* analysis (G*Power v.3.1.9.2) indicated that 27 participants were required. This calculation was based on a moderate effect size (Cohen’s $d = 0.25$), an alpha level of 0.05, and a power level of 0.80 for a 3 (load) x 2 (stability) repeated measures factorial analysis of variance (ANOVA) design (Faul et al., 2007). Participants were recruited through the University of Regina Athlete Health and Performance Initiative (AHPi) via purposive sampling and convenience sampling. Paper advertisements were placed within the AHPi High Performance Centre explaining the purpose of the research, eligibility criteria, requirements, contact information, and any other pertinent information. The inclusion criteria were male collegiate football players aged 18 and older. The exclusion criteria were: acute low back injury within the past three months, chronic LBP within the past 6 months, identified and potentially harmful spinal deformities, body fat percentage above 15%, and an inability to communicate in English. Football players were selected to participate due to the high loading and unstable nature of their sport, which implies higher stability requirements and activation of stabilizing muscles, especially through the trunk and lumbopelvic complex. The body fat percentage requirement was used to improve the accuracy and validity of the motion capture and EMG data, and limited recruitment to quarterbacks, wide receivers, defensive backs, corner backs, outside linebackers, and running backs.
3.2. Research Design

This study followed a cross-sectional observational research design. Participants performed all testing conditions, with task order being randomized by support surface condition. Load progression stayed consistent within each support surface condition. Individuals who contacted the researchers received an explanation of the testing protocol and what was required of them. Prospective participants who provided consent underwent screening via a questionnaire to rule out any individuals who did not meet the eligibility criteria. Testing was conducted in the Faculty of Kinesiology and Health Studies Biomechanics Lab. Data collection proceeded following ethical approval from the University of Regina Research Ethics Board (see Appendix A), and all participants were provided with a copy of a Study Information Sheet (see Appendix B) and provided informed written consent (see Appendix C) prior to taking part in the study.

3.3. Procedure

Each participant’s back squat 1RM was determined under the supervision of an NSCA-CSCS at least 72 hours prior to data collection in the University of Regina AHPi High Performance Center. The 1RM testing protocol included two progressive warm up sets, with participants progressively increasing weight for sets of 1-2 reps until reaching a load that could be performed for only 1 repetition. No more than four working sets were performed to avoid excessive fatigue. Using the participant’s 1RM, the following testing loads to be used for data collection were calculated: 30% 1RM (very light), 50% 1RM (light,) and 70% 1RM (moderate).

Upon arriving in the lab for data collection, the study protocol was explained again. Participants wore compression shorts with no top to facilitate the attachment of the EMG electrodes and motion capture markers without obstruction. A full body dynamic warm up was
performed and supervised by a certified NSCA-CSCS to reduce risk of injury and facilitate proper movement. Surface EMG electrodes were attached bilaterally to the skin superficial to each target muscle (see 3.4.1). Reflective motion capture markers (9mm) were attached to each target segment (see 3.4.2). A Vicon T-Series motion capture system, utilizing six T10 (1MP) cameras and two V5 cameras (5 MP), was used to collect a static calibration trial with the participant in a regular standing posture and arms crossed across their chest at shoulder height. The same system was used to collect motion capture and EMG data during the subsequent dynamic trials. A squat stand was used to hold an Olympic barbell appropriately loaded for each trial. Bar placement was standardized over the upper trapezius. Participants performed two continuous repetitions with a short pause between repetitions of each load (30% followed by 50% followed by 70%) for the initial support surface condition (randomly allocated), with a 2-minute rest between each load to minimize the effects of fatigue. Before each repetition, participants were instructed to “brace” (abdominal brace) since this instruction is commonly used when the athletes perform the exercise in their regular training. Each repetition was performed with a consistent stance width to “parallel” depth (see Figure 2), whereby the line between the top of the knee and the crease of the hip was parallel with the ground (monitored by visual inspection), with participants initially performing the movement with an unloaded bar so that the depth requirements were understood. Following the completion of all repetitions at each load in the initial stability condition, the process was repeated in the second support surface condition following a 5-minute rest period. The stable support surface condition consisted of the participant performing the back squat while standing on the floor. The unstable support surface condition consisted of the participant performing the back squat while standing on two AIREX balance pads placed under the participant’s feet.
3.4. Data Collection

3.4.1. EMG

At the EMG electrode sites, the skin was shaved (if necessary), abraded, and sanitized with an alcohol swab. Using double-sided tape, eight silver 5x1mm Delsys Trigno double-differential wireless electrodes (Delsys Inc., Boston, MA, USA) were attached bilaterally to the skin over the lumbar paraspinal muscles (PS), RA, IO, and EO as follows:
• PS – At the level of the iliac crest, parallel to the spine, 2cm lateral to the spine over the muscle mass (Criswell, 2010). For the purposes of this thesis, the lumbar PS muscles include the ES, MFD, and semispinalis muscle groups.

• RA – 2cm lateral to midline, 1cm above the umbilicus over the muscle mass (Boccia et al., 2014; Dankaerts et al., 2004; Nelson-Wong et al., 2013; Ng et al., 1998).

• IO – 2cm medial to the anterior superior iliac spine (ASIS), medial/superior to the inguinal ligament (Boccia et al., 2014).

• EO – Just inferior to the rib cage at a location halfway between midline and mid-axillary line (Boccia et al., 2014; Dankaerts et al., 2004; Nelson-Wong et al., 2013; Ng et al., 1998).

The IO and EO placement sites were selected to strategically reduce crosstalk and subsequent error (Boccia et al., 2014). The EMG electrodes were pre-amplified (gain = 300V/V), and had an intra-electrode distance of 1cm, CMRR >80dB, and 20-450Hz hard-wired band pass filter. A sampling rate of 2000Hz was used; while this sampling rate is higher than what may be necessary, it allows for accurate recording of the raw signal without any risk of error associated with the sampling rate (Ives et al., 2003). The EMG system was digitally integrated with the motion capture system using a Delsys Trigger Module (Delsys Inc., Boston, MA, USA) to ensure synchronized recording of the EMG and motion capture data.

EMG signal amplitudes for each muscle were normalized using amplitude values obtained during resisted maximal voluntary isometric contractions ((MVIC). This method has shown to be effective in producing a high level of neuromuscular activity and reduce the risk of underestimation compared to other methods (Burden, 2010). The normalization tasks for the PS, RA, EO, and IO were:
• PS – Lying prone, with the lower limbs restrained on an assessment table and the torso off the edge of the table. The participant attempted to extend the torso past parallel with the table by lifting upward against manual resistance (Ng et al., 2002; Roth et al., 2016).

• RA – Lying supine, with the torso at a 45° angle relative to the table and the feet flat on the table and restrained. The participant attempted to flex the torso by lifting upward against manual resistance (Ng et al., 2002; Vera-Garcia et al., 2010).

• EO – Side-lying on the assessment table, with the lower limbs restrained. The participant attempted to laterally flex the torso by lifting upward against manual resistance (Vera-Garcia et al., 2010).

• IO – Side bridge position on floor mat. The participant attempted to maintain the position against manual resistance applied at the pelvis (Vera-Garcia et al., 2010).

The same individual applied the manual resistance used in the normalization tasks for all participants. Three repetitions (5 seconds) were performed for each normalization task with 1-minute rest intervals between repetitions (Dankaerts et al., 2004).

### 3.4.2. Motion Capture

Motion capture was used to measure the ROM of the upper lumbar spine, lower lumbar spine, pelvis, thighs, and lower legs. Using double-sided tape, markers were attached to various landmarks in order to define and track the motion of each segment. The sampling rate of the motion capture system was set at 100Hz.

The lumbar spine was separated into upper and lower segments, rather than defining the lumbar spine as one rigid segment (Ryan et al., 2017). A thermoplastic base plate was attached to the skin over the spinous processes of the T12 and L3 vertebrae, which were identified via the
palpation technique described by Snider et al., (2011). A marker attached to a 40mm wand extending out from the center of each plate and two retro-reflective markers attached to the lateral borders of each plate along its horizontal axis served as the anatomical markers for these segments. The wand marker and two bilateral markers attached to the skin 6cm lateral to the top edge of the plate served as technical markers for the segments.

At the pelvis, anatomical markers were attached over the most prominent points of the ASISs and posterior superior iliac spines (PSISs) bilaterally. Technical markers consisted of the PSIS markers and markers attached to the skin over the most superior point of the iliac crest bilaterally (Bruno, 2015).

Anatomical markers for each thigh were attached over the greater trochanter, lateral femoral epicondyle, and medial femoral epicondyle. Technical markers consisted of a cluster of four markers attached to molded thermoplastic plates, which were attached firmly to the lateral aspect of the thigh using a self-adhesive tensor bandage (Cappoz, 1995, 1996, 1997).

Lower leg (shank) anatomical markers were attached over the lateral femoral epicondyle, medial femoral epicondyle, medial malleolus, and lateral malleolus. Technical markers consisted of a cluster of four markers attached to molded thermoplastic plates, which were attached to the lateral aspect of the lower leg using Superwrap™ (FabriFoam Inc., Exton, PA, USA).

3.5 Data Analysis

Vicon Nexus software (Version 2.5, Vicon Motion Systems Ltd., Oxford, United Kingdom) was used to collect both motion capture and EMG data. Mathematical modelling and kinematic calculations involving the motion capture data, as well as all EMG data processing and
calculations, were performed using Visual3D Professional software (Version 6, C-Motion Inc., Germantown, MD, USA).

The initiation and termination of each repetition were determined visually using a graph of the hip flexion angle time series for the repetition. Initiation was set as the point at which the hip flexion angle increased beyond the baseline hip flexion angle (starting standing posture), while termination was set as the point at which the hip flexion angle returned to the baseline hip flexion angle. The mid-point of the repetition (i.e., end of the DSC and start of the ASC) was set as the point at which peak hip flexion angle was achieved between the initiation and termination of the repetition.

RMS values were calculated to quantify trunk muscle amplitudes during the squat trials and the normalization tasks. For each muscle, the peak amplitude achieved during the middle three seconds of each of its normalization task repetitions was determined and used to calculate an average value that was used to normalize the amplitudes achieved during the squat trials (%MVIC).

All segments were modelled as rigid bodies. A digital low-pass filter using a second-order, dual-pass Butterworth filter (cut-off frequency 6 Hz) was applied to the raw motion capture data (Winter, 2009). Using a geometrical joint coordinate system convention (Cole et al., 1993; Grood and Suntay, 1983), the 3D angular kinematic data for each segment was calculated using a XYZ Cardan rotation sequence (sagittal plane – frontal plane – transverse plane). Pelvic motion was defined as motion of the pelvis relative to the global coordinate system. Lower lumbar motion was defined as motion of the lower lumbar segment relative to the pelvis. Upper lumbar motion was defined as motion of the upper lumbar segment relative to the lower lumbar
segment. Thigh motion was defined as motion of the thigh segments relative to the pelvis. Lower leg motion was defined as motion of the lower leg segments relative to the thighs.

For the lumbar segments, virtual segments were created within the software to account for differences between participants and the relative orientation of the segments in their standing position. To do this, the origin of each segment was defined through the midpoint of the baseplate markers. The global co-ordinate system was used as reference to define the medial-lateral (x), anterior-posterior (y), and superior-inferior (z) axis orthogonal to the x and y-axis for each segment (Ryan et al., 2017). Through this process, segments with XY planes parallel to the laboratory floor were created, ensuring that rotations occurring during dynamic trials were calculated relative to the participant’s static standing posture.

A summary of the anatomical coordinate systems and movements of the lumbar segments (see Table 2), pelvis (see Table 3), thighs (see Table 4), and lower legs (see Table 5) are outlined below.
### Table 2
*Anatomical coordinate system of the lumbar spine. (Ryan et al., 2017; Wu et al., 2002)*

<table>
<thead>
<tr>
<th>Anatomical Coordinate System</th>
<th>Direction</th>
<th>Location</th>
<th>Movement</th>
<th>Plane of Movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Origin</td>
<td></td>
<td>Midpoint between baseplate markers</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x-axis</td>
<td>Medial-Lateral</td>
<td>Line passing through baseplate markers, positive to right</td>
<td>Flexion/extension (+ ext.)</td>
<td>Sagittal</td>
</tr>
<tr>
<td>y-axis</td>
<td>Anterior-Posterior</td>
<td>Perpendicular to x-axis, passing through baseplate, positive anteriorly</td>
<td>Lateral flexion (+ right lat. flex.)</td>
<td>Frontal</td>
</tr>
<tr>
<td>z-axis</td>
<td>Superior-Inferior</td>
<td>Perpendicular to z and y-axes, positive cephalad</td>
<td>Axial rotation (+ left rotation)</td>
<td>Transverse</td>
</tr>
</tbody>
</table>
### Table 3
Anatomical coordinate system of the pelvis. (Ryan et al., 2017; Wu et al., 2002)

<table>
<thead>
<tr>
<th>Anatomical Coordinate System</th>
<th>Direction</th>
<th>Location</th>
<th>Movement</th>
<th>Plane of Movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Origin</td>
<td></td>
<td>Midpoint between ASIS markers</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x-axis</td>
<td>Medial-Lateral</td>
<td>Line passing through ASIS markers, positive to right</td>
<td>Anterior/posterior tilt (+ post. tilt)</td>
<td>Sagittal</td>
</tr>
<tr>
<td>y-axis</td>
<td>Anterior-Posterior</td>
<td>Perpendicular to x-axis, midpoint of ASIS and PSIS markers, positive anteriorly</td>
<td>Upward movement of iliac crest (+ left side high)</td>
<td>Frontal</td>
</tr>
<tr>
<td>z-axis</td>
<td>Superior-Inferior</td>
<td>Perpendicular to z and y-axes, positive cephalad</td>
<td>Axial rotation (+ left rotation)</td>
<td>Transverse</td>
</tr>
<tr>
<td>Anatomical Coordinate System</td>
<td>Direction</td>
<td>Location</td>
<td>Movement</td>
<td>Plane of Movement</td>
</tr>
<tr>
<td>-----------------------------</td>
<td>-------------</td>
<td>---------------------------------------------------------------------------</td>
<td>---------------------------</td>
<td>-------------------</td>
</tr>
<tr>
<td>Origin</td>
<td></td>
<td>Mid-point between medial/lateral femoral epicondyle</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x-axis</td>
<td>Medial-Lateral</td>
<td>Lying in plane defined by knee markers and hip joint, perpendicular to z-axis, positive to the right</td>
<td>Flexion/extension (+ flex.)</td>
<td>Sagittal</td>
</tr>
<tr>
<td>y-axis</td>
<td>Anterior-Posterior</td>
<td>Line perpendicular to both z- and x-axis, positive anteriorly</td>
<td>Abduction/adduction (+ abduction)</td>
<td>Frontal</td>
</tr>
<tr>
<td>z-axis</td>
<td>Superior-Inferior</td>
<td>Line passing through origin and hip joint, positive proximally</td>
<td>External/internal rotation (+ ext. rot.)</td>
<td>Transverse</td>
</tr>
<tr>
<td>Anatomical Coordinate System</td>
<td>Direction</td>
<td>Location</td>
<td>Movement</td>
<td>Plane of Movement</td>
</tr>
<tr>
<td>------------------------------</td>
<td>----------------</td>
<td>---------------------------------------------------------------------------</td>
<td>-------------------------------</td>
<td>-------------------</td>
</tr>
<tr>
<td>Origin</td>
<td></td>
<td>Mid-point between medial and lateral malleolus markers</td>
<td></td>
<td></td>
</tr>
<tr>
<td>x-axis</td>
<td>Medial-Lateral</td>
<td>Lying in plane defined by malleolus markers and mid-point between medial and lateral epicondyle markers, perpendicular to z-axis, positive to right</td>
<td>Flexion/extension (+ flex.)</td>
<td>Sagittal</td>
</tr>
<tr>
<td>y-axis</td>
<td>Anterior-Posterior</td>
<td>Line perpendicular to both z- and x-axis, positive anteriorly</td>
<td>Abduction/adduction (+ abduction)</td>
<td>Frontal</td>
</tr>
<tr>
<td>z-axis</td>
<td>Superior-Inferior</td>
<td>Line passing through the origin and mid-point between medial and lateral knee epicondyle markers, positive proximally</td>
<td>External/internal rotation (+ ext. rot.)</td>
<td>Transverse</td>
</tr>
</tbody>
</table>
The hip flexion angle at which the peak amplitude (HIPemg_peak) of each muscle occurred was also identified and represented as a percentage of the total hip flexion ROM during both the DSC and ASC. For DSC, 0% was the starting hip flexion angle (HIPflex_start) and 100% was the peak hip flexion angle at the bottom of the squat (HIPflex_peak). For ASC, 100% was the peak hip flexion angle at the bottom of the squat (HIPflex_peak) and 0% was the end hip flexion angle (HIPflex_end). As a result, larger % values indicate HIPemg_peak occurred closer to the bottom of the squat for both the DSC and ASC, while smaller % values indicate HIPemg_peak occurred closer to the top of the squat for both the DSC and ASC. The following formulas were used to determine the relative (%) point in the hip flexion ROM that peak muscle activity occurred:

\[
\text{DSC: } \frac{(\text{HIPemg}_{\text{peak}} - \text{HIPflex}_{\text{start}})}{(\text{HIPflex}_{\text{peak}} - \text{HIPflex}_{\text{start}})} \times 100 \\
\text{ASC: } \frac{(\text{HIPemg}_{\text{peak}} - \text{HIPflex}_{\text{end}})}{(\text{HIPflex}_{\text{peak}} - \text{HIPflex}_{\text{end}})} \times 100
\]

### 3.6. Statistical Analysis

To address the primary research questions and hypotheses, the independent variables were load (30%, 50%, 70%), support surface (stable, unstable), and squat phase (DSC, ASC). The dependent variables were normalized peak and average trunk muscle amplitudes, and upper and lower lumbar ROMs. Separate 3 (load) x 2 (support surface) repeated-measures factorial ANOVAs with post-hoc testing were used to compare each of the dependent variables for DSC and ASC between load and support surface conditions. Bonferroni correction was used for post-hoc testing unless the F-test was significant without significant between-condition comparisons, in which case a least significant difference (LSD) correction was used.

To address the secondary research questions, the independent variables were load (30%, 50%, 70%), support surface (stable, unstable), and squat phase (DSC, ASC). The dependent
variable was the relative timing of peak trunk muscle amplitude (HIPemgpeak). Separate 3 (load) x 2 (support surface), and 2 (support surface) x 2 (squat phase), repeated-measures factorial ANOVA with post-hoc testing (Bonferroni or LSD) was used to compare the dependent variable between the load, support surface, and squat phase conditions.

Statistical significance ($\alpha$) was set at 0.05 for all analyses.
4. Results

4.1. Participants

Twenty-six individuals participated in the study (Age: 19.7 ± 2.2 years; Height: 181.2 ± 6.2 cm; Mass: 85.6 ± 8.5 kg). Average back squat 1RM and testing loads are presented in Table 6.

Table 6
Mean ± SD back squat 1RM and testing loads (kg).

<table>
<thead>
<tr>
<th>Load</th>
<th>30%</th>
<th>50%</th>
<th>70%</th>
<th>1RM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean ± SD</td>
<td>108.3 ± 18.4</td>
<td>181.3 ± 31.1</td>
<td>252.8 ± 44.2</td>
<td>361.1 ± 63.1</td>
</tr>
</tbody>
</table>

4.2. Trunk Muscle Amplitudes

When comparing the differences in trunk muscle amplitudes between load (30%, 50%, 70%) and support surface (stable, unstable) conditions, there was a significant main effect for load for PSavgDSC (70% > 50% > 30%), PSavgASC (70% > 50% > 30%), PSmaxDSC (70% > 50% > 30%), PSmaxASC (70% > 50% > 30%), RAaveDSC (70% > 50% > 30%), RAaveASC (70% > 50% > 30%), RAmaxASC (70% > 50% > 30%), EOaveASC (70% > 50% > 30%), EOmaxASC (70% > 30%), IOaveDSC (70% > 50% > 30%), IOaveASC (70% > 50% > 30%), and IOmaxASC (70% > 50% > 30%) (see Tables 7 & 8). Greater and more consistent increases in amplitude with increasing load were observed in the PS compared to other muscle groups during both phases of the squat. Increases in amplitude were generally less or non-significant from 50-70% 1RM, with the exception of the PS, RAavgDSC, RAmaxASC, EOavgASC, and IOavgDSC. There were no significant main effects for support surface, and no significant load x support surface interaction effects (see Tables 7 & 8).
When comparing differences in trunk muscle amplitudes between phase (DSC, ASC) and support surface (stable, unstable) conditions, there was a significant main effect for movement phase for PSavg (all loads: ASC > DSC), PSmax (all loads: ASC > DSC), RAavg (all loads: ASC > DSC), RAmax (all loads: ASC > DSC), EOmax (30%: ASC > DSC), and IO (all loads: ASC > DSC) (see Tables 7 & 9). The RA and PS were the only muscle groups showing significant differences between phases at all loads for both EMGavg and EMGmax. The IO showed significant increases in EMGavg (but not EMGmax) at all loads. There were no significant main effects for support surface, and no significant phase x support surface interaction effects (see Tables 7 & 9).
Table 7
Mean ± SD trunk muscle amplitudes (% MVIC) during the descent and ascent phases of the back squat for the three load (30%, 50%, 70%) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Phase</th>
<th>S30</th>
<th>S50</th>
<th>S70</th>
<th>U30</th>
<th>U50</th>
<th>U70</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>ASC</td>
<td>42.82 ± 24.21</td>
<td>51.27 ± 17.41</td>
<td>62.89 ± 19.89</td>
<td>39.43 ± 13.50</td>
<td>51.97 ± 16.94</td>
</tr>
<tr>
<td>MAX</td>
<td>DSC</td>
<td>66.33 ± 18.30</td>
<td>77.26 ± 23.02</td>
<td>90.28 ± 23.80</td>
<td>61.87 ± 22.51</td>
<td>82.31 ± 22.86</td>
<td>93.62 ± 20.88</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>86.59 ± 27.47</td>
<td>100.34 ± 32.30</td>
<td>115.43 ± 40.13</td>
<td>81.68 ± 31.20</td>
<td>100.42 ± 21.04</td>
<td>118.06 ± 33.97</td>
</tr>
<tr>
<td>RA</td>
<td>AVG</td>
<td>DSC</td>
<td>2.79 ± 2.23</td>
<td>3.46 ± 2.50</td>
<td>3.90 ± 2.78</td>
<td>2.82 ± 2.07</td>
<td>3.11 ± 2.16</td>
</tr>
<tr>
<td></td>
<td></td>
<td>ASC</td>
<td>6.08 ± 4.45</td>
<td>7.02 ± 4.56</td>
<td>8.18 ± 4.85</td>
<td>5.75 ± 3.96</td>
<td>7.25 ± 5.13</td>
</tr>
<tr>
<td>MAX</td>
<td>DSC</td>
<td>6.83 ± 5.61</td>
<td>9.14 ± 7.00</td>
<td>11.03 ± 12.50</td>
<td>7.95 ± 7.09</td>
<td>10.01 ± 9.69</td>
<td>8.83 ± 6.47</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>18.13 ± 13.41</td>
<td>23.70 ± 15.11</td>
<td>31.47 ± 19.94</td>
<td>16.72 ± 9.49</td>
<td>22.86 ± 16.40</td>
<td>32.55 ± 22.75</td>
</tr>
<tr>
<td>MAX</td>
<td>DSC</td>
<td>35.95 ± 26.01</td>
<td>30.03 ± 17.39</td>
<td>34.64 ± 24.37</td>
<td>36.23 ± 41.17</td>
<td>34.34 ± 30.29</td>
<td>30.93 ± 19.68</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>30.06 ± 20.18</td>
<td>32.90 ± 19.91</td>
<td>38.91 ± 21.88</td>
<td>25.24 ± 12.15</td>
<td>29.50 ± 15.36</td>
<td>34.97 ± 22.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td>ASC</td>
<td>19.21 ± 17.19</td>
<td>22.70 ± 19.52</td>
<td>28.35 ± 28.06</td>
<td>18.68 ± 14.34</td>
<td>22.55 ± 16.90</td>
</tr>
<tr>
<td>MAX</td>
<td>DSC</td>
<td>41.43 ± 38.13</td>
<td>45.07 ± 32.26</td>
<td>51.23 ± 38.14</td>
<td>45.45 ± 35.95</td>
<td>45.01 ± 40.40</td>
<td>48.73 ± 29.56</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>46.08 ± 47.51</td>
<td>59.77 ± 54.01</td>
<td>74.55 ± 77.72</td>
<td>45.50 ± 39.08</td>
<td>57.97 ± 50.40</td>
<td>75.64 ± 78.25</td>
</tr>
</tbody>
</table>

Phase: DSC: descent, ASC: ascent.
Load & Support Surface: S30: 30% stable, S50: 50% stable, S70: 70% stable, U30: 30% unstable, U50: 50% unstable, U70: 70% unstable.
Table 8
Statistical analysis results comparing the differences in muscle amplitude between the three load (30%, 50%, 70%) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Phase</th>
<th>Load</th>
<th>Surface</th>
<th>Load x Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>PS</td>
<td>AVG</td>
<td>DSC</td>
<td>*F(1.14,26.14)=65.79, p&lt;0.0001</td>
<td>F(1.23)=0.559, p=0.462</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70, 50-70: p&lt;0.0001</td>
<td>*F(1.1,25.306)=1.16, p=0.299</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>F(1.22,27.97)=65.32, p&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70, 50-70: p&lt;0.0001</td>
<td>*F(1.28,29.42)=1.56, p=0.299</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>MAX</td>
<td>*F(1.41,31.11)=61.23, p&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70, 50-70: p&lt;0.0001</td>
<td>F(1,23)=0.525, p=.476</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>*F(1.43,31.41)=33.20, p&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70, 50-70: p&lt;0.0001</td>
<td>F(1,22)=0.077, p=0.784</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>MAX</td>
<td>*F(1.58,33.14)=12.20, p&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70: p&lt;0.005; 30-70: p&lt;0.005</td>
<td>F(1,21)=0.063, p=0.804</td>
</tr>
<tr>
<td>RA</td>
<td>AVG</td>
<td>DSC</td>
<td>F(1.58,33.14)=12.20, p&lt;0.0001</td>
<td>F(2,44)=1.43, p=0.251</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70, 50-70: p&lt;0.005</td>
<td>*F(1.43,31.41)=33.20, p&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>F(1.41,29.70)=7.70, p&lt;0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70: p&lt;0.05; 50-70: p&lt;0.275</td>
<td>*F(1.57,33.01)=0.309, p=0.683</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>MAX</td>
<td>*F(1.52,32.01)=2.38, p=0.121</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>*F(1.36,28.64)=22.86, p&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70: p&lt;0.001; 50-70: p&lt;0.01</td>
<td>F(2,44)=1.875, p=0.082</td>
</tr>
<tr>
<td>EO</td>
<td>AVG</td>
<td>DSC</td>
<td>*F(1.08,25.94)=0.81, p=0.385</td>
<td>F(2,44)=0.327, p=0.668</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>*F(1.30,31.15)=7.476, p&lt;0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-70: p&lt;0.05; 50-70: p&lt;0.005; 30-50: p=0.437</td>
<td>F(1.47,35.31)=0.91, p=0.384</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>MAX</td>
<td>*F(1.46,34.97)=0.70, p=0.502</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>F(2.48)=5.378, p&lt;0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50: p=0.475; 30-70: p&lt;0.005; 50-70: p=0.179</td>
<td>F(2,44)=0.064, p=0.938</td>
</tr>
<tr>
<td>IO</td>
<td>AVG</td>
<td>DSC</td>
<td>*F(1.25,29.95)=4.275, p&lt;0.05</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50: p=0.123; 30-70, 50-70: p&lt;0.05 **</td>
<td>F(1.24)=1.01, p=0.326</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>*F(1,09,26.25)=4.75, p&lt;0.05</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50, 30-70: p&lt;0.05; 50-70: p=0.076 **</td>
<td>F(1.24)=0.174, p=0.68</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>MAX</td>
<td>*F(1.61,36.91)=2.34, p=0.108</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>ASC</td>
<td>*F(1.10,25.23)=6.692, p&lt;0.05</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>30-50: p=0.01; 30-70: p&lt;0.05; 50-70: p=0.151</td>
<td>F(1.23)=0.035, p=0.853</td>
</tr>
</tbody>
</table>


Phase: DSC: descent, ASC: ascent.

Load: 30-50: 30% vs. 50%, 30-70: 30% vs. 70%, 50-70: 50% vs. 70%.

* Greenhouse-Geisser correction used to calculate the F statistic (assumption of sphericity violated).

** Least significant difference (LSD) test used for pairwise comparisons.
Table 9
Statistical analysis results comparing the differences in muscle amplitude between the two movement phases (descent, ascent) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Load</th>
<th>Phase</th>
<th>Surface</th>
<th>Phase x Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>PS</td>
<td>AVG</td>
<td>30</td>
<td>F(1, 23)=42.39, p&lt;0.0001</td>
<td>F(1, 23)=1.40, p=0.250</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,24)=34.21, p&lt;0.0001</td>
<td>F(1,24)=0.02, p=0.882</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,24)=26.42, p&lt;0.0001</td>
<td>F(1,24)=0.358, p=0.555</td>
</tr>
<tr>
<td>MAX</td>
<td>AVG</td>
<td>30</td>
<td>F(1,22)=30.11, p&lt;0.0001</td>
<td>F(1,22)=2.05, p=0.166</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,22)=29.21, p&lt;0.0001</td>
<td>F(1,22)=1.006, p=0.327</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,24)=23.51, p&lt;0.0001</td>
<td>F(1,24)=2.024, p=0.168</td>
</tr>
<tr>
<td>RA</td>
<td>AVG</td>
<td>30</td>
<td>F(1,21)=24.84, p&lt;0.0001</td>
<td>F(1,21)=0.389, p=0.540</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,21)=31.05, p&lt;0.0001</td>
<td>F(1,21)=0.034, p=0.855</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,21)=30.10, p&lt;0.0001</td>
<td>F(1,21)=0.015, p=0.903</td>
</tr>
<tr>
<td>MAX</td>
<td>AVG</td>
<td>30</td>
<td>F(1,21)=29.65, p&lt;0.0001</td>
<td>F(1,21)=0.014, p=0.908</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,21)=22.70, p&lt;0.0001</td>
<td>F(1,21)=0.0001, p=0.989</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,21)=29.68, p&lt;0.0001</td>
<td>F(1,21)=0.07, p=0.799</td>
</tr>
<tr>
<td>EO</td>
<td>AVG</td>
<td>30</td>
<td>F(1,24)=0.54, p=0.471</td>
<td>F(1,24)=0.664, p=0.423</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,24)=0.26, p=0.617</td>
<td>F(1,24)=0.28, p=0.603</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,24)=3.58, p=0.07</td>
<td>F(1,24)=1.16, p=0.292</td>
</tr>
<tr>
<td>MAX</td>
<td>AVG</td>
<td>30</td>
<td>F(1,24)=4.76, p&lt;0.05</td>
<td>F(1,24)=0.33, p=0.571</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,24)=0.10, p=0.755</td>
<td>F(1,24)=0.02, p=0.881</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,24)=1.12, p=0.301</td>
<td>F(1,24)=2.62, p=0.118</td>
</tr>
<tr>
<td>IO</td>
<td>AVG</td>
<td>30</td>
<td>F(1,24)=5.63, p&lt;0.05</td>
<td>F(1,24)=0.07, p=0.792</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,24)=5.52, p&lt;0.05</td>
<td>F(1,24)=0.03, p=0.876</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,24)=4.39, p&lt;0.05</td>
<td>F(1,24)=0.08, p=0.787</td>
</tr>
<tr>
<td>MAX</td>
<td>AVG</td>
<td>30</td>
<td>F(1,24)=3.46, p=0.562</td>
<td>F(1,24)=0.05, p=0.833</td>
</tr>
<tr>
<td></td>
<td></td>
<td>50</td>
<td>F(1,22)=1.808, p=0.192</td>
<td>F(1,22)=0.033, p=0.857</td>
</tr>
<tr>
<td></td>
<td></td>
<td>70</td>
<td>F(1,24)=3.482, p=0.074</td>
<td>F(1,24)=0.234, p=0.633</td>
</tr>
</tbody>
</table>


Load: 30: 30%, 50: 50%, 70: 70%.
4.3. Lumbar Spine ROMs

When comparing the differences in upper lumbar and lower lumbar ROMs between load (30%, 50%, 70%) and support surface (stable, unstable) conditions, there was a significant main effect for load for ULUM-FLEXASC (30/50% > 70%), ULUM-LLFASC (50% > 30%), LLUM-EXTDSC (70% > 30%), LLUM-FLEXDSC (30% > 50% > 70%), LLUM-FLEXASC (30% > 50% > 70%), and LLUM-LRASC (50% > 70%) (see Tables 10-11). There were no main effects for support surface and no significant load x support surface interaction effects (see Tables 10-11).

When comparing differences in upper lumbar and lower lumbar ROMs between movement phases (DSC, ASC) and support surface (stable, unstable) conditions, there was a significant main effect for movement phase for LLUM-EXT (all loads: DSC > ASC), LLUM-LR (70%: DSC > ASC), and LLUM-RLF (30%: DSC > ASC) (see Tables 10 & 12). There was also a significant main effect for support surface for LLUM-RLF (70%: unstable > stable) (see Tables 10 & 12). Finally, there was a significant phase x support surface interaction effect for ULUM-EXT (30%) (see Tables 10 & 12). The significant interaction effect for ULUM-EXT (30%) was related to its decreasing from DSC to ASC during the stable conditions, but increasing from DSC to ASC during the unstable conditions.
Table 10
*Mean ± SD lumbar ranges of motion (degrees) during the descent and ascent phases of the back squat for the three load (30%, 50%, 70%) and two support surface (stable, unstable) conditions.*

<table>
<thead>
<tr>
<th>Movement</th>
<th>Phase</th>
<th>S30</th>
<th>S50</th>
<th>S70</th>
<th>U30</th>
<th>U50</th>
<th>U70</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ASC</td>
<td>-2.59 ± 4.85</td>
<td>-2.42 ± 4.89</td>
<td>-2.30 ± 4.88</td>
<td>-2.51 ± 4.91</td>
<td>-2.35 ± 4.87</td>
<td>-2.35 ± 4.87</td>
</tr>
<tr>
<td>LR</td>
<td>DSC</td>
<td>1.54 ± 3.24</td>
<td>1.52 ± 3.25</td>
<td>1.50 ± 3.25</td>
<td>1.61 ± 3.28</td>
<td>1.61 ± 3.23</td>
<td>1.61 ± 3.26</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>1.51 ± 3.29</td>
<td>1.51 ± 3.39</td>
<td>1.59 ± 3.43</td>
<td>1.48 ± 3.19</td>
<td>1.67 ± 3.51</td>
<td>1.60 ± 3.40</td>
</tr>
<tr>
<td>RLF</td>
<td>DSC</td>
<td>4.02 ± 4.82</td>
<td>4.00 ± 4.94</td>
<td>3.90 ± 4.84</td>
<td>3.95 ± 4.88</td>
<td>4.01 ± 4.82</td>
<td>3.92 ± 4.88</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>3.89 ± 4.75</td>
<td>3.90 ± 4.91</td>
<td>3.91 ± 4.81</td>
<td>3.87 ± 4.81</td>
<td>3.93 ± 4.84</td>
<td>3.96 ± 4.82</td>
</tr>
<tr>
<td>RR</td>
<td>DSC</td>
<td>0.05 ± 3.37</td>
<td>-0.08 ± 0.31</td>
<td>-0.20 ± 3.33</td>
<td>-0.08 ± 3.33</td>
<td>-0.12 ± 3.29</td>
<td>-0.20 ± 3.30</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>0.37 ± 3.11</td>
<td>0.12 ± 3.28</td>
<td>0.14 ± 3.32</td>
<td>0.1292 ± 3.28</td>
<td>0.16 ± 3.36</td>
<td>0.19 ± 3.44</td>
</tr>
</tbody>
</table>

<p>| FLEX     | DSC   | 17.52 ± 12.84 | 15.76 ± 12.82 | 14.29 ± 12.36 | 17.33 ± 12.81 | 15.86 ± 12.29 | 14.10 ± 11.90 |
|          | ASC   | 18.74 ± 10.79 | 17.12 ± 10.76 | 15.51 ± 10.31 | 18.49 ± 10.42 | 17.18 ± 10.08 | 15.23 ± 9.92 |
| LLF      | DSC   | 2.84 ± 1.96 | 2.86 ± 1.95 | 2.90 ± 2.06 | 2.90 ± 2.06 | 2.88 ± 1.91 | 2.79 ± 2.01 |
|          | ASC   | 2.84 ± 1.93 | 2.93 ± 2.02 | 2.86 ± 2.15 | 2.84 ± 1.97 | 2.85 ± 1.95 | 2.85 ± 2.11 |
| LR       | DSC   | 0.65 ± 2.43 | 0.73 ± 2.38 | 0.43 ± 2.53 | 0.54 ± 2.51 | 0.57 ± 2.42 | 0.44 ± 2.41 |</p>
<table>
<thead>
<tr>
<th></th>
<th>ASC</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>RLF</td>
<td>DSC</td>
<td>-0.10</td>
<td>-1.13</td>
<td>-1.24</td>
<td>-1.07</td>
<td>-1.09</td>
</tr>
<tr>
<td>ASC</td>
<td>DSC</td>
<td>-1.22</td>
<td>-1.24</td>
<td>-1.37</td>
<td>-1.19</td>
<td>-1.18</td>
</tr>
<tr>
<td>RR</td>
<td>DSC</td>
<td>1.72</td>
<td>1.71</td>
<td>1.82</td>
<td>1.72</td>
<td>1.61</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td>1.68</td>
<td>1.51</td>
<td>1.64</td>
<td>1.52</td>
<td>1.53</td>
</tr>
</tbody>
</table>


**Phase:** DSC: descent, ASC: ascent.

**Load & Support Surface:** S30: 30% stable, S50: 50% stable, S70: 70% stable, U30: 30% unstable, U50: 50% unstable, U70: 70% unstable.

**Note:** Segment/joint angles are calculated relative to those measured during a neutral standing posture.
Table 11
Statistical analysis results comparing the differences in lumbar ranges of motion between the three load (30%, 50%, 70%) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Movement</th>
<th>Phase</th>
<th>Load</th>
<th>Surface</th>
<th>Load x Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>ULUM</td>
<td>EXT</td>
<td>DSC</td>
<td>*F(1.53,38.29)=0.273, p=0.762</td>
<td>F(1.25)=0.581, p=0.453</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>*F(1.37,34.21)=0.252, p=0.778</td>
<td>F(1.25)=0.004, p=0.953</td>
</tr>
<tr>
<td>FLEX</td>
<td>DSC</td>
<td></td>
<td>*F(1.49,37.19)=0.137, p=0.122</td>
<td>F(1.25)=0.618, p=0.439</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>30-50 = p=0.971; 30-70, 50-70: p&lt;0.05 **</td>
<td>F(1.25)=1.298, p=0.265</td>
</tr>
<tr>
<td>LLF</td>
<td>DSC</td>
<td></td>
<td>F(2.50)=1.017, p=0.369</td>
<td>F(1.25)=0.103, p=0.751</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>F(1.31,32.82)=4.444, p&lt;0.05</td>
<td>F(1.25)=0.201, p=0.657</td>
</tr>
<tr>
<td>LR</td>
<td>DSC</td>
<td></td>
<td>*F(1.32,32.94)=0.039, p=0.962</td>
<td>F(1.25)=1.253, p=0.274</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>F(2.50)=0.649, p=0.527</td>
<td>F(1.25)=0.477, p=0.496</td>
</tr>
<tr>
<td>RLF</td>
<td>DSC</td>
<td></td>
<td>*F(1.47,36.83)=0.797, p=0.423</td>
<td>F(1.25)=0.075, p=0.786</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>*F(1.53,38.13)=0.154, p=0.800</td>
<td>F(1.25)=0.074, p=0.788</td>
</tr>
<tr>
<td>RR</td>
<td>DSC</td>
<td></td>
<td>*F(1.52,38.11)=3.046, p=0.072</td>
<td>F(1.25)=0.511, p=0.481</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>*F(1.22,30.46)=0.445, p=0.644</td>
<td>F(1.25)=0.332, p=0.570</td>
</tr>
<tr>
<td>LLUM</td>
<td>EXT</td>
<td>DSC</td>
<td>*F(1.54,38.57)=5.001, p&lt;0.05</td>
<td>F(1.25)=1.804, p=0.191</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>30-50: p=0.200; 30-70: p&lt;0.05; 50-70: p=0.441</td>
<td>F(1.25)=3.917, p=0.059</td>
</tr>
<tr>
<td>FLEX</td>
<td>DSC</td>
<td></td>
<td>*F(1.49,37.31)=40.885, p&lt;0.0001</td>
<td>F(1.25)=0.097, p=0.757</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>30-50, 30-70, 50-70: p&lt;0.0001</td>
<td>F(1.25)=0.244, p=0.626</td>
</tr>
<tr>
<td>LLF</td>
<td>DSC</td>
<td></td>
<td>*F(1.63,40.75)=0.027, p=0.973</td>
<td>F(1.25)=0.404, p=0.531</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td></td>
<td>F(2.50)=0.231, p=0.795</td>
<td>F(1.25)=0.276, p=0.604</td>
</tr>
<tr>
<td>LR</td>
<td>DSC</td>
<td></td>
<td>F(2.50)=5.294, p&lt;0.01</td>
<td>F(1.25)=1.809, p=0.191</td>
</tr>
</tbody>
</table>
30-50: p=1.0; 30-70: p=0.135; 50-70: p<0.01  
p=0.191  
p=0.236

<table>
<thead>
<tr>
<th>Movement</th>
<th>Ascension (ASC)</th>
<th>Right Lateral Flexion (RLF)</th>
<th>Right Rotation (RR)</th>
</tr>
</thead>
<tbody>
<tr>
<td>DSC</td>
<td>F(2,50) = 3.070, p=0.055</td>
<td>F(2,50) = 2.670, p=0.079</td>
<td>F(2,50) = 2.670, p=0.079</td>
</tr>
<tr>
<td>ASC</td>
<td>F(2,50) = 3.070, p=0.055</td>
<td>F(2,50) = 2.670, p=0.079</td>
<td>F(2,50) = 2.670, p=0.079</td>
</tr>
<tr>
<td>RLF</td>
<td>DSC</td>
<td>ASC</td>
<td>RLF</td>
</tr>
<tr>
<td>ASC</td>
<td>F(1.52, 38.11) = 0.481, p=0.621</td>
<td>F(1.52, 38.11) = 0.481, p=0.621</td>
<td>F(1.52, 38.11) = 0.481, p=0.621</td>
</tr>
<tr>
<td>RR</td>
<td>ASC</td>
<td>F(1.35, 33.67) = 0.692, p=0.453</td>
<td>F(1.35, 33.67) = 0.692, p=0.453</td>
</tr>
<tr>
<td>ASC</td>
<td>F(1.52, 37.96) = 0.555, p=0.531</td>
<td>F(1.52, 37.96) = 0.555, p=0.531</td>
<td>F(1.52, 37.96) = 0.555, p=0.531</td>
</tr>
</tbody>
</table>


**Phase:** DSC: descent, ASC: ascent.

**Load:** 30-50: 30% vs. 50%, 30-70: 30% vs. 70%, 50-70: 50% vs. 70%.

* Greenhouse-Geisser correction used to calculate the F statistic (assumption of sphericity violated).

** Least significant difference (LSD) test used for pairwise comparisons.
Table 12
Statistical analysis results comparing the differences in lumbar ranges of motion between the two movement phases (descent, ascent) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Movement</th>
<th>Load</th>
<th>Phase</th>
<th>Surface</th>
<th>Phase x Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>ULUM EXT</td>
<td>30</td>
<td>F(1,25)=0.356, p=0.556</td>
<td>F(1,25)=0.013, p=0.911</td>
<td>F(1,25)=10.178, p&lt;0.005</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.396, p=0.535</td>
<td>F(1,25)=0.085, p=0.774</td>
<td>F(1,25)=0.658, p=0.425</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=0.315, p=0.579</td>
<td>F(1,25)=0.300, p=0.589</td>
<td>F(1,25)=0.042, p=0.839</td>
</tr>
<tr>
<td>FLEX</td>
<td>30</td>
<td>F(1,25)=4.024, p=0.056</td>
<td>F(1,25)=0.440, p=0.513</td>
<td>F(1,25)=0.270, p=0.608</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.965, p=0.335</td>
<td>F(1,25)=1.162, p=0.291</td>
<td>F(1,25)=0.721, p=0.404</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=5.417, p=0.028</td>
<td>F(1,25)=0.248, p=0.623</td>
<td>F(1,25)=0.269, p=0.609</td>
</tr>
<tr>
<td>LLF</td>
<td>30</td>
<td>F(1,25)=0.013, p=0.912</td>
<td>F(1,25)=0.262, p=0.613</td>
<td>F(1,25)=0.724, p=0.403</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.828, p=0.372</td>
<td>F(1,25)=0.220, p=0.643</td>
<td>F(1,25)=0.294, p=0.593</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=3.383, p=0.078</td>
<td>F(1,25)=0.079, p=0.782</td>
<td>F(1,25)=0.531, p=0.473</td>
</tr>
<tr>
<td>LR</td>
<td>30</td>
<td>F(1,25)=1.292, p=0.266</td>
<td>F(1,25)=0.038, p=0.847</td>
<td>F(1,25)=1.112, p=0.302</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.047, p=0.830</td>
<td>F(1,25)=2.073, p=0.162</td>
<td>F(1,25)=0.337, p=0.567</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=1.148, p=0.703</td>
<td>F(1,25)=0.628, p=0.435</td>
<td>F(1,25)=1.367, p=0.253</td>
</tr>
<tr>
<td>RLF</td>
<td>30</td>
<td>F(1,25)=3.220, p=0.085</td>
<td>F(1,25)=0.272, p=0.607</td>
<td>F(1,25)=0.425, p=0.521</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=2.096, p=0.160</td>
<td>F(1,25)=0.034, p=0.856</td>
<td>F(1,25)=0.071, p=0.791</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=0.048, p=0.828</td>
<td>F(1,25)=0.293, p=0.593</td>
<td>F(1,25)=0.212, p=0.649</td>
</tr>
<tr>
<td>RR</td>
<td>30</td>
<td>F(1,25)=0.230, p=0.636</td>
<td>F(1,25)=2.686, p=0.114</td>
<td>F(1,25)=2.157, p=0.154</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.152, p=0.700</td>
<td>F(1,25)=0.174, p=0.680</td>
<td>F(1,25)=0.001, p=0.998</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=0.214, p=0.648</td>
<td>F(1,25)=0.194, p=0.664</td>
<td>F(1,25)=0.441, p=0.513</td>
</tr>
<tr>
<td>LLUM EXT</td>
<td>30</td>
<td>F(1,25)=14.219, p&lt;0.005</td>
<td>F(1,25)=2.473, p=0.128</td>
<td>F(1,25)=0.220, p=0.643</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=17.579, p&lt;0.0001</td>
<td>F(1,25)=2.048, p=0.165</td>
<td>F(1,25)=0.004, p=0.953</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=24.004, p&lt;0.0001</td>
<td>F(1,25)=1.007, p=0.325</td>
<td>F(1,25)=1.001, p=0.327</td>
</tr>
<tr>
<td>FLEX</td>
<td>30</td>
<td>F(1,25)=0.912, p=0.349</td>
<td>F(1,25)=0.255, p=0.618</td>
<td>F(1,25)=0.393, p=0.536</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=1.148, p=0.294</td>
<td>F(1,25)=0.030, p=0.863</td>
<td>F(1,25)=0.056, p=0.814</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=0.973, p=0.334</td>
<td>F(1,25)=0.283, p=0.599</td>
<td>F(1,25)=0.507, p=0.483</td>
</tr>
<tr>
<td>LLF</td>
<td>30</td>
<td>F(1,25)=0.051, p=0.823</td>
<td>F(1,25)=0.051, p=0.823</td>
<td>F(1,25)=0.215, p=0.647</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.394, p=0.536</td>
<td>F(1,25)=0.452, p=0.507</td>
<td>F(1,25)=1.100, p=0.304</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=0.013, p=0.910</td>
<td>F(1,25)=1.764, p=0.196</td>
<td>F(1,25)=1.914, p=0.179</td>
</tr>
<tr>
<td>Movement</td>
<td>Load</td>
<td>F(1,25)</td>
<td>p-value</td>
<td>F(1,25)</td>
</tr>
<tr>
<td>----------</td>
<td>------</td>
<td>----------</td>
<td>---------</td>
<td>----------</td>
</tr>
<tr>
<td>LR</td>
<td>30</td>
<td>7.277</td>
<td>0.012</td>
<td>0.110</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>1.732</td>
<td>0.200</td>
<td>2.880</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>24.004</td>
<td>0.001</td>
<td>1.007</td>
</tr>
<tr>
<td>RLF</td>
<td>30</td>
<td>4.318</td>
<td>&lt;0.05</td>
<td>0.169</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>0.946</td>
<td>0.340</td>
<td>0.347</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>1.001</td>
<td>0.327</td>
<td>4.683</td>
</tr>
<tr>
<td>RR</td>
<td>30</td>
<td>1.196</td>
<td>0.285</td>
<td>0.720</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>1.672</td>
<td>0.208</td>
<td>0.324</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>3.435</td>
<td>0.076</td>
<td>1.777</td>
</tr>
</tbody>
</table>


**Load**: 30: 30%, 50: 50%, 70: 70%.
4.4. Relative Timing of Peak Trunk Muscle Amplitude

When comparing the differences in the relative timing of peak trunk muscle amplitude between load (30%, 50%, 70%) and support surface (stable, unstable) conditions, there was a significant main effect for load for PS_{DSC} (30% > 70%), EO_{DSC} (30% > 70%), and EO_{ASC} (30% > 70%) (see Tables 13 & 14). In other words, peak amplitude of the PS and EO occurred closer to the bottom of the squat during DSC at 30% load compared to 70% load, and peak amplitude of the EO occurred closer to the bottom of the squat during ASC at 30% compared to 70% load (see Tables 13 & 14). There were no significant main effects for support surface; however, there was a significant load x support surface interaction effect for EO_{ASC} (see Tables 13 & 14). The significant interaction effect was related to the timing of peak EO activity progressively decreasing as load increased in the stable condition, but increasing from 30% to 50% and decreasing from 50% to 70% in the unstable condition.

When comparing differences in the relative timing of peak trunk muscle amplitude between movement phases (DSC, ASC) and support surface (stable, unstable) conditions, there was a significant main effect for movement phase for RA (all loads: DSC > ASC) and EO (all loads: DSC > ASC) (see Tables 13 & 15). In other words, peak amplitude of the RA and EO occurred closer to the bottom of the squat during DSC compared to ASC for all loads. There was also a significant main effect for support surface for EO (30%: stable > unstable) (see Tables 13 & 15). In other words, peak amplitude of the EO occurred closer to the bottom of the squat during the stable condition compared to the unstable condition at 30% load. There were no significant phase x support surface interaction effects (see Tables 13 & 15).
Table 13
Mean ± SD relative point in the hip flexion ROM (%) that peak trunk muscle amplitude occurred during the descent and ascent phases of the back squat for the three load (30%, 50%, 70%) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Phase</th>
<th>S30</th>
<th>S50</th>
<th>S70</th>
<th>U30</th>
<th>U50</th>
<th>U70</th>
</tr>
</thead>
<tbody>
<tr>
<td>PS</td>
<td>DSC</td>
<td>94.73 ± 10.68</td>
<td>91.93 ± 19.23</td>
<td>89.10 ± 20.30</td>
<td>94.26 ± 18.30</td>
<td>91.62 ± 20.71</td>
<td>88.65 ± 19.54</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>93.31 ± 19.23</td>
<td>88.60 ± 22.00</td>
<td>92.82 ± 6.40</td>
<td>93.10 ± 10.32</td>
<td>95.02 ± 6.20</td>
<td>90.55 ± 12.17</td>
</tr>
<tr>
<td>RA</td>
<td>DSC</td>
<td>85.35 ± 21.18</td>
<td>80.58 ± 27.24</td>
<td>80.69 ± 31.51</td>
<td>79.69 ± 26.25</td>
<td>81.24 ± 27.06</td>
<td>80.92 ± 27.65</td>
</tr>
<tr>
<td>EO</td>
<td>DSC</td>
<td>92.02 ± 13.10</td>
<td>79.29 ± 23.41</td>
<td>85.18 ± 19.92</td>
<td>87.05 ± 20.12</td>
<td>81.64 ± 17.03</td>
<td>75.76 ± 27.35</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>74.73 ± 29.26</td>
<td>60.73 ± 36.51</td>
<td>51.85 ± 32.83</td>
<td>60.93 ± 36.05</td>
<td>68.59 ± 37.92</td>
<td>52.44 ± 36.14</td>
</tr>
<tr>
<td>IO</td>
<td>DSC</td>
<td>47.39 ± 31.09</td>
<td>52.22 ± 30.09</td>
<td>50.05 ± 28.59</td>
<td>44.60 ± 32.80</td>
<td>47.91 ± 29.63</td>
<td>47.29 ± 21.80</td>
</tr>
<tr>
<td></td>
<td>ASC</td>
<td>41.48 ± 31.78</td>
<td>45.95 ± 34.09</td>
<td>48.00 ± 34.75</td>
<td>42.42 ± 34.71</td>
<td>48.06 ± 33.19</td>
<td>45.36 ± 34.78</td>
</tr>
</tbody>
</table>

**Muscle:** PS: paraspinal; RA: rectus abdominis, EO, external oblique, IO: internal oblique.
**Phase:** DSC: descent, ASC: ascent.
**Load & Support Surface:** S30: 30% stable, S50: 50% stable, S70: 70% stable, U30: 30% unstable, U50: 50% unstable, U70: 70% unstable.
**DSC:** 0% = start of the movement phase (top); 100% = end of the movement phase (bottom).
**ASC:** 100% = start of the movement phase (bottom); 0% = end of the movement phase (top).
### Table 14
Statistical analysis results comparing the differences in relative point in the hip flexion ROM that peak trunk muscle amplitude occurred between the three load (30%, 50%, 70%) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Phase</th>
<th>Load</th>
<th>Surface</th>
<th>Load x Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>PS</td>
<td>DSC</td>
<td>F(2.50)=4.899, p&lt;0.05</td>
<td>F(1,25)=0.122, p=0.730</td>
<td>F(2.50)=0.002, p=0.998</td>
</tr>
<tr>
<td></td>
<td></td>
<td>30-50: p=0.434; <strong>30-70: p&lt;0.05;</strong> 50-70: p=0.155</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td>*F(1.36,32.60)=0.249, p=0.780</td>
<td>F(1,24)=1.093, p=0.306</td>
<td>*F(1.15,27.68)=2.669, p=0.109</td>
</tr>
<tr>
<td>RA</td>
<td>DSC</td>
<td>*F(1.51,31.76)=0.137, p=0.872</td>
<td>F(1,21)=0.177, p=0.678</td>
<td>F(2.46)=0.584, p=0.562</td>
</tr>
<tr>
<td></td>
<td></td>
<td>F(2.46)=2.038, p=0.142</td>
<td>F(1,23)=2.720, p=0.113</td>
<td>F(2.46)=0.106, p=0.899</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td>F(2.46)=3.471, p&lt;0.05</td>
<td>F(1,24)=2.071, p=0.163</td>
<td>F(2.48)=1.517, p=0.230</td>
</tr>
<tr>
<td>EO</td>
<td>DSC</td>
<td><strong>F(1.48,35.57)=4.222, p&lt;0.05</strong></td>
<td>F(1,24)=0.308, p=0.584</td>
<td><strong>F(1.59, 38.08)=3.573, p&lt;0.05</strong></td>
</tr>
<tr>
<td></td>
<td></td>
<td>30-50: p&lt;0.05; 30-70: p=0.134; 50-70: p = 1.0</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td>*F(1.60,36.74)=0.923, p=0.405</td>
<td>F(1,23)=0.001, p=0.972</td>
<td>F(2.48)=0.293, p=0.747</td>
</tr>
<tr>
<td>IO</td>
<td>DSC</td>
<td>F(2.48)=0.639, p=0.532</td>
<td>F(1,24)=1.275, p=0.270</td>
<td>F(2.48)=0.034, p=0.966</td>
</tr>
<tr>
<td>ASC</td>
<td></td>
<td>*F(1.60,36.74)=0.923, p=0.405</td>
<td>F(1,23)=0.001, p=0.972</td>
<td>F(2.46)=0.293, p=0.747</td>
</tr>
</tbody>
</table>


*Phase:* DSC: descent, ASC: ascent.

*Load:* 30-50: 30% vs. 50%, 30-70: 30% vs. 70%, 50-70: 50% vs. 70%.

* Greenhouse-Geisser correction used to calculate the F statistic (assumption of sphericity violated).
Table 15
Statistical analysis results comparing the differences in relative point in the hip flexion ROM that peak trunk muscle amplitude occurred between the two movement phases (descent, ascent) and two support surface (stable, unstable) conditions.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Load</th>
<th>Phase</th>
<th>Surface</th>
<th>Phase x Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>PS</td>
<td>30</td>
<td>F(1,24)=0.598, p=0.447</td>
<td>F(1,24)=0.097, p=0.758</td>
<td>F(1,24)=0.030, p=0.864</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=0.004, p=0.953</td>
<td>F(1,25)=2.369, p=0.136</td>
<td>F(1,25)=2.020, p=0.168</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,25)=1.372, p=0.253</td>
<td>F(1,25)=1.308, p=0.264</td>
<td>F(1,25)=0.440, p=0.513</td>
</tr>
<tr>
<td>RA</td>
<td>30</td>
<td>F(1,24)=119.484, p&lt;0.001</td>
<td>F(1,24)=3.080, p=0.092</td>
<td>F(1,24)=0.007, p=0.935</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,24)=66.467, p&lt;0.0001</td>
<td>F(1,22)=0.780, p=0.387</td>
<td>F(1,22)=0.045, p=0.834</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,22)=37.980, p&lt;0.0001</td>
<td>F(1,22)=0.170, p=0.684</td>
<td>F(1,22)=0.783, p=0.386</td>
</tr>
<tr>
<td>EO</td>
<td>30</td>
<td>F(1,25)=19.762, p&lt;0.0001</td>
<td>F(1,25)=5.114, p&lt;0.05</td>
<td>F(1,25)=1.764, p=0.196</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,24)=4.965, p&lt;0.05</td>
<td>F(1,24)=1.764, p=0.197</td>
<td>F(1,24)=0.885, p=0.356</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,24)=22.875, p&lt;0.0001</td>
<td>F(1,24)=1.690, p=0.206</td>
<td>F(1,24)=1.480, p=0.236</td>
</tr>
<tr>
<td>IO</td>
<td>30</td>
<td>F(1,25)=0.549, p=0.466</td>
<td>F(1,25)=0.163, p=0.689</td>
<td>F(1,25)=0.333, p=0.569</td>
</tr>
<tr>
<td></td>
<td>50</td>
<td>F(1,25)=1.476, p=0.236</td>
<td>F(1,25)=0.249, p=0.622</td>
<td>F(1,25)=0.808, p=0.377</td>
</tr>
<tr>
<td></td>
<td>70</td>
<td>F(1,22)=0.741, p=0.339</td>
<td>F(1,22)=0.439, p=0.514</td>
<td>F(1,22)=0.043, p=0.837</td>
</tr>
</tbody>
</table>

Load: 30: 30%, 50: 50%, 70: 70%.
5. Discussion

The primary purpose of this study was to examine the effects of progressive external load and support surface stability on trunk muscle activity and lumbar spine ROM in collegiate football players performing the back squat. While the activity of lower limb and low back muscles have been examined during the back squat (Wisløff et al., 2004; Clark et al., 2012), little published research has taken a more comprehensive look at changes in trunk muscle activity, especially that of the IO (Clark et al., 2012). Similarly, while incorporating instability into strength training exercises such as the back squat might provide a more functional approach to improving trunk muscle recruitment (Saeterbakken et al., 2013; Behm et al., 2005; Lawrence et al., 2015), little research has addressed this approach under varying load conditions. While increasing the load and incorporating an unstable support surface have shown specific effects in certain trunk muscles, the relationship between the two has not been investigated.

5.1. Primary Research Questions

5.1.1. Progressive Load and Trunk Muscle Activation

The results support the first hypothesis, as all muscles experienced an increase in either EMGpeak or EMGavg as load increased, during either the descent (DSC) or ascent (ASC) phases of the movement. For the PS muscles, successive increases in load were associated with significant increases in EMGpeak and EMGavg during both phases of the movement. Significant increases in PS muscle activity are common during the back squat given the amount of shear force experienced by the spine during the movement, especially at greater depth (Hamlyn et al., 2007; Clark et al., 2017). The RA experienced significant increases in EMGavg, but not
EMGpeak, with each successive increase in load during the DSC. In contrast, significant increases in EMGpeak were observed with each successive increase in load during the ASC, while EMGavg only experienced significant increases from 30-50% and 30-70%. Bressel et al. (2009) did not report any significant changes in RA EMGavg or EMGpeak from 50-75% 1RM. While Clark et al. (2017) observed increases in RA EMGavg from 75-95% 1RM, its activity was consistently lower than that of the PS and EO. Given that external trunk flexion moments are experienced for the entirety of the movement, the function of the RA in creating stability during the squat likely lies in its co-contraction with other muscles. Finally, the PS and RA showed significant differences between DSC and ASC at all loads for EMGavg and EMGpeak, highlighting the increased demands on the anterior and posterior trunk musculature in maintaining spinal stability under load during ASC.

While the IO saw significant increases in EMGavg from 30-50% and 30-70% during the ASC, increasing load from 50-70% did not elicit any further significant changes during this phase. In contrast, the IO saw significant increases in EMGavg from 30-70% and 50-70% during the DSC, but not 30-50%. This suggests that the relative contribution of the IO in maintaining lumbopelvic stability is greater at lower loads during ASC, and greater at higher loads during DSC. This may relate to the IO’s importance in the abdominal bracing mechanism, where it has been demonstrated to have high relative activity (Maeo et al., 2013). This may also imply the role of abdominal bracing in maintaining spinal stability at the bottom of the squat under heavier load where shear force would be the highest. Willardson et al. (2009) observed no significant differences in IO activity during the back squat from 50-75% 1RM, which may imply a tendency for IO activity to peak at lower loads. However, while Bressel et al. (2009) also observed no significant differences in IO activity from 50-75% 1RM with no external feedback, they found
that cueing the participant increased activity over two fold. Participants in the current study were cued to “brace” prior to performing each repetition, which may explain the significant increases that were observed. In contrast to the IO, the EO only saw increases in EMGavg (30-70%; 50-70%) and EMGpeak (30-70%) during the ASC of the movement. This suggests that the EO has a lesser role relative to the IO in maintaining spinal stability during the DSC, but is more active when moving higher loads against gravity (ASC). Maeo et al. (2013) observed significantly less activity from the EO relative to the IO during abdominal bracing. Previous research does demonstrate the capacity of the EO to reach high levels of activity with increasing loads as high as 95% 1RM, though changes were far greater during the ASC (Bressel et al., 2009; Clark et al., 2017). While co-contraction of the EO is important for abdominal bracing, its function as it relates to global trunk muscle activation reinforces its increased capacity for power generation in higher loading movements during the ASC (Comerford et al., 2001).

5.1.2. Surface Stability and Trunk Muscle Activation

The results do not support the second hypothesis, as no significant differences were observed between stable and unstable surface conditions at any load for the EO and IO (or any of the muscles tested). This most likely suggests that the level of support surface instability was insufficient to elicit additional increases in activity within these muscles. Previous research generally suggests that the PS muscles do not experience significant increases, and may even demonstrate decreases in activity, between stable and unstable surfaces at moderate loads (Behm et al., 2005; Lawrence et al., 2015; Bressel et al., 2009; Schwanbeck et al., 2009), with similar findings observed for the RA (Bressel et al., 2009; Willardson et al., 2009, Schwanbeck et al., 2009).
The IO’s role in maintaining spinal stability during anticipated destabilizing movements has been observed (Westad et al., 2010; Garcia-Vaquero et al., 2012). It is possible that the increases in activity elicited by increasing load challenged the IO to a point where reducing surface stability made no difference in increasing activity further. Willardson et al. (2009) did not report any significant differences in IO activity between a 50% 1RM back squat on a stable surface vs. a bosu ball, which may introduce a greater level of instability than the AIREX™ pads used in this study. The main difference between these stability conditions is that the bosu ball places a single unstable surface under both feet, while the AIREX™ pads allow for two separate unstable surfaces (one pad under each foot). Given that maintaining a neutral trunk position is of great importance during the back squat to optimize force and power production, the combination of total trunk muscle activity and increases in IAP may limit lateral flexion moments to the point where further increases in activity from these stability conditions is not required. That said, under moderate loads, instability must be limited to ensure safety. Bressel et al. (2009) observed a 64% increase in EO activity when a back squat at 50% 1RM was performed on a bosu ball. These findings are conflicted by those of Willardson et al., (2009), who observed increases in EO activity from 50-75% 1RM, but not from stable to unstable conditions at 50% 1RM. Clark et al. (2017) found significant increases in EO activity for the back squat (less stable) compared to a machine back squat (more stable). Hamlyn et al. (2007) found that an unstable sidebridge exercise (isometric lateral flexion) was able to produce equivalent EO activity to that of an 80% 1RM back squat, showing how direct planar resistance produces activity that is comparable to its contribution to sagittal plane stability during dynamic movements.

After reviewing previous research on muscle activity during the back squat, Clark et al. (2012) concluded that performing the exercise at moderate loads was more effective in activating
the trunk muscles as a whole compared to other instability trunk exercises. While increasing frontal plane instability by a greater magnitude will increase EO and IO activity, this increase occurs to maintain mechanical stability. Potential increases in global stiffness from these muscles may do little to actually improve load capacity in the sagittal plane when also factoring in potential mechanical insufficiencies produced by the unstable environment. While increased sagittal plane loading causes increases in lateral trunk muscle activity, it is to a much lesser extent than the posterior trunk muscles. Therefore, the magnitude of EO and IO activity during isolated lateral flexion or low load, highly unstable movements, may be much higher than what is required of these muscles to sufficiently stabilize the trunk during heavy sagittal plane exercises, such as the back squat, clean, or deadlift (Hamlyn et al., 2007). This would suggest that challenging these muscles directly would be more effective in improving trunk strength.

5.1.3. Progressive Load and Lumbar Spine ROMs

The results support the third hypothesis, as LLUM extension was observed to increase from 30-70% during the DSC, and significant decreases in LLUM flexion were observed with each successive increase in load. It is suspected that the participants may have had a tendency to hyperextend the lower lumbar spine rather than purely moving into hip flexion with a neutral spine as they initiated the descent. In contrast, LLUM extension was significantly lower at all loads during the ASC vs. DSC, meaning participants were able to return to and maintain a more neutral position during the ASC. These findings are consistent with those of Walsh et al. (2007), who observed decreases in flexion paired with increases in extension between 40%, 60%, and 80% 1RM, though phase of movement was not specified. It would appear that decreases in lumbar flexion (or increases in extension) are not necessarily associated with increases in PS
muscle activity. Decreases in LLUM extension during the ASC may further suggest that a more neutral position is mechanically advantageous when accelerating load against gravity as opposed to decelerating it during the DSC. ULUM flexion was also observed to decrease from 30-70% and 50-70% during the ASC. It can also be noted that, relative to the neutral (standing) position, the ULUM stays in a flexed position for the entirety of the movement (DSC: 19.28° to 29.46°; ASC: 19.41° to 28.95°), while the LLUM adopts both flexed and extended positions during the movement (DSC: -12.14° to 17.52°; ASC: -7.59° to 18.74°). McKean et al. (2010) observed a greater ROM at the lumbar spine during the DSC of the squat, and minimal ROM during the ASC. Mean segment angles indicate that the LLUM moved through a greater total ROM than the ULUM for both DSC and ASC. This may be attributed to lumbar-pelvic movement coupling, as high relative movement between the lumbar and pelvis segments has been observed during the DSC (McKean et al., 2010). While decreases in flexion with increasing load are a common compensation of increased spinal loading (Adams et al., 2000; Walsh et al., 2007), these changes appear to be more prominent at the LLUM. This further suggests that the LLUM and ULUM respond differently to external loading, and therefore modelling the lumbar spine as one segment may limit the interpretation of previous reports. Finally, LLUM LR was observed to decrease from 50-70% during the DSC. Similarly, LLF of the ULUM increased significantly from 30-50% during the ASC. The importance and interpretation of these findings is unclear, as few studies have examined spinal kinematics outside the sagittal plane (i.e., flexion/extension) during the squat.
5.1.4. Surface Stability and Lumbar Spine ROMs

The results do not support the fourth hypothesis, as no significant changes in lumbar ROMs were observed between the support surface conditions. As stated previously, the amount of support surface instability used in the current study may have been insufficient to evoke any changes in lumbar ROMs. However, the amount of multi-planar instability that would be required to provoke changes in lumbar ROMs in collegiate level athletes, especially in the frontal plane, would likely be unsafe under moderate loads and have little application to performance.

5.2. Secondary Research Questions

5.2.1. Progressive Load and Relative Timing of Peak Trunk Muscle Activation

Peak activity of the PS muscles occurred closer to the bottom of the squat during the DSC at 30% compared to 70%. Peak activity of the EO also occurred closer to the bottom of the squat during both the DSC and ASC at 30% compared to 70%. While the bottom of the movement is where the spine would be expected to experience the most shear force (due to more forward torso lean), a higher load caused peak activity in the PS and EO to occur earlier in the movement during DSC. This suggests that with further increases in load, peak activity of these muscles may be reached at a higher point in the movement, implying higher shear forces earlier in the movement. The findings for the EO also suggest that frontal plane lumbopelvic stability requirements may be greater earlier on in both phases of the squat at higher loads. Lateral “hip shift” is a compensation that has been recognized to occur when there is a strength imbalance between lower limbs, whereby the pelvis will “shift” to one (dominant) side (Snarr et al., 2015). It is possible that as load (and therefore the possibility of compensation) increases, the EO
reaches peak activity earlier to resist lateral flexion moments that may be produced by shifting of the pelvis away from the midline.

Peak activity of the RA occurred closer to the bottom of the movement during DSC compared to ASC at all loads. It is possible that the RA’s role in maintaining spinal stability may be more prominent in a lower position as the PS approaches peak activity during the DSC. In contrast, it is also possible that the high amount of PS activity observed when moving load against gravity (especially in the beginning) has an inhibitory effect on the RA, given their antagonistic functions. Shear force would be expected to peak at the beginning of the ASC, as forward trunk lean is more significant and the load is being moved against gravity. Increased RA activity in this bottom position may promote spinal flexion and be counterproductive to maintaining a neutral spine, especially at higher loads.

5.2.2. Surface Stability and Relative Timing of Peak Trunk Muscle Activation

A significant interaction effect for support surface was only observed for EO, where peak activity occurred closer to the bottom of the squat during the stable condition compared to the unstable condition at 30% load. A potential explanation for this finding relates to the compliancy of the balance pads that were used in the unstable condition. When a participant stepped onto the balance pads, the depth that their feet sank into the pads would have been proportional to the load. A heavier load would have increased the density of the pad by compressing the material under the participant’s feet, thereby potentially (and inadvertently) increasing the stability of the surface. In other words, the level of instability may have been higher at lighter loads.
5.3. Implications

In summary, surface instability, as simulated by AIREX™ pads, had no significant effects on the average or peak activity, and few effects on relative timing of peak activity, in the trunk muscles examined in the current study. Load caused activity to increase in all muscle groups from 30-70%, though increases in certain muscles were specific to the phase of movement. Surface instability had no influence on lumbar spine ROM, while load progression caused decreases in ULUM flexion and increases in LLUM extension during the DSC. These findings suggest that the level of support surface instability required to increase trunk muscle activation is beyond what was used in this study at low to moderate loads. Furthermore, the activation of the tested muscles may be sufficient in providing spinal stability in healthy athletes, and further activation (induced by instability or not), may not be needed at the tested loads. Previous studies suggest that increasing bilateral instability (bosu ball) during the back squat is also generally ineffective in increasing the activity in most trunk muscles, with some exceptions for the EO at 50% 1RM (Willardson et al., 2009; Bressel et al., 2009). For the present study, it was hypothesized that by incorporating minimal instability during the back squat, the bracing mechanism could be enhanced during this exercise. However, this does not appear to be the case. It is possible that some of the participants subconsciously widened their stance in order to increase stability by increasing the size of their base of support.

The PS muscles can be recognized as the primary stabilizer of the spine in maintaining a neutral position and reducing shear forces during the back squat based on our findings. Differences in IO activity may provide useful insight on its response to sagittal trunk loading, since more activity was observed at higher loads during the DSC and at lower loads during the ASC. This would imply that the IO plays a significant role in maintaining lumbopelvic stability.
during higher force, lower limb eccentric loading. These same findings may also be perceived as a limited ability to maintain the abdominal bracing technique during the concentric portion of the squat, relative to the eccentric portion, as loads increase. This may suggest that slower eccentric tempos or pauses could be incorporated to train the IO in its role of bracing the trunk during the back squat as opposed to further increasing load. In contrast, the EO had increased activity at higher loads during the ASC. This would suggest that the EO plays a significant role in maintaining stability as shear and spinal stability demands increase, and can be trained more effectively by emphasizing the concentric portion of the squat, with increases in load further challenging the muscle. It also appears that the EO plays a much different role in providing lumbopelvic stability compared to the IO, with decreased activity observed when the IO is more active. These muscles seem to have independent but coordinated activation that collectively maintains stability, with their relative contribution changing with the phase of movement. The consistent increases in RA activity with increasing load during the DSC imply a more important role of the muscle in reinforcing spinal stability than previously thought. Similar to the IO, the RA could be trained using specific cueing tempo alterations.

Increases in LLUM extension with increasing load during the DSC have been observed previously (Adams et al., 2000; Walsh et al., 2007), though typically more significant increases in lumbar extension are seen at greater loads (Adams et al., 2000; Walsh et al., 2007). This may be attributed to individuals attempting to load the hips by “pushing the hips back”, leading to compensatory hyperextension of the spine. This may suggest the need to modify certain cues used to reinforce technique when initiating the DSC in order to maintain a neutral spine, as increases in extension during the back squat under load have been observed to increase spinal compression forces (Adams et al., 2000; Walsh et al., 2007). The ULUM was also observed to
maintain a flexed posture (relative to the normal standing posture) through the entirety of the movement. While initially this may seem to imply greater risk of flexion related injury at the ULUM (e.g., disc herniation), this can likely be attributed to differences in the magnitude of forward trunk lean and hip flexion in the starting position with the barbell placed on the back to keep the CoM over the mid-foot compared to the normal standing posture. Significant pelvic movement may not occur until the repetition is initiated, despite a shift in torso position once the bar is placed on the back. The differences in ULUM vs LLUM ROMs (e.g., flexion/extension) indicate that there are regional differences in how the lumbar spine moves during the back squat. This also suggests that the LLUM undergoes greater stress than the ULUM, given the flexion/extension cycles that are experienced (Foss et al., 2012). These differences in regional lumbar ROMs highlight the importance of modelling the lumbar spine as multiple segments.

Increases in load were associated with an earlier peak activity in the PS during DSC, suggesting that force output of the PS may peak prior to reaching the most stressful part of the ROM (in terms of shear force), or that shear force peaks before reaching the bottom of the movement. Similar findings for the EO during DSC and ASC suggest that lumbopelvic stability requirements are higher closer to the top of the movement in both phases at higher loads. Peak activity in the RA that occurred closer to the top of the movement during ASC compared to DSC at all loads may be related to individuals having a tendency to hyperextend the trunk during ASC, which would be controlled by isometric or eccentric contractions of the RA.

Trunk strength and stability, from a sports performance standpoint, is a component of full kinetic chain movements and requires the tolerance of characteristically similar forces, torques, and velocities (i.e., specificity) of the task demands of the sport (Silfies et al. (2015). For adaptations to be made, these specific variables must be challenged to some extent over a period
of time. The findings of this study confirm that for the given level of instability in the squat, while it may replicate lower limb power production in more sport specific conditions, does not influence trunk muscle activity or lumbar ROMs. Conversely, increases in load not only increase trunk muscle activity, but change the timing of peak muscle activity. This effectively identifies potential methods of utilizing the squat at different loads or emphasizing specific phases or ranges of the movement in order to more effectively target specific muscles in bilateral conditions. Given that these muscles can be collectively challenged in combination with those in the lower limb during the back squat, it can be effectively utilized as a sport-specific lower limb and trunk strengthening exercise. While some of these muscles may not be isolated to the same extent as they are with more traditional trunk strengthening exercises (e.g., side plank), they are activated in a more sport-specific and functional manner in terms of incorporating total trunk stiffness with full kinetic chain movement. Additionally, muscle strength and activation in more direct trunk strengthening exercises may not be indicative of their ability to generate functional trunk stability (Okada et al., 2010).

5.4. Limitations

Performing research on competitive athletes can be particularly difficult as their overall training volume between the weight room and on the field is fairly high, depending on the time of year. In the present study, athletes were tested primarily in the off season (January to August), so as to not influence in-season performance. Given the higher volume of resistance training in the off season, some level of residual fatigue from previous training days can be impossible to avoid at times (even with athletes having 48 hours of lower body rest prior to testing in the
present study), and may vary based on which training block the athletes are in at the time of testing.

For this study, only lean athletes were selected (under 15% bodyfat, using previous team testing data) to improve the validity of the EMG and motion capture data. Occasionally, there were trials (squat or normalization) where large (artificial) spikes were observed in the EMG data, likely caused by motion artifact. Some of these trials were not included in the subsequent analyses, thereby reducing the number of trials used for specific loads and stability conditions for some participants. It is possible that a more secure fastening system could have been used to limit motion artifact, but it was deemed preferable that the movement feel as natural as possible while not interfering with the placement of the spinal motion capture markers. The PS normalization (MVIC) values were likely not valid, as 100% MVIC was exceeded at 50% and 70% 1RM during the ASC, with even larger increases having been demonstrated at higher loads (Clark et al., 2017). It is suspected that the athletes did not fully exert themselves during the normalization trials as instructed, or that the resistance used was insufficient to elicit maximal contractions. This makes some of our normalized EMG results hard to validate, as this overestimation is not observed in all muscle groups. It was also observed that the EO and IO normalization exercises, while eliciting higher activity in the intended muscle for some participants, did not show any differences in others and in a few cases produced the opposite effect (e.g., the IO exercise produced larger values for the EO than the EO exercise). Knowing this, it would be best to perform tests and choose the test that elicits the highest activity for each muscle group.

Although previous research brings into question the capacity of the TrA to contribute to spinal stability under higher loads and while bracing (Grenier 2007; Southwell, 2016), it is
possible that some IO EMG readings included TrA crosstalk activity. Additionally, while the EMG electrode attachment sites were outlined on the skin with a marker, some electrodes occasionally had to be re-applied due to sweat during testing, which may have reduced the reliability and validity of the readings (Burden, 2010; Balshaw et al., 2012). Unfortunately, this could not be completely avoided.

For the pelvis, the motion capture markers were often placed on the participants’ compression shorts/garments. This may have introduced error in calculating pelvic motion, and potentially that of the lower lumbar segment and thighs since motion of these segments was calculated relative to the pelvis. Additionally, the centre of the spinal base plates could not be securely attached to the skin over the lumbar spine for some individuals with prominent PS muscles, which may have introduced errors in lumbar kinematic calculations.

Squat stance width was not controlled between the stable and unstable conditions since it was assumed that participants would maintain a consistent stance width (their normal squat stance). Forcing all athletes to adapt to a specific stance may have influenced their 1RM and coordination during the lift. Additionally, differences in movement tempo between testing conditions was not controlled and may have influenced loading of the trunk and lower limb and the resulting muscle activation and kinematic patterns measured in the study. Lastly, a 1RM determined under stable conditions will likely be greater than in unstable conditions, as instability limits force production of the agonists of the lower limb (Clark et al., 2017). However, due to safety concerns imposed by testing 1RM in unstable conditions, the loads used for both stability conditions were calculated relative to the participant’s 1RM performed on a stable surface. The influence of this on the results and their generalizability is unknown.
5.5. Conclusions & Future Research

The incorporation of an unstable support surface into compound, functional movements has been an area of interest to the sport performance community for many years. While the use of an unstable surface to increase lumbopelvic stability demands during the back squat may be effective at low loads with fairly high levels of instability, it is debatable whether its application is useful to enhance sport-specific trunk strength and stability adaptations in athletes at the given loads and level of instability. The back squat, being a bilateral exercise, is inherently stable, which allows for the use of relatively heavy training loads. Capacity to maintain trunk stiffness under high load likely has more application to high loading sports than low-moderate loads in unstable conditions. Therefore, it seems that the back squat is an effective exercise to train trunk strength and stability, and can be utilized at moderate to high loads with proper cueing to train these components of performance. If the goal is to improve hip and trunk stability and proprioception, unilateral exercises may be more effective in targeting these areas, while also decreasing the likelihood of developing unilateral deficiencies. It would be valuable to quantify the level of instability and how it affects balance (e.g., measuring excursions of the center of pressure) to more effectively analyze the unstable condition, as this has not been examined in this context. Future research should examine how trunk muscle activity and spinal kinematic patterns differ between bilateral and unilateral exercise variations.
References


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Appendix A: REB Certificate of Approval

Research Ethics Board
Certificate of Renewal Approval

Principal Investigator: Ian Rowan-Legg
DEPARTMENT: Faculty of Kinesiology and Health Studies
REB#: 2018-033

Supervisor: Dr. Paul Bruno

Title: Trunk Muscle Activation Patterns and Lumbar Ranges of Motion During the Barbell Back Squat with Progressive Load and Instability

Original Date of Approval: March 20, 2018
New Expiry Date with this Renewal: March 20, 2020
Today's Date: March 12, 2019

Full Board Meeting: [ ]
Delegated Review: [X]

Renewal Certification
The University of Regina Research Ethics Board has renewed the above-named research project for an additional 12 months.

Any significant changes to your proposed methods, procedures or related documents should be reported to the Research Ethics Board Chair for consideration in advance of its implementation.

Ongoing Review Requirements
In order to receive annual renewal, a status report must be submitted to the REB Chair for Board consideration within one month of the current expiry date each year the study remains open, and upon study completion. Please refer to the following website for the renewal and closure forms:

https://www.uregina.ca/research/or-faculty-staff/ethics-compliance/human/ethicsforms.html

Ara Steininger
Research Ethics Board

Please send all correspondence to:
Research Office
University of Regina
Research and Innovation Centre 106
Regina, SK S4S 0A2
Telephone: (306) 585-4776 Fax: (306) 585-4900
research.ethics@uregina.ca
Appendix B: Study Information Sheet

Project Title: Trunk Muscle Activation Patterns & Lumbar Ranges of Motion During the Barbell Back Squat with Progressive load and Instability

Researchers: Ian Rowan-Legg, BKin, CSCS  Paul Bruno, DC, PhD

Introduction
You are being asked to take part in a research study investigating the impact of loaded stability training on trunk muscle activity and lumbar spine range of motion in collegiate level football players. Before you decide whether or not to take part, it is important that you understand why the research is being done and what it will involve. Please read the following information carefully and ask questions of the researchers if there is anything that is not clear or if you would like more information.

What is Required of Me if I Take Part?

Initial Strength Testing (~30 minutes)
Prior to the data collection session, a strength testing session will be scheduled a minimum of 72 hours before the data collection in order to test the participant’s 1 repetition maximum (RM) back squat. This is necessary for accurately selecting loads for data collection.

Data Collection Session (~60 minutes)
You will be asked to wear a pair of compression shorts (provided). Small markers and surface electromyography sensors will be attached to specific locations around your low back, pelvis, and thighs using double-sided adhesive tape. During this session, you will be required to:

1) Perform 4 static, short duration (5 seconds), maximal effort muscle contractions in various conditions (lying on stomach, back, and side).
2) Perform an instructed dynamic warm up.
3) Perform 3 sets of 3 repetitions of the back squat at 30%, 50%, and 70% 1 RM, in both a stable and unstable condition (6 sets total). Rest periods of 3 minutes will be allotted between sets, and 5 minutes between stability conditions.

While you perform these tasks, a motion capture system will be used to track the position of the markers and collect data related to the activity in the underlying muscles.

Do I Have to Take Part?
Participation in this project is completely voluntary, and there is no penalty for deciding not to participate. Whether you participate or not in the project will not be communicated to your coaches, and not participating will NOT affect your standing with the coaches or the team. If you do decide to participate, you may keep this information sheet and will be asked to sign a consent form. Also, if you do decide to participate, you are free to withdraw at any time and without giving a reason. You may also withdraw from the study at any point prior to the dissemination of the study findings (e.g., publications, presentations), which is estimated to begin in September 2018, by contacting the primary investigator (contact information below).

What are the Benefits of Taking Part in this Study?
The results of this project will allow researchers and strength coaches to better understand how collegiate level football players’ trunk musculature responds to loaded stability training, and give insight into potential training strategies for performance and injury prevention.

What are the Risks of Taking Part in this Study?
There is a risk that you may sustain a musculoskeletal injury (e.g., sprain, strain) while performing the tasks. Additionally, a 1 RM testing is required to accurately give loads for the data collection session. The 1 RM back squat testing will be fatiguing and may result in soreness in the following 24-72 hours. The data collection process will use much lighter loads (see above) at very low volume, but may still result in some temporary fatigue and minor soreness following the session.
To ensure that the muscle recordings are of a high quality, your skin will be prepared before the EMG sensors are applied. For each sensor, this will consist of shaving a small area of skin (if necessary), light abrasion, and an alcohol swab. This may cause slight discomfort, and if so it will only be temporary. Since the adhesive tape that will be used to attach the sensors and quite sticky, you may notice mild local discomfort during their removal and small red marks on your skin afterward. If either of these occurs, they normally go away in a short time. Please inform the researchers if you have any known reactions to adhesive tape (e.g., rash, itchiness). If such a reaction occurs and persists for more than 24-48 hours, you are advised to contact the researchers and seek a medical assessment.

**Will My Taking Part be Kept Confidential?**
All information collected about you during the course of this research will be kept completely confidential. Your signed informed consent form will be stored in a locked filing cabinet. The only persons who will have access to these forms are the researchers involved in this study. Any published information based on the data collected in this study will not have your name or any other identifying markers appended to it.

**What Will Happen to the Results of this Study?**
The results of this study will be submitted as part of the principal investigators Master’s Thesis, and may be published in academic journals and presented at academic conferences.

**Who Has Reviewed this Study?**
The ethics for this study have been reviewed and approved by the Research Ethics Boards of the University of Regina.

**Contact Information for Further Queries:**

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<tr>
<td>Ian Rowan-Legg</td>
<td>Faculty of Kinesiology and Health Studies</td>
<td>University of Regina</td>
<td>(306) 337-3343</td>
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<td>Paul Bruno</td>
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Appendix C: Informed Consent Form

**Project Title:** Trunk Muscle Activation Patterns & Lumbar Ranges of Motion During the Barbell Back Squat with Progressive Load and Instability

**Researchers:** Ian Rowan-Legg, BKin, CSCS
Paul Bruno, DC, PhD

1. I confirm that I have read and understand the information sheet for this research project and have had the opportunity to ask questions.
2. I understand that my participation in this study is completely voluntary, and that not participating will **NOT** affect my standing with my coaches or the team.
3. I understand that I am free to withdraw at any time without penalty and without giving a reason by contacting the researchers using the contact information below.
4. I understand that I may withdraw from the study at any time after data collection by informing the researchers (contact information below).
5. I acknowledge that I have received a copy of this consent form.
6. I agree to take part in the study.

_________________________        _____________        _______________________
Participant Name (please print) Date Signature

_________________________        _____________        _______________________
Researcher Name (please print) Date Signature

Note: All participants in this research study will be entered into a draw to win one of two $50 prizes.

**Contact Information for Further Queries:**

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This research project was reviewed and approved on ethical grounds through a review process by the University of Regina. Any questions regarding your rights as a participant may be addressed to that committee through the U of R Research Ethics Office at research.ethics@uregina.ca or (306) 585-4775.