Neuromuscular assessment of trunk muscle function in loaded, free barbell back squat: Implications for development of trunk stability in dynamic athletic activity.

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Acknowledgements

The journey began in 2010 when I decided to review the literature on core stability and trunk muscle activation in the back squat exercise. That may well have been the end of the story had Angus Hunter not arranged for me to present my findings to Journal Club at my alma mater, Exercise Science and Sport Medicine Division, University of Cape Town. Feedback from Tim Noakes, Mike and Vicki Lambert was that it was a worthwhile topic to research and I should consider doing a PhD

I embarked on a PhD by publication so that I could work through the topic; effectiveness of loaded squat in activating the trunk stabilizers, while developing research and publication skills and enhancing academic credibility.

I chose two friends, Angus Hunter and Mike Lambert, as academic supervisors and can happily say that those friendships have grown thanks to the PhD. Testimony to the quality of their academic support is the fact that I completed this journey. As an applied practitioner, who was fully immersed in that world, I needed guidance and support in every step; research design, neuromuscular electromyography, data analysis, interpretation and publication. So, thanks Angus and Mike for your support, patience and friendship.

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Dedication

This journey would not have been possible without the love and support of my family; Jason, Lauren and Ingrid. While working on the PhD, we relocated on three occasions, twice to new countries. On each occasion, Ingrid managed to *make a loving home* for the family in her inimitable style.

Abstract

Traditional core stability training was developed as a method of treating and preventing back pain. It was however, seamlessly applied to healthy and athletic populations without scientific evidence supporting its efficacy. Traditional core stability focussed on isolating and training the anatomical region between the pelvis and diaphragm, using isometric or low load exercises to enhance spinal stability. Scientific research challenged this approach for healthy function and athletic performance, resulting in a more functional anatomical definition, which included pelvic and shoulder girdles. Hence, a revised definition of dynamic trunk stability; the efficient coordination, transfer and resistance by the trunk, of force and power generated by upper and lower appendicular skeletal extremities during all human movement. This led to an integrated exercise training approach to dynamic trunk stability. Although early evidence suggested loaded compound exercises preformed upright, in particular back squat, were effective in activating and developing trunk muscles, evidence was inconclusive.

Accordingly, the aims of this PhD were to investigate neuromuscular trunk function in loaded, free barbell back squat to understand training implications for trunk stability in dynamic athletic activity. Five research studies were conducted; 4 are published and 1 is being prepared for re-submission.

The literature review revealed evidence that back squat was an effective method of activating trunk stabilzers and showed that these muscles were load sensitive (study 1). A survey of practitioners reported an understanding and appreciation of the challenge against core stability training for athletic populations. Furthermore, perceptions were aligned with growing evidence for dynamic and functional trunk stability training (study 2). A test-retest neuromuscular study established *inter*day reliability and sensitivity of electromyographical measurement of trunk muscle activity in squats (study 3). Trunk muscle activation in back squat was higher than hack squat at the same relative, but lower absolute loads (study 4). Trunk muscle activation was lower in squats and bodyweight jumps in the strong compared to weak group (study 5). Furthermore, activation of the trunk muscles increased in each 30^o segment of squat descent and was highest in first 30^o segment of ascent for all loads (study 5).

In conclusion, this series of studies confirmed acute effect of squats on trunk stabilizers and demonstrated that external load increases activation in these muscles. Parallel squat depth is important in optimizing trunk muscle activation. Finally, high levels of squat strength result in lower trunk muscle activation in loaded squats and explosive jumps.

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Paper 2

Clark, D. R., Lambert, M. I., & Hunter, A. M. (2018). Contemporary perspectives of core stability training for dynamic athletic performance: A survey of athletes, coaches, sports science and sports medicine practitioners. *Sports Medicine-Open, 4(1), 2-10.*

Paper 3

Clark, D., Lambert, M. I., & Hunter, A. M. (2016). Reliability of Trunk Muscle Electromyography in the Loaded Back Squat Exercise. *International Journal of Sports Medicine,* 37(6), 448–456.

Paper 4

Clark, D. R., Lambert, M. I., & Hunter, A. M. (2017). Trunk muscle activation in the back and hack squat at the same relative loads. *Journal of Strength and Conditioning Research,* July 12. Published ahead of Print.

Paper 5

Clark, D. R., Lambert, M. I., Grigson, C., & Hunter, A. M. (2018). Impact of training status on trunk muscle activation in back squat, squat jump and countermovement jump. *To be submitted.*

Abbreviations used in the thesis

ANOVA	analysis of variance
ARV	average rectified value (Electromyography signal)
BS	back squat
BW	body weight
CI	confidence interval
cm	centimetre
CMJ	countermovement jump
CS	core stability
CST	core stability training
CV	coefficient of variation
EMG	electromyography
ES	effect size
HS	hack squat
Hz	Hertz
ICC	intra-class correlation coefficient
iEMG	integrated electromyography
iMVF	isometric maximal voluntary force
kg	kilogram
LOA	limits of agreement
MVC	maximal voluntary contraction
MVIC	maximal voluntary isometric contraction
MRFD	maximal rate of force development
m/s	metre per second (velocity)
NWB	no weight belt
p	p value
RFD	rate of force development
RM	repetition maximum
RMS	root mean square
s	seconds

SD	standard deviation
SJ	squat jump
sEMG	surface electromyography
SM / SMmax	system mass maximum
TMA	trunk muscle activation
W	Watt

Muscle site abbreviations

AL	adductor longus
AM	adductor major
AMG	adductor magnus
BF	biceps femoris
EO	external oblique
ES	erector spinae
GMA	gluteus maximus
GME	gluteus medius
GN	gastrocnemius
LSES	lumbar sacral erector spinae
RA	rectus abdominus
RF	rectus femoris
SO	soleus
ST	semitendinosus
TA	tibialis anterior
TRA	transversus abdominus
ULES	upper lumbar erector spinae
VL	vastus lateralis
VM	vastus medialis

Study 5: Groups

SG	strong group
MG	middle group
WG	weak group

Study 5: Tertiles (Appendix 2)

E-1	1st eccentric tertile (0-30^{o} knee flexion)
E-2	2nd eccentric tertile (30-60^{o} knee flexion)
E-3	3rd eccentric tertile (60-90^{o} knee flexion)
C-1	1st concentric tertile (90-60^{o} knee flexion)
C-2	2nd concentric tertile (60-30^{o} knee flexion)
C-3	3rd concentric tertile (30-0^{o} knee flexion)

Introduction

Background

This PhD and a suite of research studies are the response to a question that arose in applied strength and conditioning during the early 2000s: *Is traditional core stability training appropriate for the development of trunk stability for dynamic athletic activity?* Clinical research into back pain and the role of deep and superficial trunk stabilizer muscles established the concept of core stability training (CST) in mid-1990 (Panjabi, 1992; Hodges and Richardson, 1998). The transfer and application of CST to healthy, uninjured and athletic populations spread seamlessly without apparent scientific justification (Lederman, 2010). There was however, significant commercial interest and investment (J. Willardson, 2007). In the exercise and fitness industry, renowned for gimmicks and fads, CST gained traction and spread to all sectors including strength and conditioning.

Context

Traditional approach to CST is incongruous to physical development of athletes for dynamic athletic performance for several reasons. The most obvious concern is the reductionist approach of CST, isolating and training muscular, skeletal and neural structures between the diaphragm and pelvis (Panjabi, 1992; Hodges and Richardson, 1998). In dynamic sporting activity, this anatomical region functions as an integral part of the full kinetic chain and therefore should be trained or developed within that context. The second major issue with traditional CST is the isometric nature of the exercise format. This flaunted well established exercise training principles of specificity and overload, particularly relevant for dynamic athletic, movement, characterised by any combination of high velocity, force and torque.

There has been extensive research on the topic of core stability in healthy populations and for athletic performance (Kibler, Press and Sciascia, 2006; J. M. Willardson, 2007; Hibbs *et al.*, 2008; Reed *et al.*, 2012; Martuscello *et al.*, 2013; Silfies *et al.*, 2015; Maaswinkel *et al.*, 2016; Prieske, Muehlbauer and Granacher, 2016; Wirth, Hartmann, *et al.*, 2016). The particular area of interest within the published literature was the loaded back squat and efficacy of this exercise in activating the trunk stabilizers. This exercise is specific to many sporting activities and can be overloaded safely and effectively. Most important however, is the manner in which it challenges the core or trunk in an integrated and functional way. In fact, for novice squatters on a progressive loaded squat programme, the ability to stabilize the trunk represents the limiting factor. In other words, primary adaptation in the first stage of a progressive load squat programme is trunk stability. Hence, development effective trunk stability is required in order to begin to overload lower limbs, the main purpose of squat training for athletic performance. What is unknown is how changes in squat load effect trunk muscle activation or how chronic squat strength training changes neuromuscular function of trunk stabilizers.

There is compelling evidence that trunk stability is dependent on effective coordination of all muscles of the trunk (Cholewicki and VanVliet Iv, 2002; McGill *et al.*, 2003; Akuthota and Nadler, 2004; Hibbs *et al.*, 2008; Behm *et al.*, 2010; Wirth, Hartmann, *et al.*, 2016). The trunk muscles with the largest moment arms are the erector spinae, quadratus lumborum and rectus abdominis which are responsible for stabilizing the spinal column (Cholewicki and VanVliet Iv, 2002; Behm *et al.*, 2010). The internal and external obliques and transversus abdominis develop intra-abdominal pressure, thereby stabilizing the spine specifically in the lumbar region (Cholewicki and VanVliet Iv, 2002; Behm *et al.*, 2010). McGill et al (1996) determined that surface EMG was an effective method of measuring EMG amplitude in these muscles for most tasks, including maximal voluntary contractions (McGill, Juker and Kropf, 1996). They explained that the magnitude of error increased for deep muscles such as psoas. In our review, the most common muscle sites used to measure trunk stability (Anderson and Behm, 2005; Hamlyn, Behm and Young, 2007; Nuzzo *et al.*, 2008; Bressel *et al.*, 2009; Willardson, Fontana and Bressel, 2009; Clark, Lambert and Hunter, 2012) were (Appendix 3):

- Rectus abdominis (RA: 2 cm lateral from the midline of the umbilicus)
- External oblique (EO: halfway between inferior costal margin of ribs and anterior superior iliac spine)
- Upper lumber erector spinae (ULES: 6 cm lateral to L1-L2 spinous processes)
- Lumbar sacral erector spinae (LSES: 2 cm lateral to L5-S1 spinous processes).

Hence, quadratus lumborum, multifidus, internal oblique and transversus abdominis were excluded. Two reasons put forward by these authors was diminished accuracy of surface EMG in measuring deep muscles and that the selected superficial muscles were reflective of neuromuscular activity during dynamic trunk stability (Anderson and Behm, 2005;

Hamlyn, Behm and Young, 2007; Bressel *et al.*, 2009; Schwanbeck, Chilibeck and Binsted, 2009).

There is agreement in the literature that an important factor contributing to confusion around the topic of core stability is the absence of agreed terms and definitions (Kibler, Press and Sciascia, 2006; Hibbs *et al.*, 2008; Behm *et al.*, 2010; Key, 2013; Martuscello *et al.*, 2013; Silfies *et al.*, 2015; Spencer, Wolf and Rushton, 2016; Wirth, Hartmann, *et al.*, 2016; Clark, Lambert and Hunter, 2018). The terms core stability and core strength have attracted much attention and debate (Hibbs *et al.*, 2008; Prieske *et al.*, 2016; Wirth, Hartmann, *et al.*, 2016). Hibbs et al (2008) described the aetiology of these terms and explored the concept that stability was required for everyday function and rehabilitation from lower back pain (LBP), while strength was necessary for dynamic athletic activity (Hibbs *et al.*, 2008). They concluded that there was no evidence for this separation and that core strength was central to stability (Hibbs *et al.*, 2008). Many now subscribe to a more functional definition of core stability; a dynamic process characterized by effective muscular function and neuromuscular control (Silfies *et al.*, 2015; Prieske *et al.*, 2016). Where muscular function includes both strength and endurance, while neuromuscular control refers to coordination of efferent and afferent neural pathways (Silfies *et al.*, 2015). Describing trunk stability as core strength and core stability is closely associated to the scientific endeavours to measure core strength and endurance (Hibbs *et al.*, 2008; Prieske *et al.*, 2016).

In a systematic review, Prieske et al (2016) found that trunk muscle strength, measured isometric and dynamically, played only a minor role in measures of fitness and athletic performance (Prieske *et al.*, 2016). Isometric tests included timed prone plank and dynamic tests, peak isokinetic torque in trunk flexion and extension (Prieske *et al.*, 2016). Regardless of the absence of an association between trunk muscle strength, fitness and performance tests, it is patently clear that these tests do not reflect complex, dynamic and multifaceted day-to-day movement or athletic activity. This highlights the folly of attempting to measure trunk stability using currently available methods and technology. Based on growing evidence in scientific literature (Prieske *et al.*, 2016; Wirth, Hartmann, *et al.*, 2016) and this glaring methodological testing error, it is clear that *trunk strength is central to trunk stability and is not directly measurable given prevailing methodology*.

Back squat (Papers 3, 4 & 5) and hack squat (Paper 4) exercises are central to research published in papers 3, 4 and 5. Free barbell back squat is a widely used exercise in programmes for sports performance, health and fitness and body building. The hack squat is more common in body building, general fitness and rehabilitation training programmes. The primary purpose of both exercises is to develop eccentric and concentric strength of the lower limb through flexion and extension of knee and hip joints and to a lesser extent ankle joint. The mechanics of free barbell back squat require that the line of force or centre of gravity in the sagittal plane remain over the base of support through the full range of movement. In the hack squat line of gravity does not need to coincide with the point where force is applied in the foot position, due to the supported trunk and fixed external load. These difference in mechanics of these two exercises obviously have implications on the neuromuscular demands of all muscles involved (Appendix 2).

Most neuromuscular research has investigated the acute responses (McCaw and Melrose, 1999; Caterisano *et al.*, 2002; Gullett *et al.*, 2009; Paoli, Marcolin and Petrone, 2009; Brandon *et al.*, 2014) and chronic adaptation (Hakkinen, 1989; Hakkinen *et al.*, 1998; Aagaard *et al.*, 2002) of prime mover muscles responsible for driving concentric load in the back squat. These muscles, the quadriceps group comprise of vastus lateralis, rectus femoris and vastus medialis. Research indicates that isometric (Carolan and Cafarelli, 1992) and isokinetic (Häkkinen *et al.*, 1998) training results in increased maximal voluntary activation of the agonists with a significant reduction in coactivation of the antagonists or muscles opposing the quadriceps, the hamstrings. There is however, little evidence for synergist or stabilizer muscle response to compound lower limb strength training. Buckthorpe and co-workers (2015) measured changes in neural activation of agonist, antagonist and stabilizer muscles after 3 weeks isometric and isointertial elbow flexion training (Buckthorpe *et al.*, 2015). Maximal dynamic strength (1RM) increased significantly more than isometric maximal voluntary force (iMVF). Agonist, antagonist and stabilizer EMG increased significantly in the follow-up 1RM test, and only in the stabilizers for iMVF test. It is highly likely that synergist adaptation to compound lower limb strength exercises, such as back squat, underpin improvements in dynamic squat strength and reported associated performance gains. Despite the obvious importance of trunk stabilizers in transferring and resisting force during the squat, there is scant neuromuscular information on trunk muscle activation response and adaptation to squat training.

Early research compared trunk muscle activation in squats to instability exercises (Anderson and Behm, 2005; Hamlyn, Behm and Young, 2007; Nuzzo *et al.*, 2008; Bressel *et al.*, 2009) and gave the first evidence that squat load directly influenced activation of these muscles. Subsequent studies compared trunk muscle activation in back and overhead squat (Aspe and Swinton, 2014) and the squat to isometric and dynamic strength exercises (Comfort, Pearson and Mather, 2011). There was a clear and obvious requirement for more information on neuromuscular trunk function in loaded, free barbell back squat.

Scope

The suite of studies in this PhD thesis straddle performance sport and applied sports science research. The main question arose in response to a strength and conditioning trend in high performance sport (PhD candidate). The research question and methods were formulated according to principles of applied sports science research through the guidance from two PhD supervisors and many of those acknowledged above (page 2). This conforms to the conceptual model proposed by Coutts (2016), where he proposes that questions innovated in high performance are investigated using robust and rigorous sports science research methods to develop and reinforce evidence based practice (Coutts, 2016).

Despite the significant rise in applied sport science research in performance sport (Coutts, 2016; Kraemer *et al.*, 2017), there is little evidence of this translating effectively to applied practice, let alone sports performance (Bishop, 2008). In response, Bishop (2008) proposed the Applied Research Model for the Sports Sciences to better facilitate translation to practice by informing initial design and conceptualization of research (Bishop, 2008). The model consists of 8 steps progressing from problem definition, concluding with implementation studies to measure the effectiveness in practice. Retrospective alignment of studies in this PhD to this model proves quite insightful. Paper 1, the review and paper 2, the survey defined and contextualized the problem into a research question fulfilling the model’s criteria for step 1 and 2. The third paper established methodological reliability of neuromuscular and kinematic measures proposed for studies 4 and 5. This, along with paper 4 comparing trunk muscle activation in hack versus back squat is classified as descriptive research and is therefore aligned to step 2 of the model. The final study falls across steps 3, 4 and 5; establishing predictors of performance, experimental testing of predictors and determinants of key performance predictors. In this study, we investigate and compare a number of key performance attributes in strong versus weak squatters. The model progresses to intervention studies which would be a recommendation arising from this PhD; conduct a training study to assess the impact of improved squat strength on trunk muscle activation and dynamic trunk stability in proxy tests of athletic performance.

The starting point for this PhD was to review the scientific research literature on the topic of muscle activation in loaded back squat exercise. This is the first published paper of this thesis (Clark, Lambert and Hunter, 2012). This was important in order to determine and confirm the specific area of our research within the context of wider neuromuscular research into the back squat exercise. Furthermore, it was important to review the related neuromuscular research methods as they applied to the back squat exercise. This informed methods used in our study to establish reliability in trunk electromyography (EMG) capture in the back squat (Clark, Lambert and Hunter, 2016). It also determined kinematic tests and EMG normalization methods used in all our subsequent research (Clark, Lambert and Hunter, 2016, 2017).

Application of traditional CST was well established and appeared to withstand the growing scientific challenge. In fact there was a view that it's use in healthy and athletic populations continued to develop and spread (J. Willardson, 2007). Surveys are an effective scientific research tool used previously to assess nutrition knowledge (Torres-McGehee *et al.*, 2012) and understanding of scientific training principles in the workplace (Durell, Pujol and Barnes, 2003). The motivation for the specific area of research in the PhD arose from a clear gulf between applied practice and scientific principles and latterly, published scientific evidence. Hence, the second question was to determine *perceptions and application of core stability training in people working and participating in sport using a survey. In the publication, survey results are analysed to determine extent to which scientific research informed CST perceptions and practice.*

Our first publication, confirmed that back squat was an effective method of activating the trunk muscles and that this activation increased with increases in load (Clark, Lambert and Hunter, 2012). It was also apparent that methodology around the measurement of trunk activation in back squat, reported in the scientific literature was inconsistent and unreliable. Surface electromyography (sEMG) data capture for muscles of the trunk required a standardized approach to ensure that findings could be interpreted, compared and have an impact on practice. This review also identified inconsistencies in EMG normalization methods. *The purpose of the third study was to establish reliability and sensitivity of the measurement of trunk muscle electromyography in the loaded back squat exercise.*

We established reliability of sEMG in measuring trunk muscle activation in the back squat and demonstrated that this method was sensitive to typical load changes in this exercise. Most previous trunk muscle activation research had compared the back squat to isometric (Comfort, Pearson and Mather, 2011) or unstable exercises (Anderson and Behm, 2005; Hamlyn, Behm and Young, 2007; Nuzzo *et al.*, 2008; Bressel *et al.*, 2009). The requirement to stabilize the free barbell in the back squat is a unique and important feature of the exercise. In the one published study, there was no difference in trunk muscle activation between the more stable Smith machine squat and the free barbell squat for the rectus abdominus and erector spinae muscles (Schwanbeck, Chilibeck and Binsted, 2009). This was contrary to what we believed based on our review of the literature. *Hence, the fourth study of the PhD compared trunk muscle activation in free barbell back squat to machine supported hack squat for a range of moderate to heavy loads.*

The survey demonstrated that perceptions and practice amongst people working and participating in sport did reflect the current information in the scientific literature. The review and survey concluded that there was a requirement for more data on efficacy of commonly used exercises in activating trunk stabilizers. A further recommendation was to investigate the adaptations to long-term squat training. Specifically, how does acquired back squat strength through regular, progressive training change trunk stability in dynamic athletic activity? *Hence, the fifth and final study aimed to determine how squat training status influenced trunk muscle activation in the back squat, squat jump and countermovement jump.*

Study 1: Scientific review - muscle activation in the loaded free barbell squat

The scope of the review included all publications that reported muscle activation measured by sEMG in back squat. The review did include data for other exercises where these were compared to back squat. Studies included reported neuromuscular activation data for all muscle sites of the lower limb, hips, thighs and trunk region.

Section headings of the review paper reflect topics covered in the scientific literature. Most of these topics were areas of interest in the applied setting, where research had aimed to verify common applications and variations of back squat exercise using neuromuscular analysis. Common applications include technical squat variations such as stance width, hip rotation and squat depth and programming manipulations such as external load and instability.

A limitation of the review process and publication was the use of a systematic narrative review method rather than a systematic or meta-analysis method. The variety and breadth of the sub-topics related to muscle activation in the back squat was more suited to a systematic narrative approach. Furthermore, differences in research design and methods precluded a systematic of meta-analysis review. The selected method presented and debated findings of selected research publications. It included qualitative analysis in the discussion and quantitative analysis in tables that reported muscle sites investigated in each study (Table 1) along with study design and key findings (Table 2). The outcome however, did serve to guide and inform neuromuscular research in back squat trunk muscle activation. The review summary confirmed there was sufficient evidence that loaded barbell squats were effective in activating trunk stabilizing muscles. Recommended practical applications gave clear direction for subsequent research to determine the role of back squat in developing dynamic trunk stability.

MUSCLE ACTIVATION IN THE LOADED FREE BARBELL SQUAT: A BRIEF REVIEW

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ABSTRACT

Clark, DR, Lambert, MI, and Hunter, AM. Muscle activation in the loaded free barbell squat: A brief review. *J Strength Cond Res* 26(4): 1169–1178, 2012—The purpose of this article was to review a series of studies ($n = 18$) where muscle activation in the free barbell back squat was measured and discussed. The loaded barbell squat is widely used and central to many strength training programs. It is a functional and safe exercise that is obviously transferable to many movements in sports and life. Hence, a large and growing body of research has been published on various aspects of the squat. Training studies have measured the impact of barbell squat loading schemes on selected training adaptations including maximal strength and power changes in the squat. Squat exercise training adaptations and their impact on a variety of performance parameters, in particular countermovement jump, acceleration, and running speed, have also been reported. Furthermore, studies have reported on the muscle activation of the lower limb resulting from variations of squat depth, foot placement, training status, and training intensity. There have also been studies on the impact of squatting with or without a weight belt on trunk muscle activation (TMA). More recently, studies have reported on the effect of instability on TMA and squat performance. Research has also shown that muscle activation of the prime movers in the squat exercise increases with an increase in the external load. Also common variations such as stance width, hip rotation, and front squat do not significantly affect muscle activation. However, despite many studies, this information has not been consolidated, resulting in a lack of consensus about how the information can be applied. Therefore, the purpose of this review was to examine studies that reported muscle activation measured by electromyography in the free barbell back squat with the goal of clarifying the understanding of how the exercise can be applied.

KEY WORDS resistance training, strength tests, athletic performance

INTRODUCTION

The squat exercise has a long history in fitness training, exercise for rehabilitation, and strength training for performance in sport. It is a functional movement that is performed loaded or unloaded by flexing and extending the hip, knee, and ankle joints in a manner similar to many movements that occur in daily activity and sport. The squat exercise is regarded as a closed kinetic chain exercise where the force is expressed through the end (length) of the limb while it is fixed to the ground (11).

Variations of the loaded barbell squat are widely used in the physical preparation programs for athletes in many sports. The primary reasons for this are the functional nature of the squat exercise movement, the ability to overload the muscles during this exercise, and the relative safety (9) of the squat when performed in a squat rack or cage. As a consequence, this exercise and a selection of variants have been subjected to research. For example, training studies have measured the impact of barbell squat loading schemes (14) on selected training adaptations including maximal strength (30) and power changes in the squat (2,14,18,29,31). Squat exercise training adaptations and their impact on a variety of performance parameters, in particular countermovement jump, acceleration, and running speed (7,34), have also been reported. Kinematic, kinetic, and electromyographic (EMG) studies have reported muscle activation of the lower limb resulting from variations of squat depth (5), foot placement (10,24,25), and training status and training intensity (26). There have also been studies (36) on the impact of squatting with or without a weight belt on trunk muscle activation (TMA). More recently, a number of researchers have reported the effect of instability on TMA and squat performance (2,3,13,20,21,23,27,33).

The barbell squat was established as a key exercise in the physical preparation of athletes before the growing body of scientific evidence describing muscle activation in variations of this exercise. However, despite the growing body of

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evidence, there is no consensus on how the information should be applied. The organic manner in which this research has been conducted and published necessitates a review of the findings as they relate to athletic training and therefore inform practice. In particular, because of (a) the narrow topic area, defined by muscle activation in the free barbell back squat; (b) the wide range of methodologies used in the studies; and (c) the spread of muscle sites reported, we decided that a systematic narrative review method was the most appropriate way to achieve our objectives of consolidating the evidence so that it can be applied in a more evidence-based way.

Therefore, the purpose of this article is to review the studies ($n = 18$) that investigated the muscle activation during the free barbell back squat and applied variations of this exercise as used in the physical training of athletes for performance. The review does include data from studies that reported muscle activation in the leg extension, leg press, front squat, and Smith machine squat, only where these exercises have been compared with the free barbell back squat.

A PubMed search of the academic literature was performed using the following terms: "free barbell back squat," "loaded back squat," "back squat," "electromyography," "EMG," and "muscle activation," limited to English articles and human subjects. Literature was also sourced from links to related articles, hand searches, and the bibliographies of academic articles. The searches retrieved 18 full articles, where muscle activation in the loaded free barbell back squat was reported.

Muscle Sites Analyzed

In all the studies reviewed, EMG activity in muscles of the lower limb was measured and reported more than twice as often (56 of 80) as muscles of the trunk (24 of 80) (Table 1). Interest in hamstring and quadriceps activation was the higher than any other muscle group reported; biceps femoris and vastus medialis activation were each reported in 12 of the 18 studies. This reflects the fact that these muscles are traditionally the primary muscles of interest in loaded squat training, hence the interest in factors and variables that may influence muscle activation and therefore training efficacy of the squat exercise.

More recently, a number of investigators (2,3,13,23,27,33) have reported TMA for the loaded squat exercise under different conditions. The trunk muscles reported most frequently are the rectus abdominus (6 of 18), external oblique (4/18), and various aspects of the erector spinae (8 of 18). In all the studies reviewed, the measurement of the activation of the muscles of the trunk is used to assess the impact of stability vs. instability in the squat.

Normalization Procedures

One of the challenges with EMG analysis is that the amplitude of the data is highly variable and influenced by a number of factors. Measurements can differ across muscle sites, and intrasubject day-to-day fluctuations also contribute to the variability. Although it is not the purpose of this article to review EMG normalization procedure, a basic understanding of the process and the procedures used in the studies reviewed justifies comment.

One method of accounting for this variation is to normalize the EMG data against a reference value. The most common normalization method is to express the EMG activity under investigation as a percentage of the EMG activity during a maximal isometric voluntary contraction (MVIC), usually performed at the start of test session.

The majority of the studies reviewed (3,12,13,23,25, 28,32,33,36) used an MVIC procedure to normalize EMG data. In studies with repeated measures within the same individual, mean EMG was reported without normalization (2,27). Recent evidence (1) has been published to support a dynamic method of normalization (i.e., EMG data collected while cycling at 70% peak power) as being more repeatable than maximal voluntary contraction (MVC). Dankaerts et al. (8) also showed that normalization with submaximal voluntary contractions was more reliable than normalization with MVC when measuring TMA. In his recent review, Burden (4) states that using a dynamic MVC with the same muscle action is better than an isometric MVC at an arbitrary angle to normalize the EMG for a dynamic test effort.

Muscle Activation and the Loaded Back Squat

An inclusion criterion for this review was that the exercise under investigation was the free barbell back squat as this is the most widely practiced version of the loaded squat (12), especially in strength training for athletic performance. Given this popularity, it is not surprising that this version of the squat is used in most research where muscle activation is investigated to confirm commonly held coaching theories. A summary of research design and key findings for all reviewed studies is presented in Table 2.

Stance Width and Hip Rotation

It is a long-held belief in coaching that by manipulating squat stance width, specific muscle groups in the lower limb can be targeted, thereby influencing training adaptation. For example, in track sprint cycling, there is a strong belief that stance width in the squat should be equal to the width between the bicycle pedals and that feet should be parallel as they are when pedalling. The 2 studies (22,24) that measured lower limb muscle activation in 3 different stance widths for 2 submaximal relative squat loads had one common finding; lower limb muscle activation increased as a result of the increase in load. Paoli et al. (24) subjects perform the squat at loads of 0, 30, and 70% 1 repetition maximum (RM) with feet at hip width (greater trochanter width), 150 and 200% of hip width. They reported an increase in activation for all 8 muscles monitored with each increment in load; however, the gluteus maximus was the only muscle that had increased activation as stance width increased. These differences were only significant at the lowest loads (0% 1RM) and heaviest

TABLE 1. A summary of subjects and electrode placements in all studies reviewed.*

		Quadriceps			Harmstrings		Calf/shank			Thigh and hips					Anterior trunk			Posterior trunk	
Paper	Subjects	VM	VL	RF	BF	ST	GN	TA	SO	GMA	GME	AM	AL	AMG	RA	EO	TRA	ULES	LSES
Paoli et al. (24)	6 men with 3 years of experience	x	x	x	x	x				x	x	x							
McCaw and Melrose (22)	9 male lifters with 7 years of experience	x	x	x	x					x			x						
Caterisano et al. (5)	male strength trainees with 5 years of experience	x	x		x					x				x					
Zink et al. (36)	14 healthy men currently in weight training		x		x					x				x				x	
Wretenberg et al. (35)	14 national power and strength trainees		x	x	x														
Nuzzo et al. (23)	9 men with squat to BW ratio of 1.78														x	x		x	x
Hamlyn et al. (13)	8 men and 8 women with weight training experience														x	x		x	x
McBride et al. (20)	9 men currently in weight training	x	x		x		x												
McBride et al. (21)	men with squat to BW ratio 1.53		x		x														x
Willardson et al. (33)	12 men, average squat 1RM 135 kg														x	x	x	x	
Bressel et al. (3)	men with squat:BW ratio 1.6														x	x	x	x	
Smilios et al. (28)	16 men, average squat 1RM 129 kg	x	x	x	x														
Gullett et al. (12)	9 men and 6 women with ≥1 year of squat experience	x	x	x	x	x													
Schwanbeck et al. (27)	3 men and 3 women with 2–5 years of experience in strength	x	x		x		x	x							x				x
Anderson and Behm (2)	14 men, 7.8 years of experience with weights		x		x				x						x			x	x
Wilk et al. (32)	men with an average of 11 years of lifting experience			x	x		x												
Pick and Becque (26)	16 men: 8 years of experience and 8 <2 years of experience	x	x																
Pereira et al. (25)	5 men and 5 women			x								x							
Total		8	12	7	12	2	3	1	1	4	1	2	1	2	6	4	2	6	5 79

*AL = adductor longus; AM = adductor major; AMG = adductor magnus; BF = biceps femoris; BW = body weight; EO = external oblique; GMA = gluteus maximus; GME = gluteus medius; GN = gastrocnemius; LSES = lumbosacral erector spinae; RA = rectus abdominus; RF = rectus femoris; RM = repetition maximum; VL = vastus lateralis; VM = vastus medialis; SO = soleus; ST = semitendinosus; TA = tibialis anterior; TRA = transversus abdominus; ULES = upper lumbar erector spinae.

Table 2. A summary of the design and key findings of the reviewed studies.*

Study	Design	Key finding
Paoli et al. (24)	3 stance widths at 3 loads 0, 30, and 70% 1RM	EMG increased with load, gluteus maximus; EMG increased with stance width at 0 and 70% 1RM
McCaw and Melrose (22)	3 stance widths at 65 and 75% 1RM	Quadricep actvation increases with load, adductor longus and gluteus maximum EMG increase with stance width
Caterisano et al. (5)	3 squat depths at 0–125% body mass	Only gluteus maximus activation increases with squat depth
Zink et al. (36)	Weight belt vs. NWB for 1 repetition at 90% 1RM	Weight belt use does not affect trunk and leg muscle activation compared with no weight belt
Wretenberg et al. (35)	Squat to parallel vs. full depth at 65% 1RM	No difference in activation for parallel compared with full squats, but powerlifters produce higher EMG than strength trainees
Nuzzo et al. (23)	Stability ball exercises vs. squat and deadlift at 50, 70, 90% 1RM	Squat (and deadlifts) produces equal or more TMA than stability ball exercises
Hamlyn et al. (13)	Squat and deadlift at 0 and 80% 1RM vs. isometric unstable trunk exercises	80% 1RM squat produced more activation of LSES than stability ball exercises
McBride et al. (20)	Stable vs. unstable isometric squat	Unstable isometric squats reduce force, RFD, and agonist muscle activity with no increase in synergistic activity
McBride et al. (21)	Stable vs. unstable dynamic squat at 70, 80, and 90% 1RM and 3 fixed loads	Stable squats produce equal or greater activation of the VL, BF, and erector spinae than unstable squats
Willardson et al. (33)	Stable vs. BOSU squat at 50% 1RM and stable squat at 75% 1RM	TMA at 50 and 75% 1RM: stable squats were equal or greater than for 50% squats on BOSU
Bressel et al. (3)	Stable vs. BOSU squat at 50% 1RM and stable squat at 75% 1RM and verbal cue	Verbal cue resulted in increased TMA compared to squat without cue at the same load
Smilios et al. (28)	4 × 20 repetitions at 50% 1RM, then 4 maximum repetitions at 40 and 80% 1RM at 3 and 30 min	Muscle activity increases to compensate for metabolic fatigue
Gullett et al. (12)	Front vs. back squat 70% 1RM tested separately	No difference in muscle activity between the front and back squat at the same relative load
Schwanbeck et al. (27)	8RM smith machine squat vs. barbell back squat	Overall muscle activation was 43% greater for the free weight squat compared with Smith machine
Anderson and Behm (2)	Smith machine vs. free barbell squat vs. free barbell on balance disks at body weight and 60% 1RM	Unstable squat resulted in increased TMA but reduced activation of the prime movers
Wilk et al. (32)	12RM squat vs. leg extension vs. leg press	Highest EMG for the quadriceps and hamstrings occurred in the squat compared with the leg press and extension
Pick and Becque (26)	Squat 1RM and repetitions to failure at 85% 1RM, trained vs. untrained subjects	Trained subjects produced higher EMG during back squat 1RM test and repetitions to failure at 85% 1RM
Pereira et al. (25)	External hip rotation 0°, 30°, and 50° for RM parallel squat	Hip rotation did not affect quadriceps activation in any phase of a 1RM squat, hip adductor activation was greater at 30° external rotation

*BF = biceps femoris; LSES = lumbosacral erector spinae; NWB = no weight belt; RM = repetition maximum; VL = vastus lateralis.

load (70% 1RM). McCaw and Melrose (22) also used the 3 stance widths; shoulder width and 75 and 140% of shoulder width and 2 test loads; and 65 and 75% squat 1RM. On average, EMG of the quadriceps (rectus femoris, vastus lateralis, and vastus medialis) was 20% higher for the 75% 1RM for all stance widths and phases of the squat compared with the 60% 1RM. Interestingly, stance width did not affect muscle activation of the quadriceps contrary to the popular belief. Activation of the adductor longus was 28% greater for the heavier load and was highest during the ascent in the wide stance squat. Similarly, the activation of the gluteus maximus during the ascent was double that compared with the descent.

Pereira et al. (25) compared squatting with parallel depth while the hip was in neutral position and rotated to 30° and 50°. Ten subjects performed a RM parallel squat in the 3 hip positions, and muscle activity of the rectus femoris and the hip adductors (adductor longus and gracilis) was recorded. Muscle activity of the hip adductors was significantly greater with the rotation of the hips from neutral to 30° and from 30° and to 50° only in the last 30° of flexion to parallel and the first 30° of extension or ascent. Hip rotation did not change activation of the rectus femoris, the main extensor muscle in the squat. All muscle activity was significantly greater in the deepest phase (last 30°) of the squat in flexion and extension regardless of hip rotation.

Extreme stance widths of 40% wider than shoulder width (22) or twice the width of the hips (24) result in greater activation of adductors of the thigh and gluteus maximus. High adductor activation is also increased by turning the feet out or rotating the hips (25). The studies by Paoli et al. (24) and McCaw and Melrose (22) demonstrate that activation of quadriceps, the agonist, increases as a result of an increase in external load in the squat. Therefore, it may be concluded that if the purpose of the squat exercise is to overload the primary movers, then the stance, width of feet, and hip rotation should be dictated by the technical demands of executing the squat safely and to the appropriate or selected depth. These guidelines will vary according to individual physical mechanics and are covered by strength coaching principles. Central to this is ensuring that the vertical travel of the load in the sagittal plane remains in the line of gravity. In applied terms, this means that from a side view (sagittal plane), the vertical path of the bar during the squat is kept close to a perpendicular line emanating from the middle of the foot throughout the range of movement.

There is also evidence (25) that regardless of foot position, activation is highest in the last phase of the descent to parallel and the first phase of the ascent. This means that partial or quarter squats will result in reduced muscle activation of the prime movers and therefore arguably produce an inferior training effect in comparison to parallel or full squats. Although the loads used in these studies are representative of those used in athletic strength training, it is also common to train at higher relative loads where it would be unlikely that the widest stance widths in these studies would be practical or safe.

Squat Depth

A range of squat depths are used practically; however, it is generally believed that a squat depth to parallel is most effective for improving athletic performance (6). This belief is supported by a study (5) where the activation of muscles of the quadriceps, hamstrings, and buttocks for squats to 3 depths, partial, parallel, and full with knee angles of 135, 90 and 45°, respectively, was measured. Caterisano et al. (5) used a load of between 0 and 125% body weight because of the difficulty of standardizing the load across the 3 depths and found that activation of the gluteus maximus increased with the increase in squat depth. Increased muscle activation with squat depth was only found in the gluteus maximus in the full squat (35% mean integrated EMG) compared with 17% in the partial squat. No differences were found across the squat depths for the remaining 3 thigh muscles; biceps femoris, vastus medialis, and vastus lateralis. A limitation of this study was the selection of the test load as this would have represented very different relative loads for the squat at each depth. For example, a load that may have been a moderately high load for the full squat would have been a light load for the partial squat in the same individual.

Given the findings in the studies looking at stance width (22,24) where an increase in absolute load resulted in increased activation, one would expect that if Caterisano et al. (5) had managed to test using a relatively equivalent load for each squat depth, then they may well have found a difference in the activation for each depth of squat. A possible solution would be to determine a 1RM for the squat for each depth and then test muscle activation at the same relative submaximal percentage for each depth.

Wretenberg et al. (35) measured the activation of the vastus lateralis, rectus femoris, and the long head of biceps femoris in powerlifters ($n = 6$) and strength trainees ($n = 8$) with a combined average 1RM of 200 kg for the full squat. All subjects performed both parallel squats and full squats with the bar in the high position for the strength trainees and low position for the powerlifters. Mean peak muscle activation for all muscles was higher in the powerlifters for both squat depths, although this was only significant for the rectus femoris. Wretenberg et al. (35) found no difference in muscle activation across the 2 depths of the squat.

The authors suggested that this was because of their greater body mass and absolute mass lifted in testing for the powerlifters. The average body mass of the powerlifters was 5.7 kg greater than the strength trainees, and the powerlifters' average full squat 1RM was 0 kg greater than for the strength trainees. This meant that average test load of 65% of the 1RM represented 123% of the body mass for the strength trainees compared with 190% for the powerlifters. The average test load for the powerlifters was 65 kg heavier than that used for the strength trainees while the body mass difference was only 5.7 kg in favor of the powerlifters. Hence, the greater overall activation found in the powerlifters is probably explained entirely by the greater absolute loads lifted by them compared with the strength trainees rather than the difference in body mass. However, if we ignore the comparison between powerlifters and strength trainees, the pooled data indicate that, at a relative submaximal load, there was no difference in muscle activation of the muscles of the upper leg, including the prime movers, between squatting to parallel or full depth.

The real question in applied strength training for performance in sport is whether loaded squats performed to different depths at the same relative load result in different muscle activation and therefore different training stimuli. By using the same absolute load, Caterisano et al. (5) failed to answer this question; however, their results did give some indication that greater depth increases activity of the gluteus maximus. In the study of Wretenberg et al. (35), there was no difference in activation between parallel and full squats for well-trained subjects at a submaximal load. Pereira et al. (25), however, divided the eccentric and concentric movements into 30° segments and found that activation was highest for the deepest segment of both phases in a RM parallel squat. This suggests that depth of squat does impact on muscle activation. The results of the study by Wretenberg et al. (35) suggest that absolute load lifted per kilogram body mass had an impact on activation regardless of the depth of the squat. Both these studies focussed exclusively on the muscles of the thighs; it would be of significant interest to assess the impact of squat depth on TMA.

Weight Belt

The use of a supportive abdominal belt or weight belt is common in heavy lifting in weight training and in manual handling tasks. Zink et al. (36) conducted a study to measure the impact of squatting at a high relative load (90% 1RM) with or without a weight belt on joint kinematics and muscle activity of the following muscles: vastus lateralis, biceps femoris, adductor magnus, gluteus maximus, and erector spinae. There were no significant differences for mean EMG and time to peak EMG for any of the muscles in the concentric or eccentric phase of a parallel squat at 90% 1RM with a weight belt compared with no weight belt. Squatting with a weight belt, however, was significantly faster, for the total movement and for each phase separately, than when performed without a weight belt.

It appears that when stability is enhanced by the use of a weight belt, squat performance in terms of the velocity of movement and bar kinematics is improved. The authors of this research concede that although this may possibly represent an opportunity for the development of more work and power, at the same time, this may undermine the training effect resulting from slower and more stable training.

External Load

The loads used for the free barbell squat tests in the studies reviewed ranged from 0 to 90% of 1RM and included loads determined as a percentage of body mass and according to the RM method (Table 2). Wretenberg et al. (35) conducted a study to assess the effect of squatting with the high bar position compared with a low bar and whether this was different for full squats compared with the parallel squat. They used highly trained subjects who performed squats at 65% of their full squat 1RM. The loads reported were representative of the submaximal loads commonly used in strength training for the development of dynamic athlete performance. Two research groups used near maximal loads of 90% of 1RM. Zink et al. (36) used parallel squats at this intensity to assess the impact of a weight belt vs. no weight belt on muscle activation. Nuzzo et al. (23) used a range of loads: 50, 70 and 90% of 1RM in the squat and deadlift, to assess muscle activation in comparison to 3 stability ball exercises.

A methodological challenge facing investigators comparing 2 or more variations of the squat is how to determine a relatively equal test load for each variation to ensure that the dependent variable is the variation of the squat and not the load. Wretenberg et al. (35) and Caterisano et al. (5) failed to account for this in their studies comparing squats at different depths. The former used a load of 65% of the 1RM for the full squat for both test depths, full and parallel squat. Caterisano et al. (5) overcame this by selecting a test load which the subjects could complete for the 3 squat depths with the correct technique. If we assume that squats performed to different depths each represent a different physical challenge, then the test load for each depth should be based on the same submaximal relative percentage calculated from the maximal ability for each depth. Wilk et al. (32), Pereira et al. (25), and Schwanbeck et al. (27) overcame this challenge by using 12, and 8 RM, respectively, for each of the test variations. This is achieved by determining the maximal load which the subject can complete for the given number of repetitions, and in the study of Schwanbeck et al. (27), this was determined for the free bar back squat and the Smith machine back squat. Possibly, the most accurate theoretical method is to determine the 1RM for each of the squat variations and then calculate the submaximal test load for each as a percentage of the maximum (1RM). Gullett et al. (12), in a biomechanical comparison for the front and back squats, tested 1RM for each of these exercises, and McBride et al. (21) determined both stable and unstable squat 1RM.

There is an indication that increments in load for the same squat variation have an impact on muscle activation (22,24,35). Therefore, the test loads should be relatively equal if the research question is whether muscle activation is affected by a specific squat variation.

Instability

Although the squat is an established method of developing strength through the whole body, evidence that it is an effective method of developing core stability or trunk strength is relatively new (13,23). The concept of using instability as part of the protocol, whether it be for fitness for health, conditioning for sport, or exercise for rehabilitation, is based on the principle of core stability (15,19). Initially, the recommendations for core stability training were to isolate the contraction of the deep stabilizing muscles of the trunk (16,17). The use of unstable training surfaces was introduced in the belief that this would increase the challenge placed on these stabilizing muscles. As a consequence, a number of studies (2,20,21,33) have been conducted to assess the impact of instability during the squat on muscle activation. There have also been studies comparing muscle activation in the squat performed on a stable surface and selected unstable trunk exercises (13,23). For example, Anderson and Behm (2) reported greater muscle activation in the key muscles of the thigh and trunk when performing squats on 2 balance disks compared with a Smith machine. Hamlyn et al. (13) compared the TMA during squats and deadlifts at 80% of 1RM with 2 trunk strengthening exercises performed on an unstable surface. The results showed that first, the squat produced significantly greater activation of the lower sacral erector spinae than the other 3 exercises, and second, the deadlift resulted in greater activation of the upper lumbar erector spinae. Nuzzo et al. (23) showed that TMA in squats and deadlifts was greater than or equal to that found in 3 stability ball exercises in male subjects with an average squat 1RM to body mass ratio of 1.78 (21). The latter 2 groups (13,23) concluded that upright free weight training on a stable base was effective in challenging and developing trunk stability through effective TMA. Finally, McBride et al. (20,21) assessed the impact of instability on force and muscle activity (vastus lateralis, biceps femoris, and erector spinae) in both isometric (20) and dynamic squats (21). They found that instability in an isometric squat significantly impaired force and power capabilities without an advantage for muscle activity. Unstable squat 1RM (83 kg) was significantly lower (44 kg) than the stable squat 1RM (128 kg), and muscle activity at the same relative loads was equal or less in the unstable trial compared with the stable squat. McBride et al. (20,21) concluded that stable squats were more effective in producing force, power, and muscle activation, including TMA, than unstable squats.

Anderson and Behm (2) assessed the parallel squat under 3 conditions of stability: in a Smith machine, with a free bar, and with a free bar standing on balance disks. Three loads were tested for each condition: body mass, 29.5 kg (the load of the bar in the Smith Machine), and 60% of body mass. They found that muscle activation in the muscles of the legs and the trunk stabilizers was highest during the most unstable squat with the free bar on the balance disks. They also report an increase in activation with the increase in load for all muscles apart from the hamstrings and abdominal stabilizers. Furthermore, they showed that EMG activity was highest during the concentric movement compared with the eccentric phase for all squat variations in the following muscles: soleus, vastus lateralis, biceps femoris, upper lumbar erector spinae, and abdominal stabilizers. Duration of activity of the abdominal stabilizers during the transition between descent and ascent was significantly higher for the unstable squat.

Although the purpose of many of the studies referred above was not primarily to measure and describe TMA in the loaded squat, they all showed that this exercise is an effective method of challenging the trunk stabilizers. There is also evidence that the introduction of instability impairs force and power production in the squat without necessarily increasing TMA.

Acute Fatigue

Strength training in the practical setting is usually performed with a certain amount of acute muscular fatigue, and as such the effect of this on muscle activation is of interest. Smilios et al. (28) measured the power output and EMG activity during a moderate load muscular endurance session (4 sets of 20 repetitions at 50% 1RM). They measured power output and EMG in a set of 4 repetitions at a light load (40% 1RM) and heavy load (80% 1RM) immediately before and after the endurance sets and again at 30 minutes after this. The subjects, who were resistance-trained men with an average squat 1RM of 129 kg, performed the parallel back squat. Power output was significantly reduced in sets 3 and 4 of the muscular endurance protocol, although EMG activity increased from set to set for the quadriceps but not the biceps femoris muscles.

Average power immediately after the endurance work was reduced by 14 and 21% for the power tests at 40 and 80% 1RM. These improved but remained 8 and 14% lower for the 2 power test loads, respectively, after 30 minutes of recovery. Average quadriceps EMG activity at 3 and 30 minutes post endurance effort was decreased by 12 and 14% for the 40% 1RM power tests and 6 and 10% for the 80% 1RM power tests. Biceps femoris EMG activity did not change in the 40 and 80% 1RM power tests performed at 3 and 30 minutes after the endurance protocol.

The authors hypothesize that the increase in EMG activity of the agonists during the endurance sets, despite the loss of power across the sets, was because of an increased central drive to maintain work though increased motor unit recruitment (28).

Front Vs. Back Squat

A number of variations of the loaded barbell squat are used in programs for the development of athletes. The 2 most common versions are the back squat, where the bar is carried across the back of the shoulders, and the front squat, with the bar across the front of the shoulders. There is a belief that this technical difference produces a different physical challenge and therefore training effect. This would be reflected by a difference in activation of the primary muscles across these 2 squats.

Subjects performed 2 trials of 3 repetitions for each squat variation: front and back squat at 70% of the 1RM (12). The 1RM scores were determined for each squat on 2 separate prior occasions. During the investigative efforts, EMG activity for the following muscles was recorded: rectus femoris, vastus lateralis, vastus medialis, biceps femoris, semitendinosus, and erector spinae. These authors found no difference in muscle activity across the 2 squat variations; however, they did show that average muscle activity was significantly higher during the ascent than the descent for the 6 muscles that were monitored.

Using a test load of 70% of 1RM for each of the 2 squat variations meant that the absolute load lifted was 61.8 ± 18.6 kg for the back squat and 45.8 ± 14.1 kg for the front squats. Muscle activity was not found to be different for the front squat compared with the back squat at 70% 1RM. The common belief in coaching is that these 2 exercises offer different physical challenges primarily because of the difference in the position of the load in relation to the line of gravity throughout the movement. It is possible that the submaximal load used in this experiment failed to elicit this difference; however, this is surprising given the low level of squat training exposure reported for subjects in this study.

Free Bar Vs. Smith Machine Squat

Schwanbeck et al. (27) compared a free bar squat with a squat performed in a Smith machine to assess muscle activity in the legs (tibialis anterior, gastrocnemius, vastus lateralis, vastus medialis, and biceps femoris) and in the trunk (lumbar erector spinae and rectus abdominus). The test load for each exercise was determined as an 8RM as this load represented a common method of determining training intensity for athletic conditioning. This method resulted in the test loads being 14–23 kg heavier for all subjects for the Smith machine squat than the free bar squat.

Despite this, the free bar squat elicited 43% higher average activity for all muscle groups than the Smith machine squat. Closer inspection of individual muscles shows that only 3 leg muscles had significantly higher activation in the free bar squat, gastrocnemius (34%), biceps femoris (26%), and vastus medialis (49%) compared with the Smith machine squat. The muscles of the trunk followed this trend but failed to reach significance. The authors claim that this was because of the low number of subjects ($n = 6$) in the study.

Squat Vs. Leg Extension and Leg Press

Wilk et al. (32) compared the muscle activity of the quadriceps and hamstrings in 3 exercises: the squat, the leg press, and the leg extension. They found that the highest activation occurred in the closed kinetic chain squat compared with both the leg press and leg extension and that this was significant for the vastus lateralis, medial, and lateral hamstrings. The maximal activation of the quadriceps presented at a knee angle between 88° and 2° during the concentric phase. For the hamstrings, the peak activation occurred between 60° and 74° knee flexion also during the concentric movement. This confirms the belief that squatting with a free external load represents a greater neural challenge to the prime movers than exercises that isolate the limb as in the leg press and leg extension exercises. It has been suggested that this may be because of the increased demand to stabilize the free bar load in the squat; however, this would not necessarily present as increased EMG of the prime movers but rather the stabilizing muscles. The higher activation in the free bar squat may well be because of the fact that the load is lifted vertically against gravity compared to during the machine leg exercises where the load is applied via levers.

Training Status and Muscle Activation in the Squat

All the studies reviewed so far have compared muscle activation for variations of the back squat in a range of subjects including those untrained, moderately trained, and well trained in the back squat and of both genders (Table 1). Pick and Becque (26) reported muscle activation of 2 primary movers in the squat, vastus lateralis and vastus medialis, for a back squat set to failure at 85% 1RM in trained (1RM = 184 kg) and untrained male subjects (1RM = 120 kg).

They found that the trained subjects (± 0.9 repetitions) completed significantly more repetitions at 85% 1RM than the untrained group (7 ± 0.7 repetitions) and therefore demonstrated greater relative submaximal lifting capacity. Muscle activity was recorded during 1RM testing, and EMG during the repetitions to failure was reported as a percentage of 1RM EMG. This was higher in both the 1RM test and the repetitions to failure for the trained compared with the untrained group for both the individual muscles and combined data. Of particular importance was that this difference was significant toward the end of the test, at 80 and 0% of repetitions to failure.

The study is characterized by the high relative back squat 1RM of 1.6 of body mass reported for an untrained group and the fairly marked difference in body mass between the 2 groups; the untrained group (74.8 kg) was 15.1 kg lighter than the trained subjects (89.9 kg). This meant that the test load difference on average was 54 kg heavier for the trained group than the untrained group. It appears that the difference in absolute load may explain the difference in the EMG measured during both the 1RM test and repetitions to failure.

Summary

- Increasing stance width and hip rotation increase activation of the adductors and gluteus maximus and not the primary movers of the squat exercise.
- Muscle activation is not different in squats to varying depths at moderate loads. Within a parallel squat, it appears that activation is greatest in the last phase of the descent and the first phase of the ascent.
- Activation of the muscles of the legs and trunk increases as a consequence of increases in absolute external load.
- It is important to use equal, relative, submaximal test loads calculated from a maximum for each specific squat if the aim is to measure the differences in activation between 2 types of squat or squat variation such as depth of squat.
- Muscle activity is not influenced by the use of a weight belt.

- Squatting on an unstable base increases the activation of the leg and trunk muscle, but it impairs force and power production in this exercise.
- The squat at moderate external loads is a more effective method of activating the trunk stabilizers compared with other instability trunk exercises.
- Acute fatigue in a submaximal squatting task results in increased muscle activation corresponding to a loss of power across the task. Power and EMG is reduced for up to 30 minutes in a low and high load power test.
- Muscle activation in muscles of the leg, thigh, and trunk is the same in the front and back squat at 70% of 1RM.
- Free bar squat elicits higher overall EMG than squats in a Smith Machine, leg press, and leg extension.
- Highest activation occurs in the concentric phase of the squat.
- In sets to failure at 85% 1RM, trained subjects completed significantly more squat repetitions than untrained subjects and produced higher muscle activation in both the 1RM test and the set of repetitions to failure.

PRACTICAL APPLICATIONS

The free barbell back squat is superior to more supported squats performed in a Smith machine and closed kinetic chain leg exercises in activating the prime movers. There is also evidence that the level of activation of the agonist muscles is increased with the increase in absolute external load. It is also clear that many technical alterations to the squat, including stance width, hip rotation, and the use of a weight belt, do not enhance the activation of the prime movers. At moderate loads, even fairly significant alterations, as found in the front squat, do not alter the activation of the prime movers compared with the back squat. These data suggest that increases in load are largely responsible for increased activation. Also the concentric phase produces the highest activation and within the eccentric phase, the last third of the descent to parallel elicits the highest activation. Therefore, if the aim is to increase the strength of the known prime movers, the technique should ensure effective completion of the squat to parallel at the desired load.

At loads of greater than 50% of 1RM, the back squat to parallel is an effective method of developing trunk activation and therefore arguably trunk stability. The application of the loaded squat for the development of trunk and core stability is an area for future research.

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Study 2: A survey of contemporary perspectives of core stability training

The review confirmed that loaded barbell squat was effective in activating the trunk muscles. The next question was to determine the extent to which these developments in the scientific process and literature had entered applied thinking and practice. This was the justification and purpose of the survey, the second section of the thesis. The background section of the survey presents an in-depth review of the scientific challenge to traditional CST for healthy and athletic populations. Followed by a detailed explanation of the purpose of the survey and how questions were built around prominent themes in the scientific CST debate (Appendix 1). Furthermore, in the survey discussion, findings were analysed against the prevailing issues in the scientific literature to measure the extent to which research had influenced applied perceptions and practice.

The limitations of the survey are covered in the final paragraph of the discussion in the published paper. However, it is worth pointing out the further potential bias that may have arisen as a result of recruiting participants from the principle investigators email contacts and LinkedIn connections. This most certainly contributed to the high number of respondents from strength and conditioning.

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ORIGINAL RESEARCH ARTICLE **Open Access**

Contemporary perspectives of core stability training for dynamic athletic performance: a survey of athletes, coaches, sports science and sports medicine practitioners

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Abstract

Background: Core stability training has grown in popularity over 25 years, initially for back pain prevention or therapy. Subsequently, it developed as a mode of exercise training for health, fitness and sport. The scientific basis for traditional core stability exercise has recently been questioned and challenged, especially in relation to dynamic athletic performance. Reviews have called for clarity on what constitutes anatomy and function of the core, especially in healthy and uninjured people. Clinical research suggests that traditional core stability training is inappropriate for development of fitness for heath and sports performance. However, commonly used methods of measuring core stability in research do not reflect functional nature of core stability in uninjured, healthy and athletic populations. Recent reviews have proposed a more dynamic, whole body approach to training core stabilization, and research has begun to measure and report efficacy of these modes training. The purpose of this study was to assess extent to which these developments have informed people currently working and participating in sport.

Methods: An online survey questionnaire was developed around common themes on core stability training as defined in the current scientific literature and circulated to a sample population of people working and participating in sport. Survey results were assessed against key elements of the current scientific debate.

Results: Perceptions on anatomy and function of the core were gathered from a representative cohort of athletes, coaches, sports science and sports medicine practitioners ($n = 241$), along with their views on effectiveness of various current and traditional exercise training modes. Most popular method of testing and measuring core function was subjective assessment through observation (43%), while a quarter (22%) believed there was no effective method of measurement. Perceptions of people in sport reflect the scientific debate, and practitioners have adopted a more functional approach to core stability training. There was strong support for loaded, compound exercises performed upright, compared to moderate support for traditional core stability exercises. Half of the participants (50%) in the survey, however, still support a traditional isolation core stability training.

Conclusion: Perceptions in applied practice on core stability training for dynamic athletic performance are aligned to a large extent to the scientific literature.

Keywords: Core, Stability, Dynamic, Trunk, Athletic, Performance, Loaded, Functional, Compound, Exercise

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Key points

- Core stability training for healthy and athletic populations has recently been questioned and challenged in scientific literature. The narrow definition of both the anatomy, spinal region between pelvis and diaphragm, and the method of training the core through the isolation of muscles in this region does not relate to full body core function that characterises dynamic athletic performance.
- The survey reveals that this is reflected in opinions of people working and participating in sport. Half of the participants identified the area between and including the pelvic and shoulder girdles as the core. Majority supported functional loaded exercises such farmer's walk (87%) and barbell squats (84%) as effective exercises for the development of core stability.
- Despite the support for a more functional approach, selected traditional core stability training methods do retain a certain amount of support; isometric plank exercise (56%) and unstable stability ball exercises (41%). Many respondents (42%) felt that core function should be measured subjectively through observation of sporting and or exercise performance.
- Trunk is the preferred name of the anatomical region for almost half (45%) the participants while 35% supported the term core.

Background

The absence of a universally accepted definition of core stability (CS) is well noted in the scientific literature [1–8]. A number of these publications have proposed a definition, focussing either on function, anatomical constituents of the core or both. Several reviews have questioned and challenged core stability training (CST) for prevention and treatment of back pain [9–11] and for improvement of function and performance in healthy and athletic populations [1, 5–7, 12–14]. There is a view [1, 7] that CST in its current form evolved from clinical research [15] in the 1990s. The application of a clinical exercise approach in healthy and athletic populations has been criticised, primarily on the basis that teaching an isolated muscle pattern in uninjured athletes is unfounded [6, 10, 16]. Despite this, CST as an intervention spread to all exercise disciplines across clinical, fitness and sports performance settings with significant commercial interest and support [14].

Most review articles on this topic recognised that the application of traditional CST in healthy and athletic groups lack scientific justification [3, 7, 14, 17]. This resulted in a body of research investigating CST in healthy populations [18–22] along with aforementioned review articles [1, 6, 7, 12–14]. Reviewers have noted that research cannot progress this topic effectively until there is a standardised agreement on the anatomical structure and function of the core [1, 6, 7]. A further limitation reported by most reviewers is the absence of a valid and reliable test of core function [1, 12]. As a result most research on the topic is methodologically limited [12, 13] and therefore ineffective in confirming or challenging the concept and practice of CST for health and performance. A case has been made in the literature for a more functional definition of anatomy of the core, applicable to healthy and athletic populations [1, 8]. Similarly, it is proposed that the description of core function is revised to encompass normal healthy and athletic human movement [8].

Several comprehensive reviews over the last decade have examined the research on the effectiveness of various CST methods for athletic performance [1, 6, 7, 12–14]. Reviews covered the variations in CST including instability training, trunk rotation exercises, functional training and exercise intensity. Martuscello et al. proposed a five core exercise classification system based on their review of the research [6]. The categories were traditional core exercise (sit-ups), core stability exercises (isometric plank), ball or device exercises (stability ball), free weight exercise (squat and deadlift) and noncore free weight exercise (upper body). In a recent study conducted in an applied performance sport setting, Spencer et al. proposed a comprehensive spinal exercise classification [2]. The classification incorporated static and dynamic exercises that were either functional or non-functional according to spinal displacement across four physical outcomes: mobility, motor control, work capacity and strength. Both studies [2, 6] clarify the range and nature of core stability exercises used in the literature and practice; however, there is concern that many core stability intervention studies are diluted by other exercises and activities preventing a clear assessment of impact of CST [7, 12, 13]. Furthermore, in athletic populations, a reductionist approach or selective activation to improve integrated function is unsubstantiated [1, 2, 7, 12].

The proposed protection against injury and improved athletic performance from CST has been the subject of many research studies and review papers. Silfies et al. concluded that following a review of 11 studies, there was limited evidence to support the use of CST to prevent upper extremity injury and improve athletic performance [3]. The authors questioned whether performance in core stability tests reflected physical or athletic capability and level of conditioning, rather than solely core stabilization. Tests included the isometric front and side bridge, single-leg raise [10], star excursion test [11] and closed kinetic chain upper extremity stability test [12]. A systematic review conducted by Prieske et al. [12] concluded that CST compared with no training or regular sports-specific training does improve trunk muscle strength measured predominantly by isometric plank. However, increases in trunk muscle strength only had a small effect on physical fitness and athletic performance measures in trained individuals. CST compared to alternative physical training methods in trained individuals had little impact on trunk muscle strength, physical fitness and athletic performance measures. Both studies strongly suggest that high levels of general fitness are associated with better performance in CS tests and therefore a lower risk of injury and better athletic performance test scores [3, 12].

Separating the core into smaller local and larger global muscles has little bearing on core stability for dynamic movement in healthy people. In Lederman's [10] words, this is an anatomical classification with no functional relevance. The role the core plays in stabilising the body is dynamic and responsive to many postural challenges that occur in normal movement and complex, reactive environment of sport [14]. The concepts of core strength and core stability have been reviewed the literature [1, 5, 23]. Whether these are separate attributes [5] or whether core strength is required for core stability [23] remain unresolved questions [1]. In this context, core stability is an integrated, functional motor task [7, 24] and training should reflect this according to movement patterns [14, 24], forces [7, 24] and torque and velocity [8, 24].

A limitation identified by Prieske et al. [12] was the lack of validity of tests used in most of the research. Trunk muscle strength in most studies was measured by timed isometric test (prone bridge) which, firstly, does not reflect force and velocity of movement of dynamic athletic activity [12]. Secondly, CST programmes in many of the studies incorporated prone plank or similar isometric exercises in the exercise intervention, which rendered timed isometric prone plank an inappropriate test of trunk muscle strength in these cases. Most reviews conclude there is not a valid method of measuring the effect of CST on trunk muscle strength within the context of improving dynamic athletic performance [1, 13, 14, 17, 25, 26]. As a result, many researchers have resorted to using conventional performance tests such as countermovement jump and sprint tests [12, 13, 27].

The first three levels of Martuscello's [6] core exercise classification system appear to contravene the established overload training principle [28] when applied to an athletic population. Traditional low load core exercises, minimal range or isometric core stability exercises and ball/device exercises are all characterised by low force, low velocity and restricted range of movement. Hence, these do not represent training overload in preparation for activities that characterise most sports and athletic events. Researchers have begun to investigate trunk muscle activation in a number of dynamic, loaded free weight exercises to determine their suitability for the development of dynamic trunk strength and stability [29–37]. Surface electromyography methodology shows there is good evidence that loaded exercises performed in a standing position are an effective method of overloading the trunk stabilization system in a dynamic manner. While several reviewers recognise this development [6, 7, 14], it is best summarised by Wirth et al. (2016), '… we recommend the use of classical strength-training exercises as these provide the necessary stimuli to induce the desired adaptations.'

The flawed foundations of CST for dynamic athletic performance have been exposed in the scientific literature. Research is underway to better understand the most effective training methods for the development of trunk stability. The aim of this survey is to assess the current perspectives of CST in the applied sports setting to determine how well scientific literature informs these opinions. Our hypothesis is that opinions of those who work and participate in sport will reflect scientific debate on key core stability training topics.

Methods

The online survey questionnaire (Additional file 1) was developed around common themes on core stability as defined in the current scientific literature. The online survey was created and distributed using Bristol Online Survey (BOS) tool (Tower Hill, Bristol, UK). The questionnaire comprised four sections: anatomy of the core, function of the core, methods of measuring core function and methods of training the core. The survey concluded with general questions about the application of core strength training for dynamic athletic performance.

The survey question on the anatomy of the core is based on definitions in the literature. We used the definition of local and global stabilization of intersegmental spine proposed by Bergmark (1989) [38]; the passive spinal column, active spinal muscles and neural control unit as described by Panjabi [39]; axial skeleton between pelvic and shoulder girdle including rib cage, spinal column and associated muscle and nerves proposed by Behm et al. [8]; and lumbo-pelvic hip complex according to Faries and Greenwood [23]. Categories of exercises and selection criteria for CST used in the survey question were drawn from published studies that investigated muscle activation using these manipulations. The question around core strength and core stability were based on reviews of this topic [1, 7].

A pilot survey was conducted using the postgraduate sports studies group ($n = 20$) at the University of Stirling. The questionnaire was modified according to feedback from the pilot survey. Approval for the study was granted by the local research ethics committee in accordance with the Helsinki Declaration (2013) [40].

Participants

The survey was circulated using two methods: shared with the principal authors' 700 LinkedIn connections and sent by email to 220 qualifying contacts. All recipients were asked to share the survey with all their contacts that met the criteria of working or participating in sport.

Statistical analysis

The data analysis was descriptive and frequency was presented in the tables as number and percentage (*n* (%)). Data presented in Figs. 1, 2, 3 and 4 were analysed using Kruskal-Wallis test to assess support for each statement on 5-point Likert scale. Data presented as mean and 95% CI. Five-point scale is as follows: 1 = strongly agree or very effective and 5 = strongly disagree or not effective at all. Significant differences were further analysed using Dunn's multiple comparison post hoc test. Priori alpha level of significance was set at $p < 0.05$.

Results

Participants

The online survey was completed by 241 respondents from a range of disciplines involved in sport (Table 1). The highest return by employment group was received from strength and conditioning coaches (S&CC; 47%) followed by athletes and players (A&P; 17%) and sport medicine practitioners and physiotherapists (SM&P; 17%). A quarter of the cohort were involved in sport at university or school level (27%). A similar number (33%) were working in professional sport, either with full-time professional athletes (21%), or elite funded athletes in institutes of sport (12%). Volunteers working in recreational sport made up 15% while 9% were semi-professional in part-time paid roles.

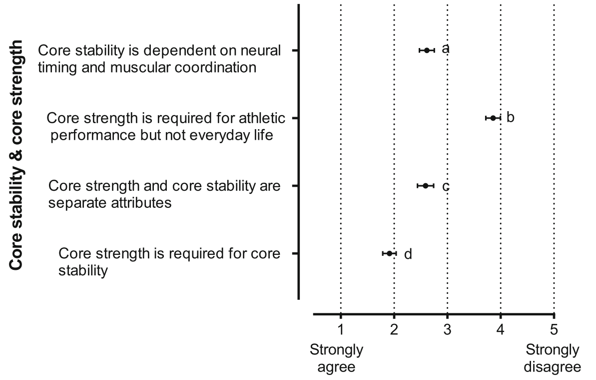

Fig. 1 Reported support for a series of statements relating to core stability and core strength. Data are reported as mean level of agreement with 95% CI. 1 = strongly agree, 5 = strongly disagree. Significant differences $p < 0.001$: a vs b, a vs d, b vs d and c vs d. CI: confidence interval

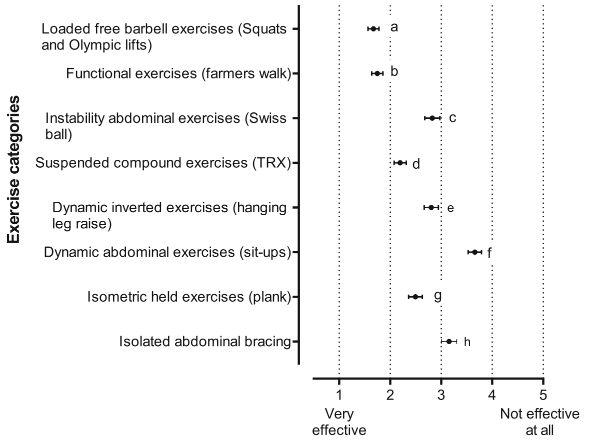

Fig. 2 Responses to a series of questions on the effectiveness of selected categories of exercise in developing core stability for dynamic athletic performance. Data are reported as mean level of effectiveness with 95% CI. 1 = very effective, 5 = not effective at all. Significant differences $p < 0.001$: a vs c, d, e, f, g and h; b vs c, d, e, f, g and h; c vs d and f; d vs e, f and h; e vs f; f vs g and h; g vs h. CI: confidence interval

Responses to all questions were analysed for all respondents ($n = 241$) and for each of the five demographic groups. There were no differences between group responses and total cohort, so data are presented and discussed for the total cohort.

The majority (87%) were qualified to degree level or higher, 40% had masters or MSc degrees and 12% had doctoral degrees. Most respondents (73%) reported to have a discipline specific professional qualification. Respondents reported to have been working in their specific discipline for an average of 8 years (range 0–36 years).

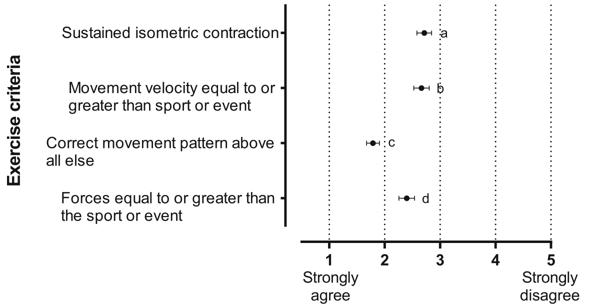

Fig. 3 Responses to which criteria should inform exercise selection for the development of core stability for dynamic athletic performance. Data are reported as mean level of agreement with 95% CI. 1 = strongly agree, 5 = strongly disagree. Significant differences $p < 0.05$: a vs c, a vs d, b vs c, b vs d and c vs d. CI: confidence interval

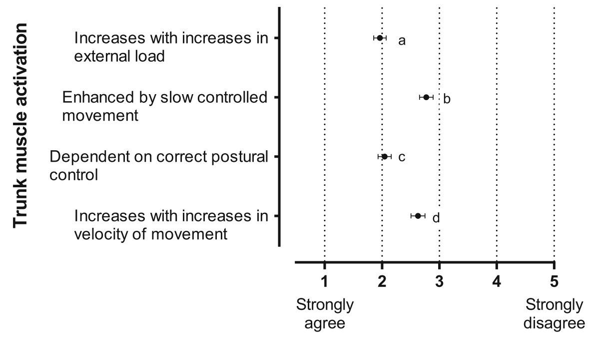

Fig. 4 Responses to a series of statements relating to ground-based loaded free barbell exercises and trunk muscle activation. Data are reported as mean level of agreement with 95% CI. 1 = strongly agree, 5 = strongly disagree. Significant differences $p < 0.001$: a vs b, a vs d, b vs c, c vs d. CI: confidence interval

Anatomy and name of the core

In response to the question on the anatomical region that comprised the core, half of the respondents (50%) identified the region between and including the pelvic and shoulder girdles and associated muscles and nerves (Table 2). Approximately, a quarter of respondents (27%) identified the region between the diaphragm and pelvic floor and associated muscles and nerves as the core, while for 18%, this was the lumbar spine, pelvis, hip joints and related muscles and nerves. Interestingly, more participants (45%) felt that the region should be called the trunk while 35% supported the term core and 18% preferred torso.

Methods of measuring core function

Respondents were asked to identify the most effective method of measuring core stability in a healthy, uninjured person. Almost a quarter (22%) reported that there was no effective method to test core stability. A number (43%) of the respondents proposed subjective assessment of core stability through observation. Of these, 17% suggested observation of sport-specific movement or exercise technique and 26%, observation of ground-based loaded barbell exercises. Objective assessments were proposed by 32% and included the timed isometric plank (19%), functional movement screen (9%) and isometric trunk bracing with biofeedback (4%).

Core function and core stability training

Core stability and core strength (Fig. 1)

The majority believed that core strength is required for stability (mean 1.9, 95% CI 1.8–2.0, $p < 0.001$) and far fewer agreed that these were separate attributes (mean 2.6, 95% CI 2.4–2.7, $p < 0.001$) (Fig. 1). Most participants disagreed with the statement that core strength was required for athletic performance, but not everyday life (mean 3.9, 95% CI 3.7–4.0, $p < 0.001$).

The effectiveness of certain exercise categories on CST (Fig. 2)

The exercise categories deemed most effective in developing core stability for dynamic athletic performance were (Fig. 2) squats and Olympic lifts (mean 1.7, 95% CI, 1.6–1.8, $p < 0.001$) and farmers walk (mean 1.7, 95% CI

Table 1 (A) Employment and (B) education information presented for all respondents (total and group)

	Total	S&CC	A&P	SM&P	SP&B	SC
All respondents	241	114 (47)	42 (17)	41 (17)	24 (10)	20 (8)
A.						
Academic, university or school sport role	*66 (27)*	29 (12)	10 (4)	*11 (5)*	*10 (4)*	*6 (2)*
Professional: full-time paid position, full-time paid athletes	50 (21)	*37 (15)*	0 (0)	9 (4)	3 (1)	1 (0)
Volunteer, recreational club sport	35 (15)	4 (2)	*21 (9)*	6 (2)	2 (1)	2 (1)
Elite professional: full-time paid position, funded, amateur athletes (Institute)	30 (12)	15 (6)	1 (0)	4 (2)	7 (3)	3 (1)
Elite non-professional, part-time, regional or national athletes	30 (12)	16 (7)	5 (2)	7 (3)	0 (0)	2 (1)
Semi-professional: paid part-time position	22 (9)	9 (4)	2 (1)	3 (1)	2 (1)	*6 (2)*
Other	8 (3)	4 (2)	3 (1)	1 (0)	0 (0)	0 (0)
B.						
MSc/Masters	*96 (40)*	*51 (21)*	7 (3)	*20 (8)*	*13 (5)*	5 (2)
Degree/Hons	84 (35)	41 (17)	*17 (7)*	9 (4)	7 (3)	*10 (4)*
PhD	28 (12)	10 (4)	2 (1)	10 (4)	4 (2)	2 (1)
Diploma	27 (11)	9 (4)	13 (5)	2 (1)	0 (0)	3 (1)
Other	6 (2)	3 (1)	3 (1)	0 (0)	0 (0)	0 (0)

Data presented as number and percentage (*n* (%)) of all respondents. Italics represent the highest response for the column
S&CC strength and conditioning coaches, *A&P* athletes and players, *SM&P* sports medicine practitioners and physiotherapists, *SP&B* sports physiologists and biomechanists, *SC* sports coaches

Table 2 Responses to the question of what (A) anatomic region makes up the core and (B) which term best describes this anatomical region

	Total
A.	
The spine and the associated muscles and nerves	5 (2)
The lumbar spine, pelvic and hip joints and associated muscles and nerves	43 (18)
The region between and including the pelvic and shoulder girdles and associated muscles and nerves	*120 (50)*
The region between and diaphragm and pelvic floor and associated muscles and nerves	65 (27)
Other	8 (3)
B.	
Torso	43 (18)
Trunk	*108 (45)*
Core	85 (35)
Upper limb	0 (0)
Other	5 (2)

Data presented as number and percentage (*n* (%)) of all respondents. Italics represent the highest response

1.6–1.9, $p < 0.001$). Conversely, support was moderate to low for traditional core stability exercises, namely suspended compound exercises (mean 2.2, 95% CI 2.1–2.3, $p < 0.001$), isometric plank (mean 2.5, 95% CI, 2.4–2.6, $p < 0.001$), hanging leg raise (mean 2.8, 95% CI 2.7–2.9, $p < 0.001$) and instability abdominal exercises (mean 2.8, 95% CI 2.7–3.0, $p < 0.001$). Participants identified two exercise categories that were more ineffective than effective; abdominal bracing (mean 3.2, 95% CI, 3.0–3.3, $p < 0.001$) and sit-ups (mean 3.7, 95% CI, 3.5–3.8, $p < 0.001$).

The exercise selection criteria for effective CST (Fig. 3)

Correct movement pattern (mean 1.8, 95% CI 1.7–1.9, $p < 0.001$) was identified as most important exercise selection criteria for development of core stability for dynamic athletic performance (Fig. 3). Exercises characterised by forces that were equal to or greater than the force in the sport or event, were supported by 60% of the cohort (mean 2.4, 95% CI 2.3–2.5, $p < 0.05$). Most were either undecided or disagreed on the importance of velocity of movement (mean 2.6, 95% CI 2.5–2.8, $p < 0.05$) and sustained isometric contraction (mean 2.7, 95% CI 2.6–2.8, $p < 0.05$) in core stability exercises for athletic performance.

Ground-based free barbell exercises and trunk muscle activation (Fig. 4)

Most participants agreed that increases in external load in standing barbell exercises would increase trunk muscle activation (mean 2.0, 95% CI 1.9–2.1, $p < 0.001$) (Fig. 4). Equally important in this form of resistance training was correct postural control (mean 2.0, 95% CI 1.9–2.2, $p < 0.001$). Slow controlled movement (mean 2.8, 95% CI 2.7–2.9, $p < 0.001$) and increases in velocity (mean 2.6, 95% CI 2.5–2. 8, $p < 0.001$) of strength training exercises were not seen as important in eliciting trunk muscle activation in ground-based free barbell exercises.

Finally, results for the general questions on the application of core stability exercises are presented on Table 3. Most participants (85%) felt that it was appropriate to include specific exercises to train core stability in healthy, uninjured individuals. Less than half (45%) felt that it was effective to exercise the core stabilisers in isolation, while a majority (65%) agreed that core stability is developed during normal progressive exercise training.

Discussion

Core stability training for healthy and athletic populations has been scrutinised and challenged in recent years in scientific literature [6, 7, 10, 13, 41–43]. Descriptions of the core by anatomic structures are entirely dependent on the chosen definition of core function [1]. The original narrow definition presented in early research focussed on the spinal region between the diaphragm and pelvis [44]. This approach identified muscular and neural dysfunction associated with back pain. Hence, core function was isolated to this region and proposed training intervention isolated the involved muscles. This approach did not transfer to healthy individuals and athletes where core function is obviously at the centre of dynamic movement characterised by force and velocity through the length of the body [10]. Core stability described by Fletcher (2016), '…is the kinetic

Table 3 Answer to a series of questions about the application of core stability

		Total
Do you think it is necessary to include specific exercises to train core stability in a healthy, uninjured athlete's exercise programme?	Yes	*206 (85)*
	No	30 (12)
	Do not know	5 (1)
Do you think it is possible to isolate and train the core stabilization system?	Yes	*120 (50)*
	No	82 (34)
	Do not know	39 (16)
Do you think it is effective to isolate and train the core stabilization system?	Yes	89 (37)
	No	*108 (45)*
	Do not know	44 (18)
Do you think that the core stability is automatically developed during normal, progressive exercise training?	Yes	*157 (65)*
	No	67 (28)
	Do not know	17 (7)

Data presented as number and percentage (n (%)) of all respondents. Italics represent the highest response for each question link transferring torques between the upper and lower extremities in sporting actions' [45]. Consequently, constituent anatomy of the core is described in the literature to reflect, i.e. region between and including pelvic and shoulder girdles and associated skeleton, muscles and nerves [1, 8]. Our survey results suggest this shift has permeated applied sports setting; half of the respondents agreed with this definition of the core while a quarter identified with the original description, i.e. structures between diaphragm and pelvic floor including muscles and nerves.

Surveys have been used effectively to assess nutrition knowledge [46] and understanding of scientific training principles [47] in the workplace. Response rate to our survey ($n = 241$) was good in comparison to similar surveys which gathered information from both athletes (Wade et al., $n = 57$) [48] and people working in sport (Taylor et al., $n = 28$) [49], (Durell et al., $n = 137$) [47] and (Torres-McGehee et al., $n = 579$) [46]. Furthermore, the representative quality of our cohort is reflected by the spread of respondents, with 33% in full-time professional positions, either working with professional athletes (21%) or full-time Institute of sport athletes (12%). A quarter (27%) were involved in sport in an academic setting, either school or university and a quarter (27%) were in non-professional roles, either volunteering (15%) or part-time (12%). The majority were qualified to degree level (87%) and half had postgraduate degrees (52%). Most had an industry-specific qualification and on average were well experienced (mean 8 years) in their discipline. The cohort is therefore representative of people working and participating in sport. Furthermore, they were reasonably well informed, indicating survey results that represent unbiased perceptions of the wider population.

Our survey investigated perceptions around core stability and core strength (Fig. 1). The majority believed that core strength is required for stability and far fewer agreed that these were separate attributes. In a comprehensive review Hibbs et al. [1] concluded that these two terms had yet to be clearly defined, in fact they failed to identify any characteristics that differentiated exercises for core strength and core stability. These researchers reviewed studies that investigated core stability in response to loaded resistance exercises and traditional core stability exercises. A later systematic review proposed a five-level core exercise classification system that progressed from traditional core exercises to noncore free weight exercises [6]. Interestingly the fourth classification level was free weight exercises defined as 'dynamic, externally loaded, intent to activate lower body and core muscles'. Both these reviews suggest that the concept of strength in the term core strength relates to the overarching nature of the exercise, rather than the impact on or adaptation in the core stabilization system.

While core strength and core stability may well be viewed by some in our survey as separate entities, this has yet to be demonstrated scientifically [1]. The selection of exercises used to develop core stability for healthy function can range from low load, minimal range of movement, abdominal bracing exercises to dynamic, loaded resistance exercises [6]. Research has not been able to identify and describe adaptations that occur in muscles responsible for stabilising the core as a consequence of different exercise modes [1, 12]. It is recognised though that effective core stability is the control of movement, including high force and high velocity movement, generated by interaction between axial and appendicular skeletons [5, 7, 8]. Most survey responses disagreed with the statement that core strength was required for athletic performance, but not everyday life. This demonstrated alignment with the principle that core stability underpins both healthy function and dynamic athletic performance. In effect core strength and core stability are synonyms and are used accordingly in the literature [1, 5, 23]. This is reflected in the survey question seeking to determine whether core stability and strength are separate attributes. Responses were mixed with just over half (57%) in agreement and the rest either undecided (16%) or in disagreement (27%).

In our survey questions that assessed support for exercise categories most effective in developing core stability for dynamic athletic performance, there was clearly more support for functional, loaded exercises (Fig. 2). Squats and Olympic lifts and farmers walk that engage the full kinetic chain. Conversely support was moderate to low for traditional, non-functional core stability exercises, namely suspended compound exercises, isometric plank, hanging leg raise, and instability abdominal exercises. Two exercise categories, namely abdominal bracing and sit-ups, were regarded as ineffective rather than effective, The survey results therefore reflect the many reviews that highlighted a lack of evidence to support traditional CST for healthy individuals and recommended loaded, dynamic exercises that engage the full kinetic chain [1, 6, 7, 12–14, 45].

Correct movement pattern was identified as most important exercise selection criteria for development of core stability for dynamic athletic performance (Fig. 3). Exercises characterised by forces that were equal to or greater than force in the sport or event, were supported by 60% of the cohort. Most were either undecided or disagreed on whether velocity of movement and sustained isometric contraction were important in core stability exercises for athletic performance. Kibler et al. (2006) accurately describes the exercise criteria for effective CST: 'integrated activation of multiple segments' providing 'force generation' that produces 'interactive movement' characterised by 'proximal stability and distal mobility' [5]. Core stability development is therefore integral to all dynamic exercise training and sports specific movement, while quality of training effect is determined by specificity of movement, forces and velocity.

There is growing evidence in the literature that external load in free barbell exercises performed in a standing position is related to muscle activation of trunk stabilisers [29, 30, 33, 34, 37, 50]. Impact of this stimulus on core stability in dynamic athletic performance is more difficult to demonstrate. In a recent systematic review, Prieske et al. (2016) reported a large effect for CST on trunk muscle strength measured by timed isometric plank, compared to no or only regular sports training [12]. When compared to alternative training, such as whole-body strength training, CST had a small sized effect on trunk muscle strength. CST had a small sized effect on muscle strength (e.g. Squat 1RM), a medium sized effect on muscle power (e.g. countermovement jump) and a small sized effect on athletic performance (e.g. 5000 m run time). They concluded that CST for healthy individuals, in the absence of any other fitness training, would increase trunk muscle strength. However, when combined with other training, such as whole-body strength training, CST is not effective. They also propose that increases in trunk muscle strength from CST, has limited effect on physical fitness and athlete performance in trained individuals. Findings from the survey indicate that this information has begun to inform applied practice (Fig. 4). Most agreed that increases in external load in standing barbell exercises would increase trunk muscle activation. Equally important in this form of resistance training was correct postural control.

The survey included a series of questions (yes/no/do not know) investigating perceptions on the application of CST for dynamic athletic performance (Table 3). Most (85%) of the cohort felt it necessary to include specific exercises to train core stability in healthy, uninjured athletes. With reference to traditional CST, two questions were asked; whether it was possible to isolate and train the core stabilization system, and whether this approach was effective. Half of the group believed that this was possible, 34% felt not and the rest were undecided (16%). The isolated training approach was regarded as not effective by 45%, and 37% were supportive. Prieske's review highlighted growing evidence that specific, traditional CST is ineffective in healthy individual and athletes [12]. They also that reported that regular sports training and commonly used supplementary training, such as whole-body strength training, presents superior stimuli, that adhere to the overload training principle [28], for development of core stability in this population. Most survey respondents (65%) concurred with this by agreeing that core stability is developed through normal, progressive exercise training. The perception in applied practice conflicts with scientific literature with regards effectiveness of traditional core stability exercises for athletic performance. The majority (85%) of survey respondents believed that specific exercises were required to train core stability and half supported the use of exercises that isolated trunk stabilisers.

A limitation of the survey was the method of recruiting participants through email and direct messaging on an online professional community platform (LinkedIn). Emails and notifications may have been filtered to spam or junk folders and not reached intended participants. Participants were directed to an online survey, which may have served as a deterrent. Despite this, the number and quality of participants was good in comparison to similar surveys. A further limitation may well have been the inconsistency of prevailing terminology around the topic of CST and broader area of exercise and fitness. Steps were taken to adhere to the most commonly used terms from the scientific literature in the survey.

Conclusion

The survey has provided evidence that a revised, more functional definition of core function and constituent anatomy described in the literature is starting to be used in the practical setting. Almost half (45%) of the respondents preferred trunk as the name for this anatomical region over core (35%). The absence of a valid objective method of measuring core function (22%) means that the most effective way is through observation (43%) of exercise and athletic movement. A quarter (26%) proposed subjective assessment of movement in upright loaded resistance exercises as the most effective method of measuring core function. This coincides with the strong shift in perceptions towards more functional approach to core stability training for dynamic athletic performance. Loaded exercises in an upright position, such as barbell squat and farmers walk, were viewed as effective training methods as proposed in the literature [7, 8, 14]. Core stability as an integrated, functional motor task [7], with training reflecting this according to movement patterns [14], forces [7], torque and velocity [8], appear to be guiding practice in the workplace according to the survey. These findings along with strong support for developing core stability through normal progressive exercise training, means we found in favour of our hypothesis. Some support remained for traditional CST through specific exercises (85%) and the isolation approach (50%). Our findings lead to the following recommendations: Research to continue into efficacy of activating trunk stabilisers through selected sport specific and supplementary training modalities, including compound, loaded strength exercises. Continue to investigate the transfer of training induced trunk muscle activation to functional performance, specifically functional stability.

Additional file

Additional file 1: Core Stability Survey. (PDF 103 kb)

Abbreviations

1RM: 1 Repetition maximum; CST: Core stability training; MSc: Master of Science; S&CC: Strength and conditioning coaches; A&P: Athletes and players; SM&P: Sport medicine practitioners and physiotherapists; SP&B: Sport physiologists and Biomechanists; SC: Sport coaches

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Availability of data and materials

Appendix will include the survey questionnaire, and survey master results document will be made available and labelled Additional file 1.

Authors' contributions

DC conceived and designed survey with assistance from JT (Acknowledgements) and ML. DC managed the data collection, survey circulation, data collation, analysis and interpretation. Final analysis and interpretation for publication was done by DC with assistance from ML and AH. All authors read and approved the final manuscript.

Ethics approval and consent to participate

Approval for the study was granted by the local research ethics committee in accordance with the Helsinki Declaration (2013) [40].

Consent for publication

Survey instructions informed participants of the details of the research study, completion and submission implied consent.

Competing interests

David Clark, Mike Lambert and Angus Hunter declare that they have no competing interests.

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Study 3: Determine reliability of trunk muscle electromyography in the squat exercise

The survey findings confirmed a connection between perceptions and practices in the applied setting and key aspects of trunk stability in scientific literature. The survey found good support for more functional, integrated anatomical definition of the trunk and consequently, exercise selection for developing trunk stability. There was clear support for compound loaded exercises and a recognition of the role of force, velocity and correct technical movement in exercise training for trunk stability. However, strong support remained for traditional CST training including the isolation approach.

There are two possible explanations for this ambivalence; firstly, this may reflect time taken for scientific research to fully effect change in applied setting and secondly, illustrates absence of valid and verifiable research based exercise guidelines to comprehensively replace traditional CST. The recommendations from the survey were to continue to investigate and demonstrate efficacy of loaded, upright compound exercises in activating trunk stabilizers and to demonstrate transfer of this training stimulus to athletic function and performance.

The review study recognised variability in research tools used in published research on trunk muscle activation in back squat. EMG data analysis and presentation included integrated EMG, where raw EMG signal was reported, and normalized EMG. Two methods of normalization were reported; EMG in test effort is normalized to EMG during a maximal voluntary isometric contraction (MVIC) or normalized to a pre-identified signal during a dynamic, submaximal test effort.

Previous work from our laboratory demonstrated that submaximal, dynamic EMG normalization methods were more reliable and sensitive than normalizing to maximal isometric EMG methods (Balshaw and Hunter, 2012). This was found for vastus lateralis and biceps femoris activation in the back squat at moderate (65 & 75% 1RM) and heavy (85 & 95% 1RM) loads. Prior to that, dynamic, submaximal normalization of lower limb muscle EMG was found to be more effective than MVIC method in cycling (Albertus-Kajee *et al.*, 2010) and running (Albertus-Kajee *et al.*, 2011). Dankaerts et al (2004) reported greater within and between day reliability for dynamic submaximal EMG normalization than MVIC for trunk muscles in healthy and back pain patients (Dankaerts *et al.*, 2004).

In an extensive review of EMG normalization procedures, Burden (2010) determined that accuracy MVC forces and torques were dependant on training status and could therefore be 20-40% less than absolute maximal values (Burden, 2010). Hence, using MVC generated EMG as the denominator in the normalization process may not represent a relative consistent anchor for submaximal EMG analysis for a mixed group of participants. Furthermore, they concluded that normalizing to mean EMG captured in submaximal dynamic execution of the task under investigation is suitable for within trial analysis, but not between trails where electrodes are re-applied. Kinematic stability and consistency of the back squat technique within and between participants can be controlled through standardising range of movement or squat depth and using the same relative external load. As a result, and based on evidence supporting dynamic, submaximal EMG normalization we selected to normalize EMG for the 3 neuromuscular studies to mean concentric EMG at 65% 1RM back squat. This meant the within each muscle group mean eccentric and concentric EMG at 75, 85 and 95% 1RM was normalized to the mean concentric EMG at 65% 1RM

There was also variability in methods of determining back squat test loads, which may have undermined reliability and value of findings. Particularly after having established that trunk muscle activation is sensitive to load increments. Comparing activation in response to absolute loads does not account for individual differences in strength levels. This is overcome by using relative test loads calculated from individual maximal strength test scores corrected for body mass. Furthermore, there were no reports on the reliability and repeatability of measuring trunk muscle activation by sEMG in the back squat.

Based on the review we established a standard methodology for all subsequent studies. Selected muscle sites were based on previous published research (Anderson and Behm, 2005; Hamlyn, Behm and Young, 2007) and guidelines from the SENIAM (Surface Electromyography for non-Invasive Assessment of Muscle) (Hermens *et al.*, 1999) (Appendix 3). The detailed review of the literature leading to the selection of these muscle sites is presented in the introduction of the thesis. Synchronised linear encoder signal was used to determine mean root mean square (RMS) processed EMG or muscle activation data for eccentric and concentric phases of the squat (Appendix 4).

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Reliability of Trunk Muscle Electromyography in the Loaded Back Squat Exercise

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Key words
- back squat
- neuromuscular
- trunk muscles
- electromyography
- electrogoniometry

Abstract

Trunk muscle activation (TMA) has been reported during back squat exercise, however reliability and sensitivity to different loads alongside kinematic measures has not. Hence the aim was to determine the *inter*day reliability and load sensitivity of TMA and kinematics during back squats. 10 males performed 3 test sessions: 1) back squat 1RM, 2) and 3) 3 reps at 65, 75, 85 and 95% of system mass max (SMmax). Kinematics were measured from an electrogoniometer and linear transducer, and surface electromyography (sEMG) recorded 4 muscles of the trunk: rectus abdominis (RA), external oblique (EO), upper lumbar erector spinae (ULES) and lumbar sacral erector spinae (LSES), and a reference leg muscle, the vastus lateralus (VL). sEMG amplitude was root mean squared (RMS). No differences ($p > 0.05$) found between tests for any kinematic and RMS data. CV demonstrated moderate interday reliability (~16.1%) for EO, LSES and ULES but not RA (29.4%) during the velocity-controlled eccentric phase; whereas it was moderately acceptable for just LSES and ULES (~17.8%) but not RA and EO (27.9%) during the uncontrolled concentric phase. This study demonstrated acceptable *inter*day reliability for kinematic data while sEMG for most trunk muscle sites was moderately acceptable during controlled contraction. sEMG responded significantly to load.

Introduction

Although development of core strength and core stability is important for everyday health and sporting performance, the most effective methods for developing these characteristics are unclear, particularly as it applies to dynamic athletic performance [29]. However, the back squat is a training exercise often used to develop core strength and stability, and there is growing scientific evidence in support of its efficacy [24, 36]. Neuromuscular activation in the back squat has been well researched, although most of those investigations have focussed on activation of the muscles of the lower limb including the prime movers [15]. However, there are also an increasing number of studies [13, 36, 42] reporting trunk muscle activation in the back squat exercise. Most of this research has attempted to assess the efficacy of the loaded back squat as a method of activating and therefore developing the trunk stabilizers. Fundamentally, these studies have demonstrated that trunk muscle activation of the posterior chain (erector spinae) increase in response to increases in external load.

Additional methods have been used to further establish the effectiveness of the back squat exercise to develop core strength by comparing it to trunk isometric [3, 17, 24, 36] or trunk dynamic exercises [3] performed in a prone or supine position. The studies measured a wide selection of muscles, typically the rectus abdominis, external oblique, erector spinae (ES), longissumus and multifidus. However, many of these studies produced conflicting findings about which exercise mode produced the greatest neuromuscular activation. Most of these investigators acknowledged that comparing trunk muscle activation in the back squat to isolation trunk muscle exercises is inappropriate. Also there is a risk of misinterpreting greater activation of a selected muscle group to justify exercise mode selection in programmes designed for developing dynamic trunk stability. It has been suggested that dynamic trunk stability for injury prevention and sports performance is related to the onset and duration of electrical activity of the trunk stabilizers rather than magnitude of activation [34].

The methodology used to capture and analyse surface electromyography (sEMG) varies across

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the studies where trunk muscle activation is reported in the back squat. Mean un-normalised sEMG amplitude (iEMG) has been reported during back squat exercise [24,32,36] which can induce considerable inter- and intra-participant variability as a number of extrinsic and intrinsic factors are not accounted for [14]. Therefore, to reduce this effect, a number of studies normalize EMG data against a reference value captured during a maximal isometric voluntary contraction (MIVC) [3,13,17,42]. However, recent studies have presented an alternative dynamic method where the test EMG is normalized against a reference value captured in a standardized submaximal effort in the same movement as the test [1,8,14,30].

Accurate kinematic measurement of the squat movement facilitates the analysis of the neuromuscular data associated with the technical and mechanical execution of the exercise. Researchers have managed this by controlling the duration of descent and ascent [3], by using a force platform to calculate the position of the centre of mass [17], and by incorporating 2 position transducers in conjunction with a force platform to measure horizontal and vertical displacement [32,33]. A flexible electrogoniometer has also been used along with a position transducer in a number of squat studies [8,11,12,37] to measure angular displacement and detemine the phases for sEMG analysis.

sEMG has been effectively used in studies measuring trunk muscle activation, however the methodology used is diverse [30]. Despite this, there is evidence that trunk muscle sEMG is sensitive to load changes [15,24,36]. Importantly, to our knowledge interday reliability of sEMG measurement of trunk muscle activation has yet to be established. It is critical to establish this so researchers and practitioners can account for day-to-day “measurement noise” to enable accurate inference of EMG change in trunk muscles following an intervention.

The main aim of this study was to determine; 1) the interday reliability and sensitivity sEMG normalized dynamically to measure trunk muscle activation in response to different relative loads in the eccentric and concentric phases of the loaded barbell back squat, 2) the reliability and sensitivity of kinematic measures calculated from an electrogoniometer and linear position transducer.

Methods
▼
Participants

10 males volunteered for this study (age: 26.6 ± 8.4 years, body mass: 86.1 ± 7.8 kg, squat training age: 5.7 ± 5.0 years, squat 1RM: 142.0 ± 29.2 kg, relative squat 1RM: 1.7 ± 0.3). All were actively participating in regular strength exercise training and had at least 1 years’ experience in performing the barbell back squat exercise (back squat 1RM: 142.0 ± 29.2 kg, relative back squat 1RM: 165 % ± 30 % body mass). Approval for the study was granted by the local research ethics committee in accordance with the Helsinki Declaration (2013) and all participants signed an Informed consent form prior to testing. This study complied with the ethical standards for sport and exercise science research according to Harris and Atkinson [26]. Participants abstained from strenuous exercise and followed their usual dietary habits for 24 h prior to test sessions, which were conducted at the same time of day to account for circadian variation [5].

Experimental design

All participants completed 3 test sessions. In the first session back squat 1RM was determined and they were briefed on the format for the subsequent test sessions. The second test session was conducted within 7 days and the third test between 5 and 7 days thereafter. All back squat repetitions across each of the 3 separate test days were performed according to the previously described protocol [6], which required participants to descend to where the tops of their thighs were horizontal. All back squats were performed using barbells and discs approved by the International Weightlifting Federation (Eleiko, Sweden) and conducted in a safety power cage (FT700 Power Cage, Fitness Technology, Skye, Australia).

Initial 1RM test and familiarization session

Participants performed a standardised warm-up prior to completing the back squat 1RM test according to the protocol recommended by the National Strength and Conditioning Association of the USA [6]. The warm-up comprised 5 min stationary cycling followed by 10 min of dynamic callisthenic exercises. This was followed by a barbell warm-up comprising 5 sets of 10, 8, 6, 4 and 2 repetitions at progressive loads determined for each participant based on previous 1RM test results and current training loads. 1RM score was recorded as the highest load lifted successfully through the required range of movement within 4 attempts. Participants were instructed to control the cadence of the descent and the ascent themselves and to rest for 3 min between each warm-up and test set [11,12,19].

Test load calculation

The test loads for the muscle activation protocol in test sessions 2 and 3 were calculated from the system mass max (SMmax) [11,12,19,20]. SMmax is accurate in determining relative test and training loads and sensitive to changes in body mass in test-retest protocols. The determination of SMmax assumes that 88.6 % of body mass should be added to the external load as this is lifted when performing the squat [22]. The remaining 11.4 % represents the shanks and feet, which do not move vertically in the exercise action. The loads used for the muscle activation test protocol included 2 warm-up sets of 10 repetitions at 45 and 55 % SMmax and 4 test sets of 3 repetitions at 65, 75, 85 and 95 % of SMmax respectively (▶ **Table 1**). The external loads for the warm-up and test sets were determined according to the following equation:

$$\text{SMmax} = 1\text{RM} + (0.886 \times \text{body mass})\ (\text{kg})$$

$$\text{External test load} = (\text{SMmax} \times \text{percentage of SMmax}) - (0.886 \times \text{body mass})\ (\text{kg})$$

Muscle activation test protocol

Test sessions 2 and 3 were separated by 5 and 7 days to assess reliability, sensitivity and inter-participant variability of the kinematic test measures and sEMG data. In test sessions 2 and 3,

Table 1 Mean (kg) ± SD warm-up and test loads as a percentage of system mass max (SMmax).

	Warm-up loads			
SMmax (%)	**45 %**	**55 %**		
Mean (kg) ± SD	21.9 ± 12.2	43.8 ± 15.1		
	Muscle activation test loads			
SMmax (%)	**65 %**	**75 %**	**85 %**	**95 %**
Mean (kg) ± SD	65.6 ± 18.2	87.4 ± 21.3	109.3 ± 24.4	131.1 ± 27.6

after completing the standardised warm-up participants were prepared for the capture of sEMG and kinematic data. sEMG and kinematic data capture was confirmed during the 2 warm-ups sets before proceeding to the sEMG test protocol. Participants rested for 2 min between warm-up sets and 3 min between test sets. Squat descent was controlled to a minimum of 2 s by metronome and participants were instructed to perform the ascent in an explosive and controlled manner.

Kinematic data

A flexible, 2 dimensional electrogoniometer (TD130B, Biopac Systems, Inc., California, USA) was attached to the right leg for both EMG test sessions [8,11]. Data were used to establish the start of the descent, the transition between descent and ascent and the end of the squat as well as to determine squat depth. The duration and displacement of the eccentric and concentric phases of the squat were measured by linear transducer (Celesco, PT5A, California, USA) which was placed directly below the path of the barbell and attached to the barbell [8,11].

A bespoke Matlab (Matlab R2010A, The Mathworks Inc., USA) programme was designed to identify the initiation and completion of the descent of the barbell. This method used the goniometer Y double differentiated signal to identify the change in threshold from lowest knee angle, i.e., closest to 0°, to the highest knee angle. Following this the programme could then detect the transition from the highest knee angle for the start of the ascent through to the lowest knee angle. In addition to enabling us to accurately calculate the kinematic measures of the movement, it also provided us with an objective method of RMS selection for respective descent and ascent phases.

All kinematic and RMS data were analysed within the 2 phases of the squat; eccentric and concentric, and presented as the mean ± SD for the 3 reps for each test load; 65, 75, 85 and 95% SMmax. The displacement and duration (time) of the eccentric and concentric phases of the squat were captured by the linear transducer and used to calculate velocity and power according to the following formula:

1) Velocity = (displacement/time)*100
Acceleration = Velocity/time
Force = External load (%SMmax) × {acceleration + gravity (9.812)}
2) Power = force × velocity

Electromyography

EMG was recorded (Biopac MP100, Biopac Systems Inc., Santa Barbara, CA) from 5 muscle sites on the right-hand side of the body: 4 muscles of the trunk, namely the rectus abdominus (RA), external oblique (EO), lumbar sacral erector spinae (LSES), upper lumbar erector spinae (ULES) [2,24], and a reference muscle from the lower limb, the vastus lateralis (VL) [2,3]. Skin was prepared by removing hair, abrading the skin with emery paper and cleaning the site with an alcohol swab. Two Ag-AgCl EL258S bipolar 8 mm diameter electrodes (Biopac Systems Inc., USA) were fitted in a custom-made soft rubber mould with an interelectrode distance of 20 mm according to SENIAM (Surface Electromyography for Non-Invasive Assessment of Muscles) (1999) [28] recommendations. Electrodes were fixed longitudinally along the muscle fibre orientation according to SENIAM (ULES and VL), Anderson & Behm (2005) [2] (LSES, ULES and VL) and Hamlyn et al. (2007) [24] (RA, EO, LSES and ULES). Each electrode was filled with conductive gel and fixed in position with transparent adhesive dressing. EMG was sampled at a rate of 2 000 Hz and anti-aliased with a 500 Hz low-pass filter. EMG data were root-mean-square processed (RMS) and the mean RMS for each phase, eccentric and concentric, was calculated from the 3 reps for each test load. Mean RMS for each phase of each of the 3 test loads of 75, 85 and 95% SMmax were normalized to the mean RMS of the concentric phase of the 65% SMmax test. Previously published work of ours showed that submaximal dynamic normalization was far more reliable and sensitive than MVC methods in the back squat exercise for the VL [8]. Dankaerts et al. [21] demonstrated that submaximal not maximal isometric contraction proved to be more reliable in EMG measurement of the trunk muscles in healthy controls and patients with lower back pain.

Statistical analysis

Kinematic data is reported as mean ± SD for the 4 test loads and both the eccentric and concentric phase of the squat. Normalized RMS data is reported as a percentage for test loads at 75, 85 and 95% SMmax for both the eccentric and concentric each phase of the squat.

Absolute reliability was assessed using the intra-participant coefficient of variation (CV%) and limits of agreement (LOA) [10]. Intra-participant CV% was calculated for mean kinematic variables, i.e., duration, displacement, velocity and power for the 4 test loads and each phase of the squat. Intra-participant CV% and LOA was calculated for mean RMS for each muscle site, combined for all test loads and for each phase of the squat. Acceptable variability has been defined as CV values less than 10% [35,43], however CV values specifically for the quadriceps muscles have been described as acceptable when between 9% and 12% [1,8,11,41]. As a cautionary measure, the definition for an "acceptable" CV value is regarded as less than 12% and "unacceptable" as greater than 20% as previously reported [1]. Therefore CV values between 12–20% will be regarded as 'moderately acceptable'.

Intra-class correlation coefficient (ICC) was used to determine relative reliability or variations in participant order across repeated tests. ICC values and 95% confidence intervals were calculated using the statistical spreadsheet downloaded from www.sportsci.org. Inter-participant CV% was used to assess inter-participant variability in the RMS data for each muscle site for each test load and the mean of all test loads for each muscle site. Intraclass correlation coefficient was determined from the ANOVA F value:

$$ICC = (F-1)/(F+k-1),$$

where k = (number of observations − number of tests)/(number of participants − 1)

The ICC scores ranging between R = 0.80 and 1.00 were defined as representing "good" reproducibility, scores between R = 0.60 and 0.79 "fair" reproducibility and less than R = 0.60 "poor" reproducibility [39].

Equal variances were assumed as all data were non-significant using Levene's Test for equality of variances. Repeated measures analysis of variance (ANOVA) (Minitab Ltd., Coventry, UK) was performed on kinematic and RMS data to determine differences of test measures from test to retest and across the 4 test loads. Tukey *post hoc* analysis was used to assess differences where test load interaction was found. Significance was accepted at $p < 0.05$.

Results

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Kinematic measures

Response to load increments

Mean displacement in the eccentric phase declined by an average of 56 mm (range 22–86 mm) with each 10 % increment in SMmax from 65 to 95 % SMmax ($F_{(3,27)}=3.06$, $p<0.05$) (▶ **Fig. 1b**). There was also a significant increase in time as a result of increases in load for the eccentric phase ($F_{(3,27)}=7.49$, $p<0.05$) (▶ **Fig. 1a**). Mean eccentric velocity declined significantly with increased load with *post hoc* testing revealing the significant ($p<0.05$) increases occurred at 85 % and 95 % SMmax (▶ **Fig. 1c**). Eccentric power increased significantly ($p<0.05$) alongside load (▶ **Fig. 1d**). No significant differences or interactions were observed between test days for any of the eccentric kinematic variables (▶ **Table 2**).

Concentric displacement of the barbell was not different ($p>0.05$) between test sessions or across loads (▶ **Table 2**). Mean concentric squat duration (▶ **Fig. 2a**) increased significantly ($F_{(3,27)}=34.21$, $p<0.001$) with each test load increment (▶ **Fig. 2b**), whereas velocity decreased significantly ($F_{(3,27)}=45.68$, $p<0.001$) (▶ **Fig. 2c**). Mean power for the concentric phase significantly changed across loads $F_{(3,27)}=12.43$, $p<0.001$); *post hoc* testing revealed that power was significantly ($p<0.001$) greater for 75 and 85 % SMmax compared to the power at 65 % and 95 % SMmax (▶ **Fig. 2d**). There was no significant differences in power at 75 and 85 % SMmax nor between power at 65 and 95 % SMmax. No significant differences or interactions were observed between test days for any of the concentric kinematic variables (▶ **Table 2**).

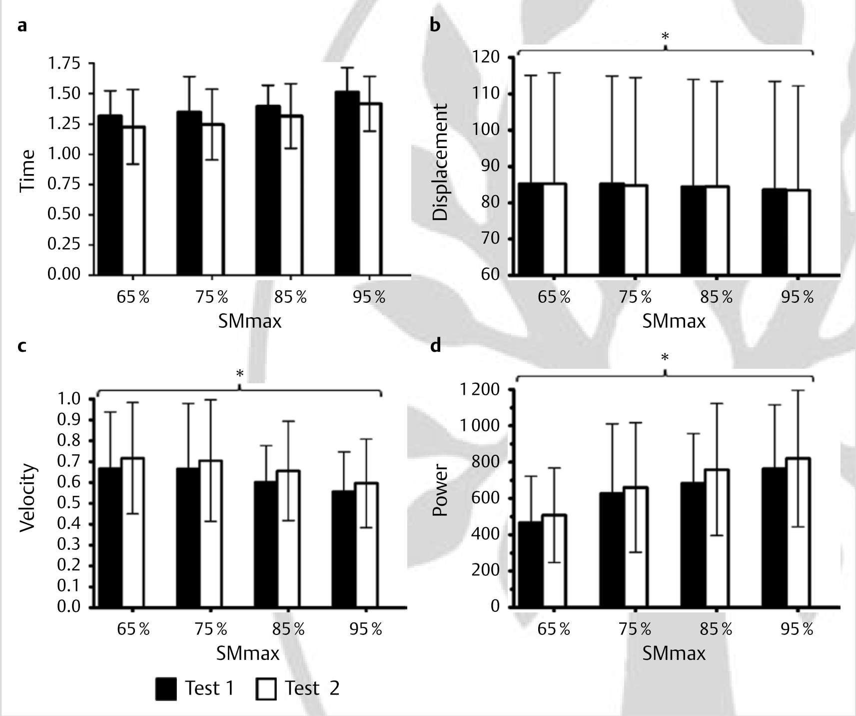

Fig. 1 Eccentric kinematic data for test 1 and 2 for all test loads: Time **a**, Displacement **b**, Velocity **c** and Power **d**. * Denotes significant load effect (*p*<0.05).

Table 2 Mean differences and Intra-participant CV % (mean ± SD) between test 1 and test 2 for the kinematic variables for the eccentric and concentric phases at the 4 test loads.

		Eccentric		**Concentric**	
	Test load	**Difference Mean ± SD**	**Intra participant CV % Mean ± SD**	**Difference Mean ± SD**	**Intra-participant CV % Mean ± SD**
Time (s)	65 %	0.9 ± 0.3	11.2 ± 11.5	0.1 ± 0.2	9.1 ± 6.2
	75 %	0.1 ± 0.2	7.6 ± 5.5	0.1 ± 0.2	7.9 ± 6.2
	85 %	0.1 ± 0.2	8.7 ± 7.6	0.1 ± 0.1	5.8 ± 3.3
	95 %	0.1 ± 0.2	7.4 ± 4.1	0.12 ± 0.2	8.3 ± 6.0
Displacement (cm)	65 %	0.0 ± 1.9	1.2 ± 0.7	− 0.0 ± 2.3	1.4 ± 0.9
	75 %	0.5 ± 1.5	1.4 ± 0.8	− 0.2 ± 2.4	1.7 ± 1.0
	85 %	− 0.0 ± 1.9	1.4 ± 1.2	0.2 ± 2.7	1.8 ± 1.3
	95 %	0.1 ± 2.5	1.7 ± 1.7	1.0 ± 2.6	1.8 ± 1.6
Velocity (m/s)	65 %	− 0.1 ± 0.1	10.6 ± 11.4	− 0.18 ± 0.1	8.8 ± 7.0
	75 %	− 0.0 ± 0.1	7.6 ± 6.0	− 0.0 ± 0.2	8.3 ± 6.8
	85 %	− 0.1 ± 0.1	9.1 ± 8.1	− 0.0 ± 0.1	6.2 ± 4.3
	95 %	− 0.0 ± 0.1	7.5 ± 4.6	− 0.1 ± 0.1	7.5 ± 6.3
Power (W)	65 %	− 42.0 ± 130.0	11.7 ± 12.6	− 74.4 ± 106.0	10.3 ± 7.7
	75 %	− 33.6 ± 86.9	8.3 ± 6.5	0.11 ± 194.0	9.54 ± 8.0
	85 %	− 74.2 ± 181.4	10.1 ± 9.0	− 27.0 ± 92.5	6.8 ± 4.6
	95 %	− 56.2 ± 100.1	8.1 ± 4.9	− 83.0 ± 98.2	8.0 ± 6.7

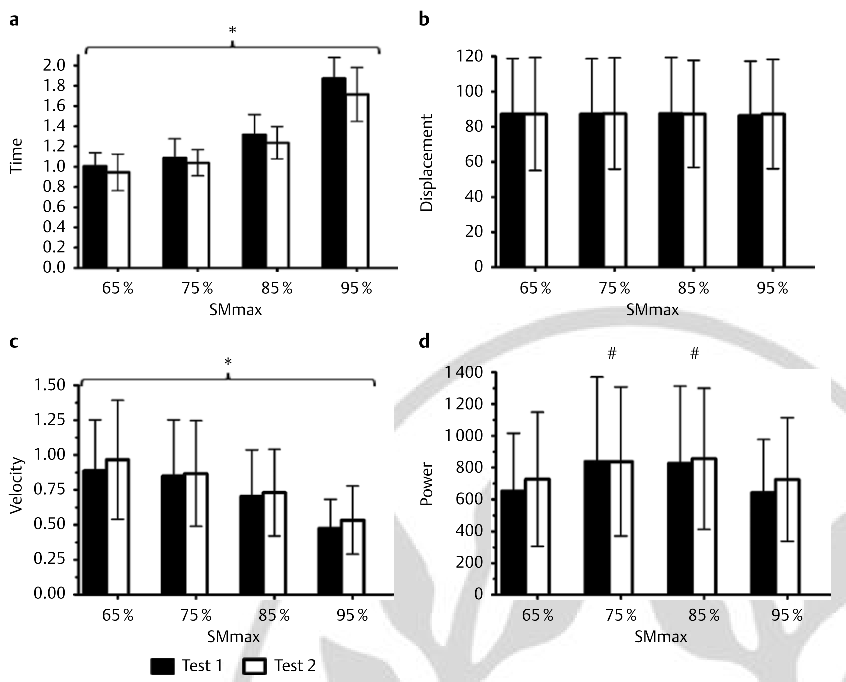

Fig. 2 Concentric kinematic data for test 1 and 2 for all test loads: Time **a**, Displacement **b**, Velocity **c** and Power **d**. * Denotes significant load effect ($p<0.05$). # Denotes significant difference from data at 65 and 95 % SMmax ($p<0.01$).

Reliability/repeatability

Kinematic mean differences and intra-participant CV% for test 1 to test 2 for each test load is presented on ▶ **Table 2** for both eccentric and concentric phases. Test-to-retest intra-participant CV% for all data ranged from 1.2 ± 0.7 % to 11.7 ± 12.6 %. This is considered as acceptable reliability as it is within the upper limit of < 12 % as described by [1,41].

Electromyography

Response to load increments

The mean RMS data showed no change from test to retest in all muscle sites in the eccentric and concentric phase (▶ **Fig. 3**). Mean RMS increased significantly ($p<0.05$) for all muscle sites by load in the eccentric phase (▶ **Fig. 3**). Mean RMS increased by load in the concentric phase for ULES ($F_{(2,18)}=7.80$ $p<0.05$), LSES ($F_{(2,18)}=31.86$ $p<0.001$) and EO ($F_{(2,18)}=3.57$ $p<0.05$) ($p<0.05$) and with a slight tendency ($F_{(2,18)}=2.14$ $p=0.14$) for RA (▶ **Fig. 4**).

Reliability/repeatability

Absolute reliability according to the differences from test to retest in mean RMS for each muscle site in each phase of the squat ranged from − 26.6 ± 25.0 % to 26.1 ± 26.3 % (▶ **Table 3**). Test-to-retest intra-participant CV% for all RMS data ranged from 12.6 ± 7.2 % to 29.4 ± 1.12 % (▶ **Table 3**). Based on intra-participant CV% the VL demonstrated the greatest absolute reliability in the eccentric phase compared to all the trunk muscles. The EO showed greater absolute reliability in the eccentric phase compared to the concentric phase based on the intra-participant CV%. The absolute reliability of both the LSES and ULES was better in both the concentric and eccentric phases than all the other muscles (RA, EO and VL) as measured by intra-participant CV%.

Relative reliability or variations in participant order across repeated tests was assessed by interclass correlation coefficient (ICC) and is presented in ▶ **Table 4**. Mean ICC for the 3 test loads demonstrated fair relative reliability for RA (R = 0.60) and EO (R = 0.71) in the eccentric phase and the LSES (R = 0.60) in the concentric phase.

Discussion

This is the first study to investigate the interday reliability of trunk muscle activation using surface EMG in the back squat exercise at loads ranging from moderate to heavy. Kinematic descriptors calculated from an electrogoniometer and linear position transducer confirmed previous data for the back squat at similar loads [8,11,12]. The RMS data, which was normalized dynamically, was shown to be moderately reliable and sensitive to load increments. Hopkins at al [31] in his review of reliability of performance tests suggested that a lower number of participants was acceptable, especially if this group were homogenous in the key area of competence. In our study the 10 participants were competent in the free barbell back squat, with a mean 1RM of 165 % body mass. Previously published work of ours reporting reliability during back squat exercise supports this notion by recruiting similar numbers [8,11]. Furthermore and importantly, this study demonstrated that trunk muscle activation increased significantly in response to 10 % SMmax increases in load for all muscle sites in the eccentric phase and in the ULES, LSES and EO in the concentric phase.

The absolute reliability of the kinematic measures in this study are within an acceptable range of similar studies [11,18,19]. The mean CV% for concentric power in this study ranged from 6.84–10.28 % for the 4 test loads, 65, 75, 85 and 95 % SMmax, while Brandon et al. (2011) [11] reported a mean CV% of 7.8 % for concentric power at 75 and 100 % of 3RM. Therefore, this provides an acceptable independent measure of reliability from which to interpret trunk muscle activation via sEMG.

In this study participants were instructed to perform the descent in a controlled and safe manner. As expected, the duration of the eccentric phase increased significantly with each load increment in accordance with safe squat technique. Bentley et al. (2010) [9] reported that a fast descent in the squat compared to a slow descent for the same load produced a larger ground reaction force. It has also been shown that a fast descent increases knee shear forces and spine compressive force [27]. As such,

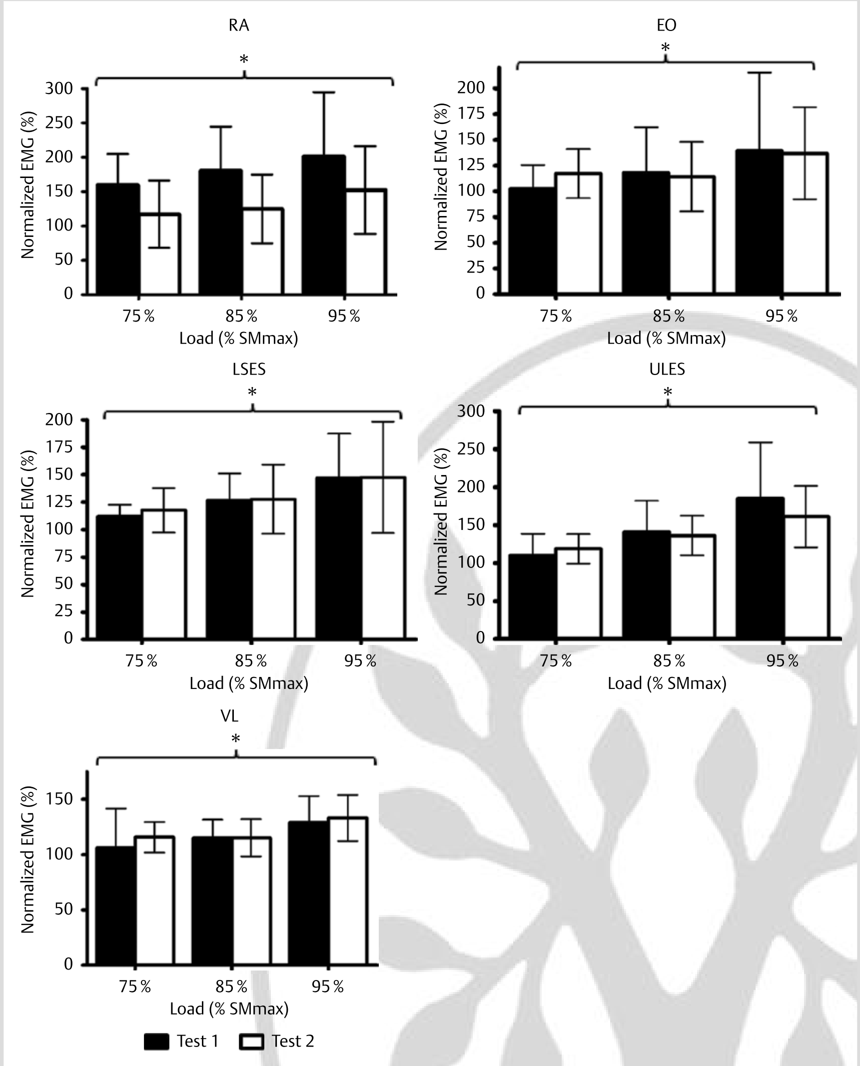

Fig. 3 Eccentric mean RMS for 3 test loads normalized to 65 % SMmax for the 5 muscle sites: RA, EO, LSES, ULES and VL. * Denotes significant load effect $p<0.001$.

expert squatters produce less vertical velocity in the descent than novice squatters [23]. The control of the load into the transition between the descent and ascent is an established coaching principle for heavy squats and is determined by individuals for each load. Brandon et al. (2011) [11] accounted for this by instructing expert squatters to apply a ‘self-selected normal tempo’ while in his review, Schoenfeld (2010) [38] recommended that the ‘squat should always be executed in a controlled fashion’. Similarly, in this study mean duration of the eccentric phase increased by 2, 4 and 8 % for the 75, 85 and 95 % SMmax squats.

Despite attempting to control eccentric displacement for all trials, this declined significantly for each load increment. This is likely the result of proprioceptive protection by participants as they approach the transition from the eccentric to concentric phase. The increased load received through the contracting muscles would have discharged the associated Golgi tendon organs [16], which would have inhibited relaxation of the hip extensors thereby reducing displacement. This challenge is magnified as the load increases and in an attempt to cope with this it appears that participants shorten the descent. Logically, the subsequent concentric displacement should mirror eccentric displacement. However, in this study the concentric phase was not effected by load. This may be explained by the instruction to participants to perform the ascent as explosively as possible resulting in the concentric phase ending slightly higher than the start point. Brandon et al. (2011) [11] in a similar study observed that the absence of control in the concentric phase represents physiological and motor skill variability in execution, which may explain the difference between mean eccentric and concentric displacement. Furthermore there is evidence that the spine temporarily shortens by up to 3.9 mm in response to axial loading of 1 × body mass due to rotation, bending and compression of the spinal cord [44]. This shortening may account for the reduced eccentric displacement resulting in the concentric displacement remaining unchanged. Following a controlled descent (eccentric phase) the participants were instructed to perform the subsequent concentric phase explosively, whereupon velocity decreased alongside greater loads. This is to be expected [3,20,45] as was the concentric classic power curve we demonstrated (▶ **Fig. 2d**) showing the established relationship between external load and power in the back squat [19,20,25,45].

Absolute reliability during the eccentric phase of *inter*day RMS, by calculating intra-participant CV % was moderately acceptable for all the muscles sites apart from RA. The RA was over this threshold, indicating that it is not a reliable measurement. This is possibly due to trunk flexion through the eccentric phase of the back squat movement causing folding of the skin in this region and excessive motion artefact of the sEMG signal. The explosive, uncontrolled nature of the concentric phase introduced an additional variable, which may explain why none of

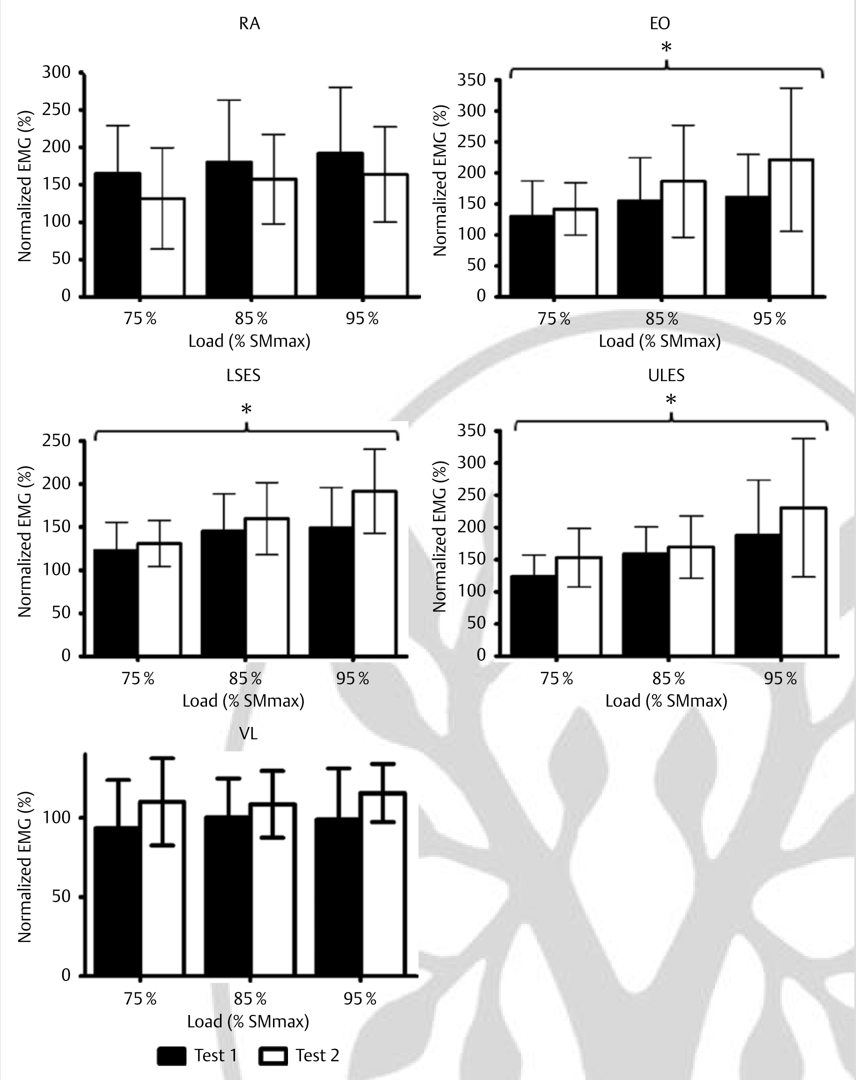

Fig. 4 Concentric mean RMS for 3 test loads normalized to 65 % SMmax for the 5 muscle sites: RA, EO, LSES, ULES and VL. * Denotes significant load effect $p<0.001$.

Table 3 Mean differences in RMS, limits of agreement and intra-participant CV % (mean ± SD) between test 1 and test 2 for the 5 muscle sites in the eccentric and concentric phases. Data for each muscle site is ranked according to intra-participant CV % data, where 1 is most reliable and 5 least reliable.

Muscle action	Muscle site	Difference between test days Mean ± SD	95 % Upper LOA Mean ± SD	95 % Lower LOA Mean ± SD	Intra-subject CV % Mean ± SD	RANK
Concentric	RA	−21.1 ± 14.7	90.3 ± 65.0	−132.5 ± 89.7	27.8 ± 3.4	4
	EO	26.1 ± 26.3	154.5 ± 115.3	−102.3 ± 69.6	28.0 ± 5.9	5
	LSES	16.2 ± 18.4	66.7 ± 48.4	−34.3 ± 25.8	16.3 ± 4.7	1
	ULES	20.7 ± 19.3	108.6 ± 101.4	−67.2 ± 63.0	19.3 ± 7.6	2
	VL	10.4 ± 8.0	54.8 ± 39.0	−34.0 ± 24.5	19.5 ± 4.3	3
Eccentric	RA	−26.6 ± 25.0	42.8 ± 40.9	−116.0 ± 80.7	29.4 ± 1.2	5
	EO	2.0 ± 8.5	64.9 ± 51.5	−60.8 ± 54.0	15.9 ± 3.6	3
	LSES	2.1 ± 2.7	46.5 ± 37.3	−42.3 ± 37.2	12.9 ± 3.9	2
	ULES	−5.1 ± 13.6	82.8 ± 61.7	−93.0 ± 82.3	19.7 ± 4.1	4
	VL	3.5 ± 4.6	41.4 ± 34.07	−34.3 ± 25.9	12.6 ± 7.2	1

RA-Rectus abdominus, EO-External oblique, LSES-Lumbar sacral erector spinae, ULES-Upper lumbar erector spinae, VL-Vastus lateralus

the muscle activation measured was reliable in this phase based on CV %. However, the LSES and ULES were found to be moderately reliable. Despite this, our data was shown to be more reliable than Hibbs et al. (2011) [30] who measured *intra*day ARV sEMG from similar muscle sites, but during static core strength exercises such as side bridge plank, medicine ball sit-hold-twist etc., as opposed to loaded back squat. This is surprising as this study [30], used sEMG electrodes that remained attached between sessions and used one normalization reference. As such, we would expect far lower CV % values than the ones shown in our study as we tested on separate days necessitating independent sEMG electrode placement each time. This methodological consideration along with separate normalization tasks should in theory create more, not less variance. Interestingly, our levels of absolute reliability (i. e., CV) did not match the relative reliability scores (i. e., ICC). For example, LSES presented

Table 4 Interclass correlation and inter-participant CV % (mean ± SD) for the 5 muscle sites and test load for the eccentric and concentric phases. ICC results regarded as fair relative reliability are presented as bold.

		Eccentric		Concentric	
Muscle site	**Test load**	**Interclass Correlation Mean (LCI–UCI) ***	**Inter-subject CV % Mean ± SD**	**Interclass Correlation Mean (LCI–UCI) ***	**Inter-subject CV % Mean ± SD**
Rectus abdominis	75 %	0.34 (−0.55–0.73)	35.2 ± 2.6	0.57 (−0.16–0.83)	45.2 ± 2.4
	85 %	0.81 (0.39–0.93)	37.7 ± 9.9	0.27 (−0.65–0.70)	42.1 ± 16.5
	95 %	0.64 (−0.00–0.87)	44.1 ± 20.1	0.53 (−0.22–0.82)	42.5 ± 17.0
	Mean	**0.60** (−**0.01–0.88)**	39.0 ± 11.1	0.46 (−0.20–0.83)	43.2 ± 12.0
External oblique	75 %	0.97 (0.66–1.76)	21.2 ± 0.7	0.16 (−0.79–0.65)	37.0 ± 10.7
	85 %	0.69 (0.09–0.88)	33.5 ± 7.4	0.39 (−0.47–0.76)	46.7 ± 14.8
	95 %	0.48 (−0.33–0.80)	43.7 ± 22.1	0.40 (−0.45–0.76)	47.3 ± 33.3
	Mean	**0.71** (**0.19–0.92)**	32.8 ± 10.1	0.32 (−0.35–0.77)	43.7 ± 19.6
Lumbar sacral erector spinae	75 %	0.33 (−0.56–0.73)	13.5 ± 6.5	0.33 (−0.56–0.73)	23.4 ± 4.4
	85 %	0.61 (−0.07–0.85)	22.1 ± 4.6	0.76 (0.27–0.91)	28.0 ± 0.9
	95 %	0.52 (−0.25–0.81)	31.0 ± 7.1	0.71 (0.15–0.89)	28.4 ± 1.7
	Mean	0.49 (−0.16–0.84)	22.2 ± 6.1	**0.60** (**0.00–0.88)**	26.6 ± 2.3
Upper lumbar erector spinae	75 %	−0.54 (−1.44–0.21)	20.8 ± 6.1	−0.09 (−1.06–0.51)	28.7 ± 8.0
	85 %	−0.03 (−1.01–0.54)	24.1 ± 10.4	0.87 (0.57–0.95)	27.2 ± 5.0
	95 %	−0.10 (−1.08–0.50)	32.5 ± 23.7	0.53 (−0.23–0.82)	46.2 ± 15.0
	Mean	−0.23 (−0.73–0.44)	25.8 ± 13.4	0.44 (−0.22–0.82)	34.0 ± 9.3
Vastus lateralis	75 %	0.10 (−0.87–0.61)	22.8 ± 15.4	0.15 (−0.81–0.64)	28.8 ± 2.1
	85 %	0.53 (−0.23–0.82)	14.5 ± 0.2	0.39 (−0.47–0.76)	22.0 ± 2.5
	95 %	0.43 (−0.41–0.78)	17.1 ± 2.0	0.50 (−0.29–0.81)	24.3 ± 9.8
	Mean	0.35 (−0.32–0.79)	18.1 ± 5.9	0.35 (−0.32–0.78)	25.1 ± 4.8

acceptable CV (12.9 ± 3.9 %) but poor ICC (0.49), whereas RA presented unacceptable CV (29.4 ± 1.2 %) but fair ICC (0.6). However, it is well known that these 2 reliability indexes express different information; CV measures consistency of measurements within participants on separate occasions, whereas ICC measures the extent to which participants maintain the same rank within the group [4]. The latter is also affected by heterogeneity of the population, with more heterogeneous results within the group displaying higher ICCs when all other conditions are equal [4]. Indeed, this occurrence could at least partially explain the apparently contradicting results within the present study, particularly given the homogenous nature of our participants. Nevertheless, to our knowledge, we are the first authors to report *inter*day reliability of RMS for TMA, which should enable coaches and researchers to accurately account for "measurement noise" recorded from back squat exercise.

Importantly, the stability of a measure in response to condition(s) is critical and we confirmed that the RMS linear load effect on muscle activation for all sites in the eccentric phase can be repeated on a separate day (▶ **Fig. 3**). This RMS linear load effect during the back squat has also been demonstrated on similar muscle sites [3,32] but never repeated on separate days. Similarly, in the concentric phase we found a repeatable load effect for muscle activation in ULES, LSES and EO (▶ **Fig. 4**) with a tendency for RA. The relationship between load and activation of the muscles of the posterior chain in the concentric phase of the squat is fairly well established [3,36]. The functions of these muscles are to both stabilize the vertebral column and to resist flexion of the spine, both obvious challenges during the concentric phase of the squat.

During the concentric phase, RMS of most trunk muscles increased alongside load whereas VL remained unchanged. This has been demonstrated previously in some studies [3,11,12,32] but not in others [8]. In our study the concentric phase was performed explosively resulting in a classic power curve [7,45], whereas Balshaw & Hunter (2012) [8] controlled the ascent, which produced no such curve. The reason for this is explosive lifts require the individual to intuitively apply the necessary force as quickly as possible to overcome the resistance, whereas in controlled lifts time is fixed. Hence, a progressive increase in motor units will be recruited to lift larger loads [40] when the duration of the concentric phase is kept the same, which necessitates overcoming increased tension within the same timeframe. As the VL is a quadriceps muscle crossing the knee joint to assist in its extension, the neuromuscular recruitment patterns are likely to reflect the kinematic demands of back squat in the absence of fatigue. Whereas, the main role of the trunk muscles is to provide structure and stability in response to load and not to velocity as we have shown.

Conclusion

▼

We have shown that sEMG of most trunk muscles possess moderately acceptable *inter*day absolute reliability which was superior during controlled eccentric as opposed to uncontrolled eccentric back squat contractions. Whereas, *all* of these muscles show acceptable sensitivity as sEMG increases as load becomes heavier in both concentric and eccentric phases of the back squat. Therefore, as long as velocity is controlled during back squat exercise, sEMG of most trunk muscles will produce moderately acceptable levels of noise but with significant increases in response to higher load. Importantly, in the ascent the demand placed on the anterior stabilizers is reduced compared to during the descent or in comparison to posterior stabilizers in the ascent. Furthermore, this is the first study to demonstrate neuromuscular activation of trunk muscles reflecting changes in load rather than velocity unlike lower limb muscles, which are affected by both parameters.

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Study 4: Comparison of trunk muscle activation in back and hack squat

The most obvious characteristic of loaded free barbell squat is the requirement of the trunk, as an integral part of the kinetic chain, to stabilize the load through the full range of movement. Many previous studies investigating core stability compared stable squats to those performed on an unstable surface. Findings indicated that instability compromised force and power production in the squat without necessarily increasing trunk muscle activation. Furthermore, previous research had shown no difference in trunk muscle activation between the more stable Smith machine squat and free barbell squat at the same relative load (Schwanbeck, Chilibeck and Binsted, 2009). Hence, comparison of the free barbell squat to a more supported version, the hack squat, would facilitate the evaluation of the challenge on trunk stabilizers posed by the free barbell (Appendix 2). The use of equivalent relative loads meant that the absolute load in hack squat would be greater than back squat, suggesting higher lower limb activation in hack squat versus back squat.

In the neuromuscular trials of study 4, back and hack squat test order was fixed. This may be seen as a limitation according to strict scientific research design suggesting random test order. Similarly, in study 5, squat and countermovement jump tests preceded loaded back squats in all neuromuscular tests. In both cases this was done to prevent postactivation potentiation (PAP), defined as ‘transient increase in muscle contractile performance after previous contractile activity’ (Sale, 2002). Sale (2002) proposed that prior heavy load efforts increase activation low frequency portion of the force / frequency curve (Sale, 2002). Hence, performing hack squat trials at the same relative, but higher absolute loads prior to the back squat trials would arguably increase activation in subsequent back squat performance. Furthermore, it is well established that prior heavy squat efforts increase countermovement jump performance (Mitchell and Sale, 2011; Esformes and Bampouras, 2013). Consequently, in study 4 and 5 test order was fixed for all neuromuscular trails to ensure that hack squat preceded back squat, squat test loads were incremental and body weight jumps preceded squat trials.

The purpose of the fourth study was to compare trunk muscle activation in the free barbell back squat to the machine hack squat at 4 equivalent, moderate to heavy loads.

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Abstract

The hack squat (HS) is likely to produce a greater 1 repetition maximum (1RM) compared to the back squat (BS). This can be attributed to the support of the trunk during the HS compared to no support during BS. This support however, may compromise trunk muscle activation (TMA), therefore producing different training adaptations. Accordingly, the purpose of this study was to compare 1RM in BS and HS and TMA at 4 relative loads, 65, 75, 85 and 95% of maximal system mass. Ten males completed 3 test sessions:1) BS and HS 1RM, 2) HS & BS neuromuscular test familiarization, and, 3) Neuromuscular test for 3 reps at 4 loads for BS and HS. BS TMA was significantly greater ($p<0.05$) than HS for all muscles and phases except rectus abdominus in concentric phase. TMA increased ($p<0.05$) with load in all muscles for both exercises and phases apart from lumbar sacral erector spinae in HS eccentric phase. Mean HS 1RM and submaximal loads were significantly ($p<0.0001$) higher than the equivalent BS loads. Duration of the eccentric phase was higher ($p<0.01$) in HS than BS but not different in concentric phase. Duration increased significantly ($p<0.01$) with load in both exercises and both phases. Despite higher absolute tests loads in HS, TMA was higher in BS. TMA is sensitive to load in both exercises. BS is more effective than HS in activating the muscles of the trunk and therefore arguably more effective in developing trunk strength and stability for dynamic athletic performance.

Key words: back squat, hack squat, trunk muscles, neuromuscular, electromyography, core stability

INTRODUCTION

The squat exercise is a compound movement that engages all muscles below the shoulders including the lower limb. The primary purpose of both the back squat (BS) and hack squat (HS) are to develop strength and power in the lower limb[1–4]. Both are widely used for the development of performance capabilities for a variety of sports[2,5] and as a rehabilitative exercise for lower limb injuries and post-surgical programmes[1,6,7]. Recent research has focused on loaded compound exercises such the squat and deadlift as a method of developing trunk strength and stability. The hack squat (HS) has been used in a number of research training studies[8–10], however no trunk muscle activation data exists for HS.

Research investigating the BS[11–15], front squat[12,13,16], and overhead squat[15] have confirmed that the loaded, free barbell squat is an effective method of activating the stabilizing muscles of the trunk. There is also evidence that in BS magnitude of activation across the majority of muscle sites is sensitive to the external load[1,11,17,18]. As a result, a number of researchers concluded that BS is an effective method for developing dynamic trunk strength and stability for healthy function and athletic performance[11,17–19].

There are variations of the squat exercise performed in a machine supported set-up. These include leg press[6,20], HS[4] and Smith machine squats[14,21–23] and are generally performed at higher absolute loads than BS[14,21]. It is believed that these more stable versions of the squat compromise and reduce TMA due to biomechanical set-up and support[22,23]. Fletcher and Bagley (2014)[14] reported an 11% greater Smith machine one repetition maximum (1RM) compared to BS. Despite this, erector spinae electromyography (EMG activity was significantly greater in BS compared to the Smith machine squat 1RM test.

The HS offers more support than the Smith machine squat; it is commonly viewed as a safe version of the loaded squat exercise, especially suitable in the absence of established barbell squat technique and for rehabilitation programmes[4,24]. The HS is performed in a machine angled posteriorly at 45^o where force is applied and resisted through padded shoulder yokes. The participant’s back is positioned on a padded board offering greater support to the trunk during squat movement[4] contributing to higher loading capacity compared to Smith machine squat and BS. To our knowledge, there is no research comparing 1RM in HS to BS. However, untrained subjects developed a 1RM of over 250 kg after 8 weeks HS training[8,9]. This is equivalent to a relative 1RM of approximately 3.3 times body mass, greater that any previously reported BS relative 1RM. This suggests the supported characteristic of HS is accompanied by the ability to lift greater maximal loads than in free bar BS.

Centre of gravity of the person and external load, or the system load, in BS must remain over base of support[25] to prevent failure and or injury. As a result, force is resisted in the eccentric phase and expressed in the concentric phase through the line of gravity which determines how the loads are experienced by the affected muscles. When squatting in a linear motion machine, such as a Smith Machine or HS, the centre or line of gravity can safely sit outside the foot stance or the point where force is applied. This is the result of anterior foot placement which is made possible by the supported trunk and fixed external load. This introduces horizontal forces which potentially change load direction experienced by muscles of the body[25], including the prime movers and trunk stabilizers. To our knowledge there is no research describing or quantifying either trunk or lower limb muscle activation in HS.

Using a two dimensional model of a free body diagram, Abelbeck (2002)[25] assessed moments and work of the hip and knee joints for 6 foot positions anterior to the line of gravity. Position 1 was under the line of gravity and at position 6, knees were flexed to 90° and thighs horizontal. Each foot position away from the line of gravity resulted in a greater moment about both joints. Net work done at the knee decreased while it increased at the hip with each anterior foot position. HS is a tilted and supported version of a linear motion machine squat. Escamilla[6] (1998) measured activation of 6 muscles of the lower limb in leg press and squat exercise at 12RM. Foot placement in the leg press was anterior to the line of gravity equivalent to position 6 in Ablebeck's[25] (2002) study. Apart from biceps femoris in extension where activation was greater in the squat than leg press, there were no significant differences in activation between the two exercises for all muscles in both flexion and extension.

It has been established that TMA, across majority of muscle sites, is sensitive to increases in external load in BS[1,11,17,18]. It is also accepted that load capacity of HS is greater than for BS[8,9]. In the BS, stabilization of the trunk is necessary to ensure that the centre of gravity of the system load remain over the base of support for the eccentric and concentric phases. Anterior foot placement in the HS, facilitated by fixed external load and trunk support, resulted in higher work at the hip joint[25] but no meaningful increase in activation of leg muscles[6]. Trunk muscle activation under these conditions is unknown. While there is an appreciation of these differences in applied strength and conditioning, these have not been measured and quantified.

Accordingly, we hypothesize that the requirement to stabilize the bar in BS places greater demands on muscles of the trunk than greater absolute loads in the more supported HS. In accordance with this, objectives of the study were to; 1) determine 1RM for HS and BS within a strength trained cohort, 2) compare TMA in HS and BS in a range of relatively equivalent external loads, and 3) determine whether TMA was load sensitive in HS and BS.

METHODS

Experimental Approach to the Problem

All subjects attended 3 test sessions (Figure. 1). In the first, a 1RM test was conducted for BS and HS. In session 2, subjects completed the neuromuscular test protocol familiarization with loads calculated from the 1RM. In the third session, the neuromuscular test protocol was repeated while EMG and kinematic measures were taken. All tests were conducted 5 to 7 days apart.

All BS repetitions were performed according to technique described by Earle and Baechle (2000)[26]. Starting with the barbell in high bar position, on the trapezius across the back of the shoulders with hip and knee joints fully extended. Feet were placed shoulder width apart with legs externally rotated by 3-5^{o} so that that the toes were turned slightly out. Hack squats[24] were performed with the back placed against the padded surface, shoulders wedged under the yokes and feet placed shoulder width apart to the front of the footplate. Both squat versions comprised of a descent through knee and hip flexion to where mid-point of the thigh joint was below mid-point of the knee joint with a minimum knee flexion of 90^{o}. The transition between the descent and the ascent was visually assessed as the point where the top of the thighs were horizontal in BS and parallel to the footplate in HS. The load was returned to the start position by extending the hip and knees in a controlled manner as fast as possible. All BS were performed using barbells and discs approved by International Weightlifting Federation (Eleiko, Sweden). BS tests were conducted in a safety power cage (FT700 Power Cage, Fitness Technology, Skye, Australia) and HS in a plate loaded Bodymax CF800 Leg Press/Hack Squat Machine.

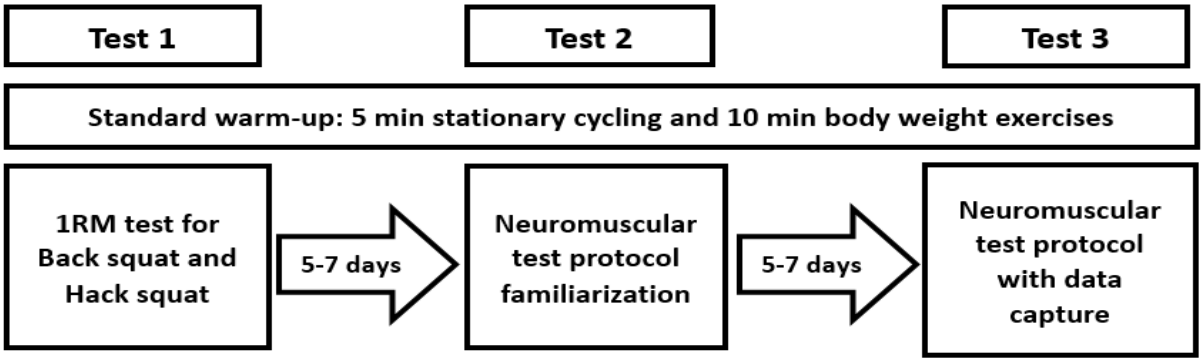

Figure 1. Experimental design illustrating the timing and content of the three test sessions and the standardised warm-up. 1RM – 1 repetition maximum.

Subjects

Ten males actively participating in regular strength training with at least 1 years' experience in BS exercise were recruited for the study. Using G*Power software (3.1) we calculated a minimum of 10 participants was required for 90% power from the effect size of RMS increase in the eccentric phase of BS from 75-95% load[17]. Subject characteristics were; age: 27 ± 8 years, body mass: 86 ± 8 kg, squat training age: 6 ± 5 years, BS 1RM: 142 ± 29 kg, relative BS 1RM: 1.7 ± 0.3, HS 1RM: 171 ± 34 kg and relative HS 1RM: 2.0 ± 0.4. In accordance with Declaration of Helsinki (2013)[27], the local research ethics committee granted approval for the study. The risks and potential benefits of the study were explained to all subjects prior to signing an informed consent form. Signed parental consent was recorded for the subjects under the age of 18. Subjects abstained from strenuous exercise and followed usual dietary habits for 24 hours prior to test sessions which were conducted at the same time of day to account for circadian variation[28].

Procedures

1RM testing

Following a standardised warm-up of 5 minutes stationary cycling and 10 minutes body weight exercises, subjects completed BS 1RM test according to an established protocol[26]. Barbell warm-up comprised 3-5 sets of diminishing repetitions at progressive loads determined for each subject from previous 1RM test results and current training loads. BS 1RM test was performed first followed by HS 1RM to avoid possible potentiation effect of higher absolute loads reported for HS[8,9]. 1RM test scores were recorded as highest load lifted successfully through required range of movement within 4 attempts in BS and HS. Subjects were instructed to control cadence of descent and perform ascent as fast as possible under control. Three minute rest periods were allocated between each warm-up and test set[24,29–31]. Correct squat depth for both exercises was established during warm-up sets and reinforced during testing by an experienced strength coach, the principle investigator, who conducted all tests.

Neuromuscular test load calculation

Test loads for sessions 2 and 3 were calculated using the system mass (SM)[17,32] approach. This is calculated by adding 88.6% of body mass to 1RM, which is equivalent to body mass minus the mass of the shanks and feet. This represents total load lifted vertically when performing the squat[32]. The neuromuscular test protocol comprised 2 BS warm-up sets of 10 repetitions at 45 and 55% SM, followed by 4 sets of 3 repetitions at 65, 75, 85 and 95% SM for BS and then HS. Back squat and HS test loads were determined according to following equation:

SM max = 1RM + (0.886 x body mass) (kg)

External test load = (SM max x percentage of SM) - (0.886 x body mass) (kg)

Familiarization and neuromuscular test trials

In test session 2, subjects completed the standardised warm-up and neuromuscular test protocol at individually calculated loads for BS and HS. During this familiarization session exercise technique, squat depth and rest times were rehearsed. In test session 3, subjects were prepared for EMG and kinematic data collection which was confirmed during 2 warm-up sets before proceeding to neuromuscular test protocol. Subjects were instructed to control descent and perform ascent as fast as possible under control for both BS and HS. Squat depth was monitored using linear transducer data and observation.

Kinematic data

The duration and displacement of eccentric and concentric phases of both exercises were measured by linear transducer (Celesco, PT5A, California, USA). The linear transducer was placed directly beneath, and attached to the barbell in BS. In HS it was placed adjacent to the footplate and attached at shoulder height to the sled of the HS machine to measure full displacement of the load along the 45° plane of travel[29,33].

A bespoke Matlab (Matlab R2010A, The Mathworks Inc., USA) programme was designed to identify initiation and completion of descent and ascent of the load in order to determine eccentric and concentric phases for EMG selection.

Electromyography

Muscle activity was measured from 5 sites on right-hand side of the body based on established bilateral symmetry of these muscles[34]; rectus abdominus (RA), external oblique (EO), lumbar sacral erector spinae (LSES), upper lumbar erector spinae (ULES) and vastus lateralis (VL)[11,23] using surface EMG (Biopac MP100, Biopac Systems Inc., Santa Barbara, CA). SENIAM (Surface Electromyography for Non-Invasive Assessment of Muscles) recommendations were followed for skin preparation and application of electrodes[35]. Hair was removed, sites abraded with emery paper and cleaned with an alcohol swab in preparation for two Ag-AgCl EL258S bipolar 8 mm diameter electrodes (Biopac Systems Inc., USA). These were housed in custom made soft rubber mould with 20 mm inter electrode distance. They were filled with conductive gel and fixed in position with transparent adhesive dressing. Electrodes were fixed longitudinally along muscle fibre orientation according to SENIAM (ULES and VL)[23], (LSES, ULES and VL) and[11] (RA, EO, LSES and ULES). EMG was sampled at a rate of 2000 Hz, anti-aliased with a 500 Hz low pass filter and root mean square processed (RMS). We have previously demonstrated acceptable absolute (CV%) and relative (ICC) reliability of mean RMS data for these trunk muscles in the back squat exercise at similar loads[17].

Mean RMS for eccentric and concentric phases were calculated from 3 reps for each load and exercise. Mean RMS data for 75, 85 and 95% SM for each phase of both exercises were normalized to mean RMS of concentric phase of 65% SM in BS and presented as mean ± SD percentage normalized RMS. It has been demonstrated that submaximal dynamic contraction, not maximal isometric contraction, offer more reliable amplitude for EMG normalization of trunk muscles in healthy controls and patients with lower back pain[33]. We have previously shown that submaximal dynamic normalization was far more reliable and sensitive than MVC methods in BS exercise for VL[17,33].

Statistical Analysis

Statistics were performed using GraphPad Prism version 6.07 for Windows, GraphPad Software, La Jolla California USA. Data were analysed with a 2-way repeated measures analysis of variance (ANOVA) for condition (x2) and load (encoder displacement and duration x2, RMS x3). 1RM data were analysed using paired *t*-tests. F ratios were considered significant at $p<0.05$. Significant condition effect was followed by *post-hoc* Sidak's procedure for multiple comparisons. All data are presented as mean ± SD for each phase of both exercises and all test loads. Where appropriate, 95% lower and upper confidence intervals (CI) and Cohen's *d* effect sizes (ES)[36] calculated by:

Cohen's $d = Mean_1 - Mean_2 / SD_{pooled}$, where $SD_{pooled} = \sqrt{[(SD_1^2 + SD_2^2) / 2]}$.

ES were then interpreted as <0.2 = trivial, ≥0.2 - 0.5 = small, ≥0.5 - ≤0.8 = moderate, ≥0.8 = large [36].

RESULTS

Electromyography

In the eccentric phase RMS was significantly (p<0.05 to p<0.0001) greater in BS vs. HS in 7 of the 9 test loads for EO, ULES and LSES (Table 1). However, there was no difference in RA RMS in the eccentric phase between BS and HS; whereas concentric RMS was significantly ($p<0.05$ to $p<0.0001$) greater in BS than HS in all muscle sites and in 8 out of 12 instances (Table 2).

Table 1. Normalized mean percentage RMS in the eccentric phase, Mean diff., 95% confidence intervals, *p*-values, Cohen's *d* and effect size (ES) and for hack squats and back squats performed at the 3 test loads, 75, 85 and 95% SM.

					95% CI of diff.			Cohen's	
	Test load	Hack squat (mean ±SD)	Back squat (mean ±SD)	Mean Diff.	Lower	Upper	*P*	*d*	ES
RA	75%	64 ±30	65 ±22	-0.9	-18.7	16.9	>0.999	-0.03	Trivial
	85%	73 ±34	68 ±22	4.7	-13.1	22.5	>0.999	0.16	Small
	95%	86 ±35	82 ±22	3.7	-14.1	21.6	>0.999	0.13	Small
EO	75%	57 ±31	87 ±33	-29.4	-48.8	-9.9	0.003*	-0.91	Moderate
	85%	62 ±27	80 ±26	-19.2	-38.6	0.3	0.054	-0.72	Moderate
	95%	70 ±31	94 ±27	-24.0	-43.4	-4.5	0.013*	-0.84	Moderate
ULES	75%	92 ±38	118 ±56	-26.1	-40.8	-11.5	0.001*	-0.55	Small
	85%	84 ±39	130 ±47	-45.9	-60.5	-31.3	<0.0001*	-1.07	Moderate
	95%	85 ±41	155 ±64	-69.2	-83.8	-54.6	<0.0001*	-1.29	Large
LSES	75%	72 ±21	88 ±12	-16.0	-34.1	2.1	0.096	-0.92	Moderate
	85%	75 ±19	95 ±15	-19.8	-38.0	-1.7	0.030*	-1.14	Moderate
	95%	75 ±24	107 ±22	-32.5	-50.7	-14.4	0.001*	-1.44	Large

Note: *Significant greater mean RMS in back squat compared to hack squat (*p*<0.05).

Table 2. Normalized mean percentage RMS in the concentric phase, Mean diff., 95% confidence intervals, *p*-values, Cohen's *d* and effect size (ES) and for hack squats and back squats performed at the 3 test loads, 75, 85 and 95% SM.

					95% CI of diff.			Cohen's	
	Test load	Hack squat (mean ± SD)	Back squat (mean ± SD)	Mean Diff.	Lower	Upper	*P*	*d*	ES
	75%	96 ±51	132 ±68	-36.4	-73.6	0.5	0.054	-0.61	Moderate
RA	85%	117 ±68	159 ±60	-41.6	-78.6	-4.7	0.024*	-0.65	Moderate
	95%	138 ±67	166 ±64	-27.4	-64.3	9.6	0.199	-0.42	Small
	75%	81 ±34	142 ±42	-61.1	-102.1	-20.1	0.003*	-1.60	Large
EO	85%	99 ±26	188 ±90	-89.0	-130.0	-47.9	<0.0001*	-1.34	Large
	95%	123 ±43	224 ±114	-100.7	-141.8	-59.7	<0.0001*	-1.16	Moderate
	75%	112 ±42	152 ±46	-39.8	-64.5	-1.4	0.039*	-0.90	Moderate
ULES	85%	133 ±90	169 ±49	-36.1	-84.7	-21.6	0.001*	-0.50	Small
	95%	128 ±62	230 ±107	-102.2	-112.3	-49.2	<0.0001*	-1.17	Moderate
	75%	97 ±37	130 ±27	-33.0	-91.3	11.8	0.170	-1.02	Moderate
LSES	85%	105 ±31	159 ±43	-53.1	-87.7	15.4	0.243	-1.42	Large
	95%	110 ±32	191 ±50	-80.7	-153.7	-50.6	0.000*	-1.92	Large

Note: *Significant greater mean RMS in back squat compared to hack squat (*p*<0.05).

RMS increased with load in the following trunk muscle sites in the eccentric phase for both exercises (Figure 2): RA ($F_{(2, 18)}$ = 13.52, *p*<0.001) EO ($F_{(2, 18)}$ = 5.258 *p*<0.05), ULES $F_{(2, 18)}$ = 6.374 *p*<0.01). There was no eccentric load effect for LSES for both BS and HS. RMS increased with load in all muscle sites and both exercises in the concentric phase (Figure 3): RA ($F_{(2, 18)}$ = 7.795 *p*<0.01), EO $F_{(2, 18)}$ = 14.70 *p*<0.001), LSES ($F_{(2, 18)}$ = 18.76 *p*<0.001) and ULES $F_{(2, 18)}$ = 6.035 *p*<0.01).

Mean VL RMS was significantly ($F_{(1, 9)}$ = 5.846 *p*<0.05) higher for BS vs HS in the concentric phase and a tendency in the eccentric phase where *post-hoc* analysis demonstrated significance for 3 test loads (75% SM *p* <0.0001, 85% SM *p* <0.01, 95% SM *p*<0.0001). Muscle activation in VL produced a significant load effect in both exercises for both phases: eccentric ($F_{(2, 18)}$ = 18.85 *p*<0.001) concentric ($F_{(2, 18)}$ = 3.711

p<0.05).

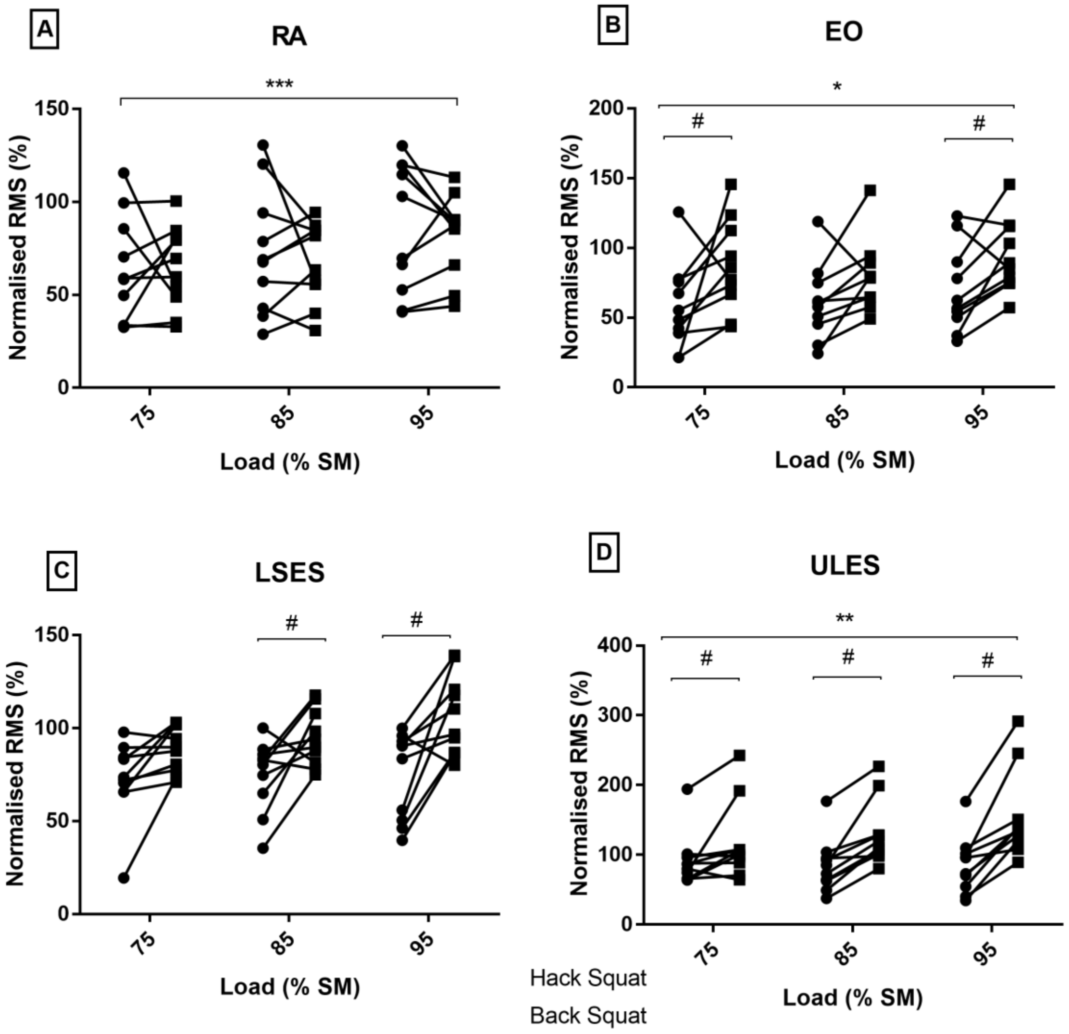

Figure 2. Mean RMS for the eccentric phase for 3 test loads, 75, 85 and 95% SM for the 4 trunk muscle sites; A – rectus abdominus, B – external oblique, C – lumbar sacral erector spinae and D – upper lumbar erector spinae. Significant load effect: * (*p*<0.05), ** (*p*<0.01), *** (*p*<0.001) and significant difference between BS and HS: # *p*<0.05 and *p*<0.0001.

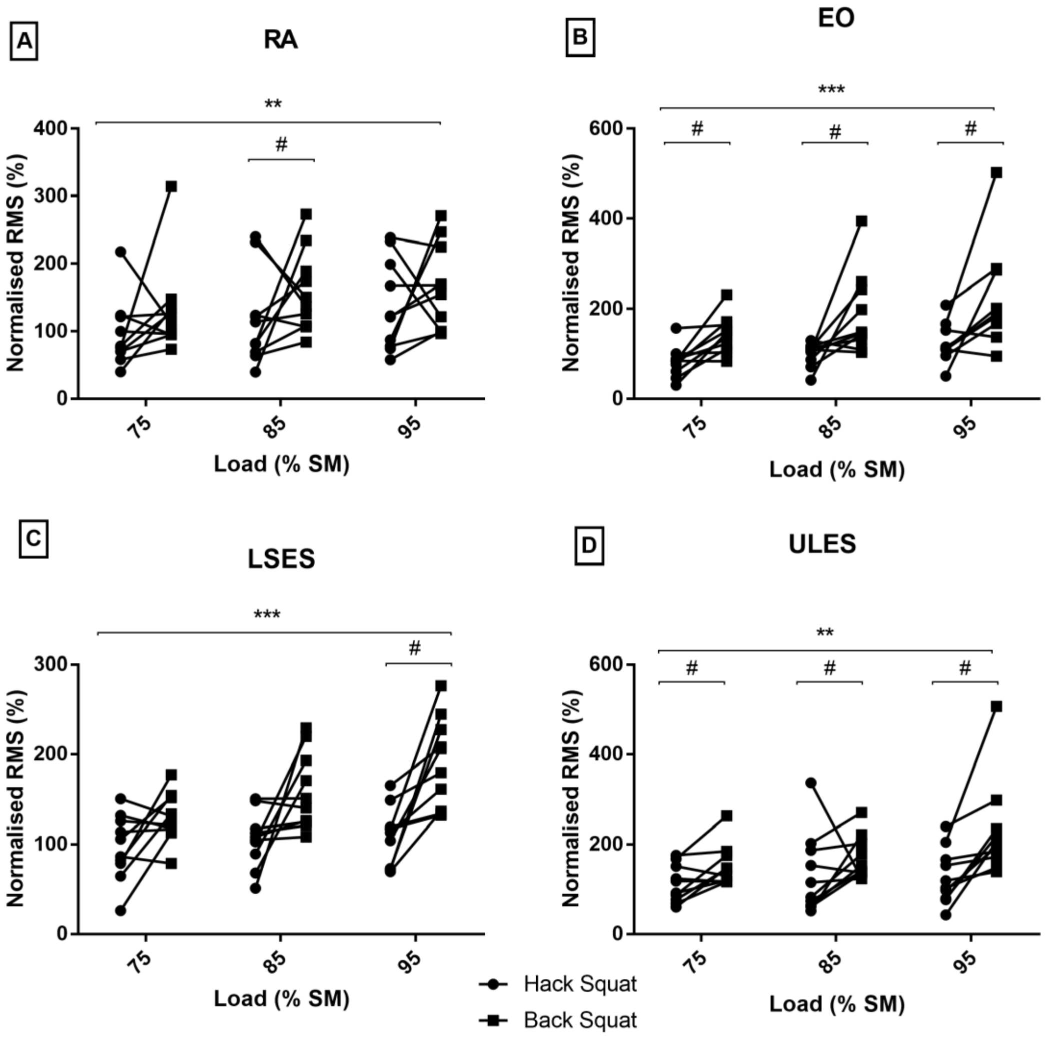

Figure 3. Mean RMS for the concentric phase for 3 test loads, 75, 85 and 95% SM for the 4 trunk muscle sites; A – rectus abdominus, B – external oblique, C – lumbar sacral erector spinae and D – upper lumbar erector spinae. Significant load effect: ** (*p*<0.01), *** (*p*<0.001) and significant difference between BS and HS: # (*p*<0.05 to *p*<0.0001).

1RM tests and test loads

The mean HS 1RM was significantly (*p* <0.0001) higher at 171 ± 34 kg when compared to 142 ± 29 kg in BS. As a result relative test loads at 65, 75, 85 and 95% SM were significantly greater in HS than BS by 16.5, 17.5, 20.5 and 23.0 kg respectively ($F_{(1, 9)}$ = 19.94 *p*<0.01).

Kinematic measures

Eccentric displacement in BS was significantly ($F_{(1, 9)}$ = 33.62 *p*<0.001) greater than in HS for 4 test loads by 21.4, 20.8, 21.5 and 22.2 cm (Figure. 2A). Eccentric displacement decreased significantly ($F_{(3, 27)}$ = 5.931 *p*<0.01) with load in both BS and HS. Duration of eccentric phase was significantly ($F_{(1, 9)} = 18.54$ $p<0.01$) greater in HS compared to BS for all test loads (Figure.3A). Duration significantly ($F_{(3, 27)} = 5.371$ $p<0.01$) increased with load for both BS and HS for eccentric phase with a significant ($F_{(3, 27)} = 2.968$ $p<0.05$) interaction effect which occurred from progressively reduced differences from 20.4% (65% SM) to 10.6 (95% SM).

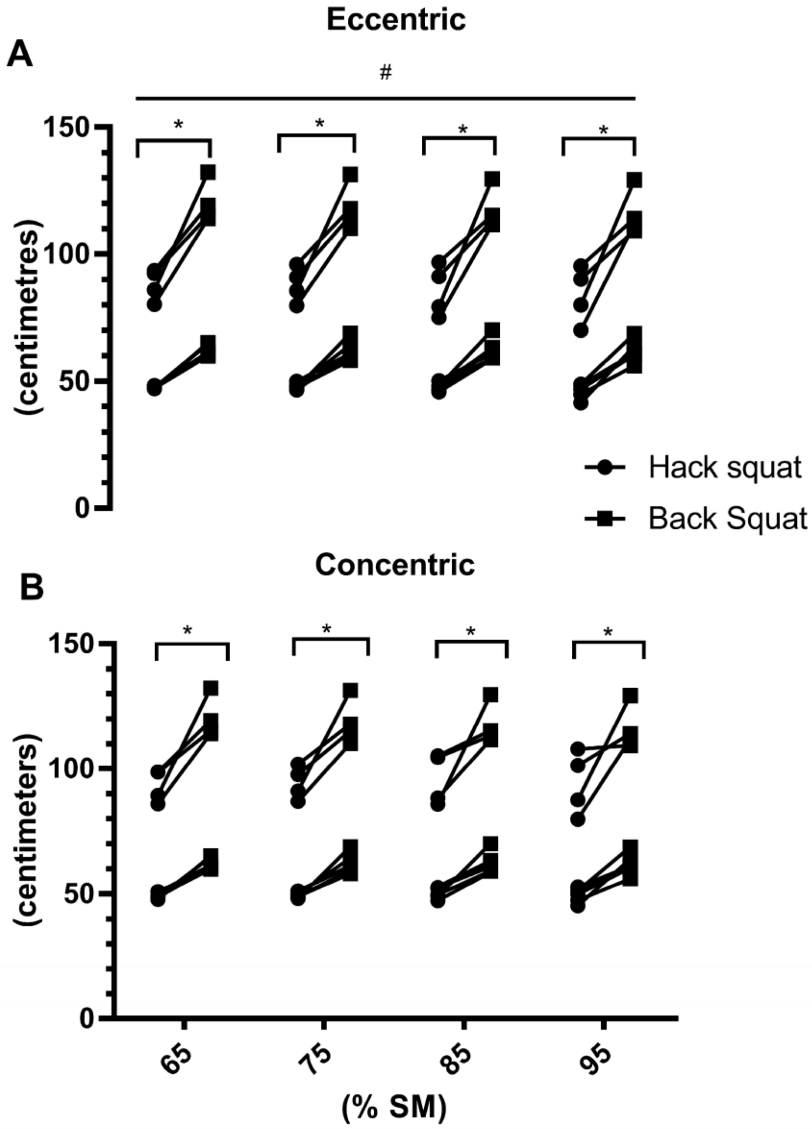

Figure 4. Kinematic data for the BS and HS where panel A is eccentric displacement and B concentric displacement. Significant load effect in both conditions: # ($p<0.01$), and significant difference between HS and BS: * ($p<0.001$).

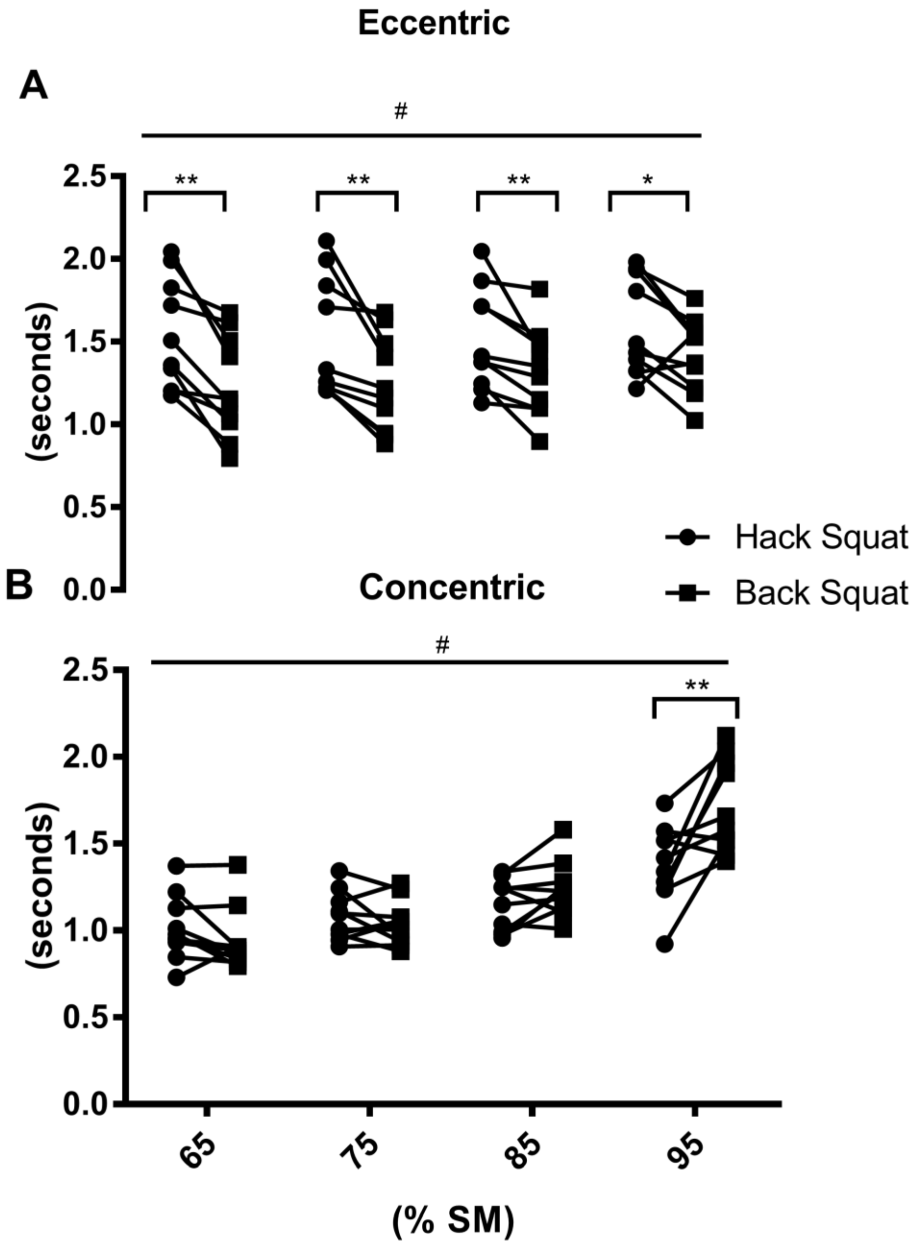

Figure 5. Kinematic data for the BS and HS where panel A is eccentric duration and B concentric duration. Significant load effect in both conditions: # (*p*<0.01) and significant difference between HS and BS: * (*p*<0.001) ** (*p*<0.0001).

Concentric displacement was significantly ($F_{(1, 9)} = 26.30$ *p*<0.001) greater in BS than HS (Figure. 2B) for all loads. There was no displacement load effect for either exercise in the concentric phase. Concentric phase duration increased significantly ($F_{(3, 27)} = 115.5$ *p*<0.0001) for BS and HS alongside increases in load. There were no differences between BS and HS for duration of concentric phase during tests at 65, 75 and 85% SM. However, there was a significant ($F_{(3, 27)} = 14.82$ *p*<0.0001) interaction effect where BS duration at 95% SM was significantly (*p*<0.0001) greater than HS (Figure. 3B).

DISCUSSION

This is the first study to compare maximal strength and TMA in HS and BS. Anecdotal evidence that HS maximal strength capacity is greater than BS is confirmed under scientific research conditions. As hypothesized, TMA in BS was greater than HS in the majority of muscle sites, at the same relative loads. Furthermore, TMA in both exercises increased with each load increment which were similar to those commonly used in applied strength and conditioning practice.

TMA was greater in BS vs. HS for all measured muscles during both phases, with the exception of rectus abdominus in the eccentric phase which demonstrated no such differences. This largely agrees with our hypothesis, although the rectus abdominus finding was also unsurprising given the previous equivocal reports of this muscle's RMS activity in Smith Machine vs. BS[14; 22]. The likely cause of this variance is the flexed trunk position during most of BS, which causes skin to fold in the rectus abdominus region, thus moving electrodes away from activated motor units and inevitably increasing measurement variability. While the role of rectus abdominus as a stabilizer in squats remains unclear it appears from our data that rectus abdominus contribution to stabilization increases with load in both phases of both exercises, and this is greater in the concentric phase of BS.

In the lateral stabilizers, activation of external oblique muscle was significantly greater in BS than HS in all instances and both phases apart from 85% SM in eccentric phase. The shared function of rectus abdominus and external oblique muscles are to create intra-abdominal pressure during exertion through the trunk[37]. Individually rectus abdominus controls lumbar extension and external oblique controls lateral flexion and rotation of the trunk[37]. Logically, these functions will be challenged more in BS than HS which suggest greater trunk muscle adaptation potential in the free bar BS.

Activation of posterior stabilizers, lumbar sacral erector spinae and upper lumbar erector spinae muscles was greater in BS than HS in 9 out of 12 instances. Importantly, in these 2 muscle sites at the heaviest load, 95% SM, activation was higher in BS than HS. Hamlyn and coworkers[11] (2007) using the mean RMS calculated from a 1 second sample from each phase, eccentric and concentric, showed that LSES and ULES activation was more than twofold higher in back squat at 80% 1RM compared the bodyweight squats. The purpose of erector spinae muscle complex is to extend the trunk, or in the case of BS prevent trunk flexion[14,15,17]. In the free bar exercise this challenge is greater where back and trunk are unsupported. During the descent activation was significantly higher in BS than HS for all three loads in ULES and for 85 and 95% SM in LSES. This was similar for the ascent however the magnitude of activation was greater for both exercises and all three loads in both ULES and LSES (Tables 1 and 2) (Concentric RMS: 97-230% vs Eccentric RMS: 92-155%). The higher activation of trunk stabilizers in the concentric compared to eccentric phase has been reported in a number of studies.[12,13,15,38]

Activation of external oblique and erector spinae muscles have been shown to increase alongside load in BS with submaximal loads of 50 and 75% 1RM[39]. In 2 studies where higher loads were used, the primary purpose was to compare TMA in deadlift exercise and a range of dynamic[18] and isometric[11] trunk exercises. Both studies reported a load effect in the posterior trunk muscles for BS but this was not significant. In our recent study we demonstrated a significant load effect in BS for all trunk muscles in the eccentric phase and for lumbar sacral erector spinae, upper lumbar erector spinae and external oblique in the concentric phase[17]. In the current study we found a load effect for both exercises, both phases and all muscle sites except for lumbar sacral erector spinae in the eccentric phase in both BS and HS. LSES activation in the BS increased by load in the eccentric phase (Table 1) but this did not reach significance, possibly due to the size of the sample. Importantly, loads in both our studies reflected loads commonly used during training for development of athletic performance. Therefore, TMA responses are representative of what may be expected for this type of activity in moderate to well strength trained populations.

In this study where load was significantly higher in HS, vastus lateralis RMS was greater in the BS for all loads and both phases. Vastus lateralis RMS increased with load in both BS and HS which is well established for this muscle during both eccentric[17] and concentric phases [29]. This is similar to earlier work from our laboratory where there was higher activation of vastus lateralis in concentric phase at 100% 3RM compared to 75% 3RM despite higher power produced in the lower load test effort[29]. Fundamentally, this demonstrates the large effect comparatively lower forces, external load in BS vs HS, have on increasing activation of prime lower limb muscle where no external support is provided for lifting weights vertically against gravity.

Mean 1RM for HS was 29 kg (18%) greater than BS, significantly more than the 11% difference between Smith Machine and BS 1RM previously reported[14]. As such, we demonstrated that absolute test loads at 65, 75, 85 and 95% SM were higher in HS than

BS. Eccentric displacement was on average 22 cm less in HS than BS across 4 test loads. This can be explained by the positioning in HS machine in which the moment about both knee and hip joint increase as the feet move anterior to the line of gravity[25]. At the same time, work done at the knee probably decreased due to reduced range of movement, while compensatory work at the hip may have increased. Therefore, the reduced overall displacement (external marker) and the higher absolute load (internal marker) in the HS possibly resulted in a greater moment and therefore work at the hip compared to the BS[25].

Eccentric displacement decreased across the 4 test loads for both squat versions. This is possibly due to compressive force of the incremental external loads causing spine shrinkage[40]. Wisleder[40] showed that an external load equivalent to body mass resulted in a mean shrinkage of 3.9 mm. This shrinkage would result in a progressively lower start point for the descent with each higher test load. This would reduce eccentric displacement despite completing a full depth squat. Interestingly, concentric displacement was not affected by load, probably due to subjects following the instruction to complete this phase as fast as possible, which may have ended in full extension overriding the shrinkage.

The eccentric phase of the BS was significantly faster for each load despite a significantly greater displacement. There was no difference between the duration of HS and the BS in the concentric phase apart from the heaviest load (95% SM) where HS was performed quicker than BS. This suggests that the instruction to ascend as fast as possible compensated the greater BS displacement and HS load respectively, in the concentric phase for 3 loads. While the instruction to descend in a controlled manner was applied to both exercises, BS descent was faster than HS. This occurred despite the greater support offered by the HS machine and the greater range of movement in the BS. A possible explanation could be familiarity with BS training reflected by mean squat training age of 6 years (Range: 1-17 years) compared to the relative novelty of the HS exercise within this group.

In our earlier study we established reliability of surface EMG in measuring trunk muscle activation in the BS[17]. The current study has confirmed and expanded those findings. The kinematic characteristics of the unsupported free bar BS are a greater range of movement, faster descent and lower absolute external loads than the HS. Importantly, this study has shown that under those conditions the BS places greater demands on the trunk stabilizers than the HS and that this increases with load. Three factors therefore explain greater trunk muscle activation in the BS, greater range of movement, faster descent and importantly, the requirement to control the unsupported external load through the full kinetic chain. This included lower limbs, hips and pelvis and, as shown by this study, the trunk. We have shown that both the BS and HS challenge the trunk stabilizers and that this activation increases in both exercises with load. However, BS is a significantly more effective method of activating the trunk stabilizers than HS. The conclusion therefore is that free barbell loaded squats are an effective exercise for the development of dynamic trunk strength and stability and for both BS and HS, trunk stability training effect is enhanced by increasing external load.

PRACTICAL APPLICATIONS

This study presents a number of interesting and novel findings particularly applicable to evidence based, applied strength and conditioning coaches. The key finding is that the free barbell back squat elicits greater trunk muscle activation than HS at the same relative load. This strengthens the case made in previous studies[11,17–19] and confirms applied anecdotal evidence that back squat is an effective method of developing dynamic trunk strength and stability. Similarly, we have presented novel research evidence to demonstrate and quantify greater absolute maximal strength capacity in HS compared to BS for a cohort of well-trained subjects. A further novel finding was the greater activation of vastus lateralis in the concentric phase of BS compared to HS despite significantly higher absolute HS loads. We also confirmed previous research[1,11,17,18] showing that increases in external load in both the BS and HS produce greater trunk muscle activation.

The implication of these findings for applied setting, is that free barbell squat is an effective exercise for the development of dynamic strength and stability in the trunk. The more stable hack squat is less effective for this purpose, however in both exercises trunk stabilization training effect can be enhanced by increasing external load.

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Study 5: Impact of back squat training status on trunk muscle activation in squats and jump tests

We established reliability and sensitivity of sEMG measurement of trunk muscle activation in back squat and demonstrated acute effect of moderate to heavy loads. Comparisons between back and hack squat confirmed that free barbell version placed higher demands on trunk stabilizers than more supported hack squat. The only previous research suggested no difference in trunk muscle activation between 8RM in free barbell back squat and Smith machine squat (Schwanbeck, Chilibeck and Binsted, 2009).

Two questions remained; how does trunk muscle activation change through the full range of squat movement for different loads and secondly how does regular progressive squat training impact on trunk muscle activation in the squat jump (SJ), countermovement jump (CMJ) and back squat? To answer the first question, we used a synchronised electro-mechanical knee goniometer (Appendix 5) to enable the analysis of RMS data in three segments or tertiles for each phase of the squat (Appendix 6). In the second question, we compared trunk muscle activation in participants with different squat training status and strength while performing loaded squats and bodyweight jumps.

Title:

Back squat strength reduces trunk muscle activation in squats and jump tests

Short Title:

Squat strength lowers trunk muscle activation

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ABSTRACT

Purpose: To measure impact of back squat training status, on trunk muscle activation in moderate to heavy back squat, squat jump (SJ) and countermovement jump (CMJ). Squat training status was determined by absolute, relative 1 repetition maximum (RM) and squat training age.

Methods: Fifty one males, with squat training experience of one year or more completed 2 test sessions. Squat 1RM was tested first and participants were assigned to either strong group (SG), middle group (MG) or weak group (WG) according to relative squat 1RM. In test 2, EMG data were collected for four trunk muscle sites; rectus abdominus, external oblique, lumbar sacral erector spinae and upper lumbar erector spinae while completing 3 reps of SJ, CMJ and squat at 65, 75 and 95% system mass max. Squat and jump phases were determined from a linear transducer and 30^{o} tertiles for the eccentric and concentric phases, from a knee goniometer.

Results: Normalized RMS for each muscle site in both squat phases significantly ($p<0.0001$) increased by similar amounts with load from 75 to 95% SM, hence data were combined for all further analysis. Trunk muscle activation was significantly ($p<0.05$) lower in SG vs WG in eccentric back squat for all loads and in heaviest concentric load (95% SM). Concentric and flight phase RMS in both jumps was lower ($p<0.05$) in SG vs WG. For all test loads RMS increased significantly ($p<0.0001$) for each respective tertile of squat descent and first concentric tertile where highest activation was recorded.

Conclusion: Greater squat strength resulted in lower trunk muscle activation in back squat and jumps compared weaker participants. Full depth parallel squats are important to achieve highest trunk muscle activation.

Key Words: surface EMG, neuromuscular activation, trunk stability, muscular strength

INTRODUCTION

Several researchers have demonstrated effectiveness of loaded, compound exercises in activating trunk stabilizers (1–5). Traditional core stability exercises, characterized by isolated (6) isometric (7) exercises do not reflect dynamic human movement in un-injured individuals (8). It is therefore appropriate to investigate the efficacy of dynamic, integrated, loaded exercises in developing trunk stability in healthy individuals. Several studies have reported effective trunk muscle activation for selected dynamic exercises (9), squats (1,2,10–12), standing, seated, unilateral and bilateral exercises (13) and deadlifts (3). Effects of specific exercise characteristics on trunk muscle activation have also been investigated, including; instability (5), external load (1,2), trunk rotation (14,15) and movement velocity (14,15). Hence, there is good evidence that variations of loaded, dynamic, compound exercises, including back squat, are effective in activating trunk stabilizer muscles (16,17). Furthermore, while acute trunk muscle activation response to back squats at different loads is well described, chronic adaptation to progressive squat training over an extended period is not known.

We have previously demonstrated acceptable reliability and sensitivity of sEMG in measuring trunk muscle activation in back squat (1). We and others have shown a load effect for trunk muscle activation in this exercise (1–4,18). Activation increased by load in eccentric and concentric phases for all four trunk muscle sites; rectus abdominus (RA), external oblique (EO), lumbar sacral erector spinae (LSES) and upper lumbar erector spinae (ULES) (1,2). Higher trunk muscle activation in concentric phase of loaded back squat compared to the descent has been previously demonstrated in our laboratory (1,2) and by others (10,18). Hip adductor and quadriceps activation in the back squat was higher in the last 30^o of descent and first 30^o of ascent than any other part of both phases of loaded parallel squats (19,20). Hence, there is evidence that for effective lower limb muscle activation the minimal optimal squat range is 90° knee flexion or parallel (19,20). Therefore, understanding how trunk muscle activation changes through the full range of the loaded back squat is important to inform squat depth for most effective development of dynamic trunk stability. Furthermore, understanding how trunk muscles adapt to squat training across all sections of the squat descent and ascent provides evidence for the efficacy of back squat in developing trunk stability.

The impact of maximal strength and training status on muscle activation in the squat has been reported for muscles of the lower limb but not trunk stabilizers. Muscle activation of quadriceps in back squat 1RM test was significantly higher in trained compared to untrained subjects (21). Higher activation in trained participants explained, in part, their capacity to lift higher 1RM loads and complete more reps in the single set to failure (21). Also, rate of force development (RFD) has been demonstrated to be underpinned by higher mean average EMG and rate of EMG rise in quadriceps following 14 weeks of lower limb training (22). This is likely to be due to higher efferent neural drive which explains both improved strength and RFD performance (22). Therefore, higher activation during maximal contractions in trained individuals is associated with increased motor unit recruitment, synchronization of motor unit discharge and improved intermuscular coordination (21,22). These neuromuscular measures therefore explain higher performances in strength, power and strength endurance associated with strength training status. As such, we would expect to see similar training adaptations in the trunk muscles, yet this has not been investigated.

Effective strength training for athletic performance should incorporate methods that address deficits across the force velocity curve by overloading the full range of this spectrum (23,24). The back squat, SJ and CMJ are established methods of training all components of the force velocity curve including; maximal strength, strength speed, speed strength and the stretch shortening cycle (23,25,26). The relationship between back squat strength and performance in CMJ and SJ is well established (27–29). There is evidence that loaded squats have an acute potentiation effect on jumps (28,29) and a chronic training effect on improving jump performance (25,27). However, despite the obvious importance of trunk stability in transferring jump force and power through full kinetic chain, there is no evidence for acute and chronic trunk muscle activation in SJ and CMJ. Hence, this is of interest, along with the impact of adaptation to back squat training on trunk stability in CMJ and SJ performance.

Accordingly, the purpose of this study is to investigate impact of back squat strength and training status on trunk muscle activation in CMJ, SJ at bodyweight and back squat at moderate to heavy loads. Our hypothesis is that trunk muscle activation will be higher in participants with high relative back squat strength compared to those with lower squat training age and strength.

METHODS

Participants

Fifty-one males actively participating in regular strength training with at least one years' experience in back squat exercise were recruited for this study (Age: 22.3 ± 3.1 years, Body mass: 81.5 ± 11.2 kg). Recruitment targeted participants across the full range of relative back squat strength according to squat training age and 1RM capability. Participants were recruited from a range of university and club sports including; rugby, football, American football, tennis and swimming. All participants were injury free at the time of the study and had no history of injury with a bearing on back squat exercise. University of Stirling, School of Sport Research Ethics Committee in accordance with the Helsinki declaration (2013) granted ethical approval for the study. All participants gave informed written consent prior to testing.

After extraction and scrutiny, back squat sEMG data for one participant was removed due to being corrupted and unsuitable for analysis.

Experimental Design

Participants attended the laboratory on two separate occasions with 3-7 days between visits. During the first visit, body mass, height and training history were collected, followed by back squat 1RM test. During the second visit, surface sEMG and kinematic data were collected for 3 repetitions of SJ, CMJ and back squat at 3 progressive loads based on 1RM. Fixed testing order was used to avoid potentiation of the exercise tested second (28). Both sessions began with a standardized warm-up and participants rested for 3 min between each test set. Testing was conducted at the same time of day to account for circadian rhythm (30). Participants followed usual dietary intake and avoided strenuous exercises for 24 hours prior to testing.

Based on relative back squat 1RM, participants were assigned to one of 3 groups for comparative analysis by dividing evenly into; Strong group (SG), Middle group (MG) and Weak group (WG) based in relative back squat 1RM (Table 1).

Back squat 1RM test

Participants performed a standardized warm-up before back squat 1RM testing. This comprised of a range of dynamic, compound body weight exercises that progressed to loaded barbell back squat for 3 sets of 10-8 reps at 20kg, 45 and 55% 1RM. The standardized warm-up was repeated prior to neuromuscular test protocol and on both occasion technical requirements of SJ, CMJ and back squat depth were rehearsed and confirmed.

The back squat 1RM and neuromuscular test protocols were conducted in a safety squat rack (FT700 Power Cage, Fitness Technology, Skye, Australia) using competition approved barbell and discs (Eleiko, Sweden). All squat efforts were assessed for correct technique and depth by primary investigator, an experienced strength coach. The 1RM test protocol, previously described by McGuigan (2016) (31), was used to determine maximal load completed with correct technique and depth within 5 attempts. Participants were required to descend to where top of the thighs were horizontal or lower followed by a continuous ascent to full extension of hips and knees. Participants were instructed to control descent cadence and drive the bar up as fast as possible, with control, during ascent (32). Three-minute seated rest was scheduled between each test repetition.

Back squat training and test loads calculated according to system mass max (SM) assume that 89% of body mass is included in external load (1,2). The remaining 11%, (i.e. shanks and feet) do not move vertically in squat exercise.

- SM = 1RM + (0.89 x body mass) (kg)
- External load = (SM x percentage of SM) – (0.89 x body mass) (kg).

This method is established in back squat research (1,2) and was applied to calculate back squat warm-up and test loads in this study

Kinematic data

A single linear transducer (Celesco, PT5A, California, USA), fitted to the safety squat cage directly above the participant was attached to the middle of the barbell for back squat and a wooden dowel for jump tests. Linear transducer measured displacement and time from initiation of the descent, transition from descent to ascent to end of ascent for each rep of back squat and jump tests. This facilitated identification of eccentric and concentric phases and jump height for SJ and CMJ tests.

An electro mechanical goniometer, incorporating a high precision rotary potentiometer (6657s-1-103, Bourns, Riverside, CA, USA), was attached to right knee to measure flexion and extension. The fixed rotary potentiometer of the goniometer was placed at the center of rotation of the knee. The fixed arm of the goniometer was attached to the lateral thigh by surgical tape. The actuating goniometer arm was attached to the lateral calf by Velcro onto a neoprene sleeve, reinforced with surgical tape. The actuating arm incorporated three hinges to allow naturel extension through the movement and a compact swiveling gimbal to accommodate small angular movement. The goniometer was manually calibrated to fixed plastic protractor (33).

A threshold was established within Acqknowledge software (Version 4.4.2, Biopac Systems Inc, CA) to produce a digital output to indicate correct knee flexion had been reached that corresponded to required back squat depth. Correct execution of technique resulted in an audible sound. When the sound was not heard, the repetition was excluded and repeated. Minimum knee flexion that corresponded to back squat depth, where thighs were horizontal or lower, was established for each participant during the first warm-up set with the barbell.

Linear transducer data synchronized to root mean square (RMS) data were used to demarcate eccentric and concentric phases in back squat and jump tests and flight phase in jump tests. Back squat goniometer data were used to segment eccentric and concentric phases into three equal 30^o (degree) tertiles for RMS analysis (19).

Neuromuscular test

On arrival, participants were weighed and screened for injury or illness prior to electrodes being fixed to 4 trunk muscle sites. Muscle sites were shaved, abraded and cleaned with an alcohol swab (34). At each muscle site, 2 adhesive electrodes (Ambu WhiteSensor WS, Ambu, Cambridgeshire, UK) were attached longitudinally along muscle fibre orientation with a 20 mm inter-electrode space according to Surface EMG (sEMG) for Non-Invasive Assessment of Muscles (SENIAM) guidelines (34). Electrodes from two muscle sites were connected to a BioNomadix 2 Ch. EMG Wireless

Transmitter (BN-EMG2). Two wireless transmitters were secured in a harness; one on upper back for posterior muscle sites; lumbar sacral erector spinae (LSES) and upper lumbar erector spinae (ULES), and the second for rectus abdominus (RA) and external oblique (EO) on the mid-chest. All connector cables were secured to minimize artifact noise and position of transmitters did not interfere with execution of back squat or jumps.

Standardized warm-up was completed up to and including SJ and CMJ rehearsal, thereafter participants were prepared for neuromuscular and kinematic data capture. Wireless transmitters were attached and goniometer fixed to the right knee. Transmitters were matched with a receiver unit in Biopac MP150 (Biopac Systems Inc., Santa Barbara, CA) which facilitated transmission of high resolution sEMG signal at a rate of 2000 Hz. Three back squat warm-up sets followed: 10 reps at 20 kg (barbell), 10 reps at 45% SM, 8 reps at 55% SM. Neuromuscular and kinematic data signals were confirmed and corrected where necessary during these sets. Minimum squat depth was determined during 10 reps at 20 kg and programmed into the Biopac system via the knee goniometer.

The barbell was replaced by a wooden dowel (< 0.25 kg) and attached to the linear transducer for SJ and CMJ tests. In both jump tests, participants were instructed to reset completely between each rep, perform maximal concentric efforts and ensure dowel remained in contact with shoulders throughout. In SJ, participants began by squatting to parallel, where thighs were horizontal and paused for count of 3 before jumping. Squat depth in CMJ was self-selected and followed immediately by a maximal jump effort (26). Three-minute seated rest was allocated between each test set.

The wooden dowel was replaced with 20 kg barbell (Eleiko) and back squat tests were conducted for 3 reps at 65, 75 and 95% SM with 3 minutes rest between each set.

Back squat test loads were calculated using system mass max formula, incorporating individual 1RM score and body mass measured at start of neuromuscular test session.

Surface Electromyography

Trunk muscle sEMG was measured from 4 sites on right-hand side of the body; RA, EO, LSES and ULES. Bilateral symmetry has been established for these muscles (35). Reliability of sEMG analysis of trunk muscle activation at these sites has been demonstrated in our laboratory (1,2) and by others (4,36).

sEMG was sampled at a rate of 2000 Hz and anti-aliased with a 500 Hz low pass filter. Raw sEMG signal was processed via Biopac MP150 amplifiers by applying an averaged RMS filter with a rolling 100-ms wide Bartlett window. Once processed, mean RMS was extracted for each phase using the synchronized linear transducer data, and for each tertile using synchronized knee goniometer signal. Mean RMS for 3 reps of all test sets in back squat and jump tests were normalized to mean RMS of concentric phase at 65% SM. Normalizing RMS data to a dynamic submaximal data point is well established for trunk muscles in healthy participants and has been effective in measuring trunk muscle activation in back squat studies (1,2).

Data analysis

Muscle activation data are presented as mean (±SD) normalized RMS percentage and were analyzed and reported for all participants (n-50) and by group (WG, MG and SG). Back squat neuromuscular test data was analyzed and presented for concentric and eccentric phase and by tertile for each phase (19). Knee goniometer data was used to divide eccentric and concentric phases into 3 equal 30^{o} segments or tertiles for RMS analysis (19). Tertiles are referred to as E-1, E-2, E-3, C-1, C-2, C-3, where E-1 is first tertile of the descent, C-1 first tertile of ascent and C-3, final tertile leading back to full extension.

RMS data for the jumps are presented for 3 phases; eccentric, concentric and flight. The concentric RMS sample was limited to the segment of that phase where participant was in contact with the ground and able to apply downward force. The end of the concentric phase identified from linear transducer data as concentric or upward displacement point that corresponded with upright standing position at start of eccentric phase. The flight phase was from the point of leaving the ground to point where peak displacement was reached. This began at the end of concentric phase, from the point where displacement exceeded upright standing position.

Statistical analysis

All statistical analyses were performed on GraphPad Prism software (Version 7.00, La Jolla, CA). Normal distribution of relative back squat 1RM data was established using D'Agostino & Pearson normality test (alpha=*0.05*). A two-way analysis of variance (ANOVA) was used to determine the differences in participant data (back squat 1RM, SJ and CMJ height and training history) and trunk muscle activation (normalized RMS percentage) measures between groups and test conditions. Level of significance of ($p<0.05$) was selected to determine statistical differences.

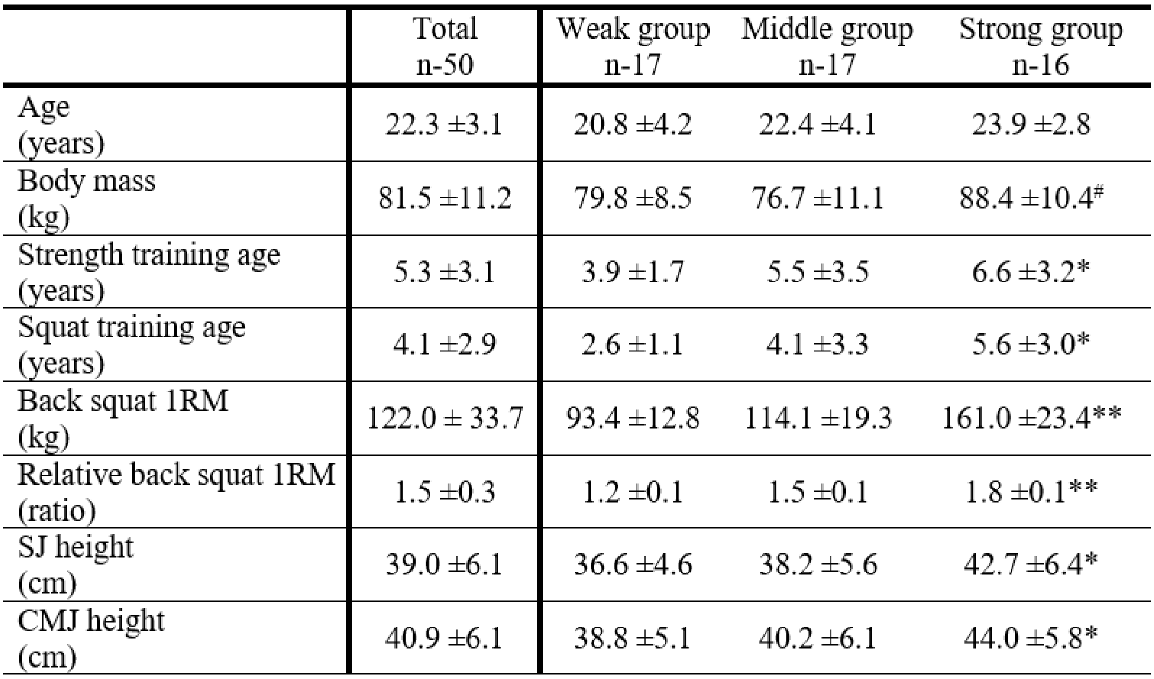
RESULTS

Participant data

Mean back squat 1RM for all participants was 122 kg (SD ± 34 kg) which translated to a relative 1RM (1RM/Body mass, kg) of 1.5 (SD ± 0.3, Range: 0.8 to 2.2) (Table 1). Relative back squat 1RM data passed D'Agostino & Pearson normality test ($p<0.05$). The three groups were significantly different according to absolute back squat 1RM ($F_{2, 47} = 51.1$, $p<0.0001$) and relative back squat 1RM ($F_{2, 47} = 125.7$, $p<0.001$). WG and SG were significantly different for strength training age ($F_{2, 47} = 3.6$, $p<0.05$), squat training age ($F_{2, 47} = 5.0$, $p<0.05$), SJ height ($F_{2, 94} = 8.8$, $p<0.01$) and CMJ height

($F_{2, 94} = 8.8$), $p<0.05$). Body mass was significantly higher in SG compared to MG ($F_{2, 47} = 5.6$, $p<0.01$).

Table 1. Mean (±SD) descriptive data for all participants (Total) and by group.



	Total n-50	Weak group n-17	Middle group n-17	Strong group n-16
Age (years)	22.3 ±3.1	20.8 ±4.2	22.4 ±4.1	23.9 ±2.8
Body mass (kg)	81.5 ±11.2	79.8 ±8.5	76.7 ±11.1	88.4 ±10.4#
Strength training age (years)	5.3 ±3.1	3.9 ±1.7	5.5 ±3.5	6.6 ±3.2*
Squat training age (years)	4.1 ±2.9	2.6 ±1.1	4.1 ±3.3	5.6 ±3.0*
Back squat 1RM (kg)	122.0 ± 33.7	93.4 ±12.8	114.1 ±19.3	161.0 ±23.4**
Relative back squat 1RM (ratio)	1.5 ±0.3	1.2 ±0.1	1.5 ±0.1	1.8 ±0.1**
SJ height (cm)	39.0 ±6.1	36.6 ±4.6	38.2 ±5.6	42.7 ±6.4*
CMJ height (cm)	40.9 ±6.1	38.8 ±5.1	40.2 ±6.1	44.0 ±5.8*

1RM – 1 Repetition maximum, SJ – Squat jump, CMJ – Countermovement jump, WG - Weak group, MG – Middle group, SG – Strong group. Significant differences: * WG vs SG (p>0.05), # MG vs SG (p>0.01) and ** all groups (p>0.01)

Neuromuscular analysis: Back squat

All participants. RMS for all participants combined increased significantly alongside load from 75 to 95% SM for all individual muscle sites in eccentric phase: RA-18%, EO-17%, LSES-33% and ULES-36% ($F_{2, 588} = 47.9$, $p < 0.0001$) and concentric phase: RA-45%, EO-63%, LSES-46% and ULES-79% ($F_{2, 588} = 225.3$, $p < 0.0001$). As a result, RMS data for 4 muscle sites was combined for all further analysis.

Concentric RMS for each test load is significantly ($F_{1, 294} = 93.1$, $p<0.0001$) greater than eccentric activation by: 15% (65% SM), 20% (75% SM) and 52% (95% SM). Activation in both phases increased significantly ($F_{2, 294} = 126.6$, $p<0.0001$) with each load: 65-75% SM-14%, 65-95% SM-56% and 75-95% SM-42% and showed significant interaction effect ($F_{2, 294} = 15.3$, $p<0.0001$) across the three test loads.

Group analysis. When assessed by group, RMS in the SG was significantly lower than WG for all test loads in eccentric phase ($F_{2, 141} = 26.7$, $p<0.05$) and at 95% SM in concentric phase ($F_{2, 141} = 26.7$, $p<0.05$) (Figure 1). There were no group differences at 65 and 75% SM in concentric phase ($F_{2, 141} = 1.8$, $p<0.17$).

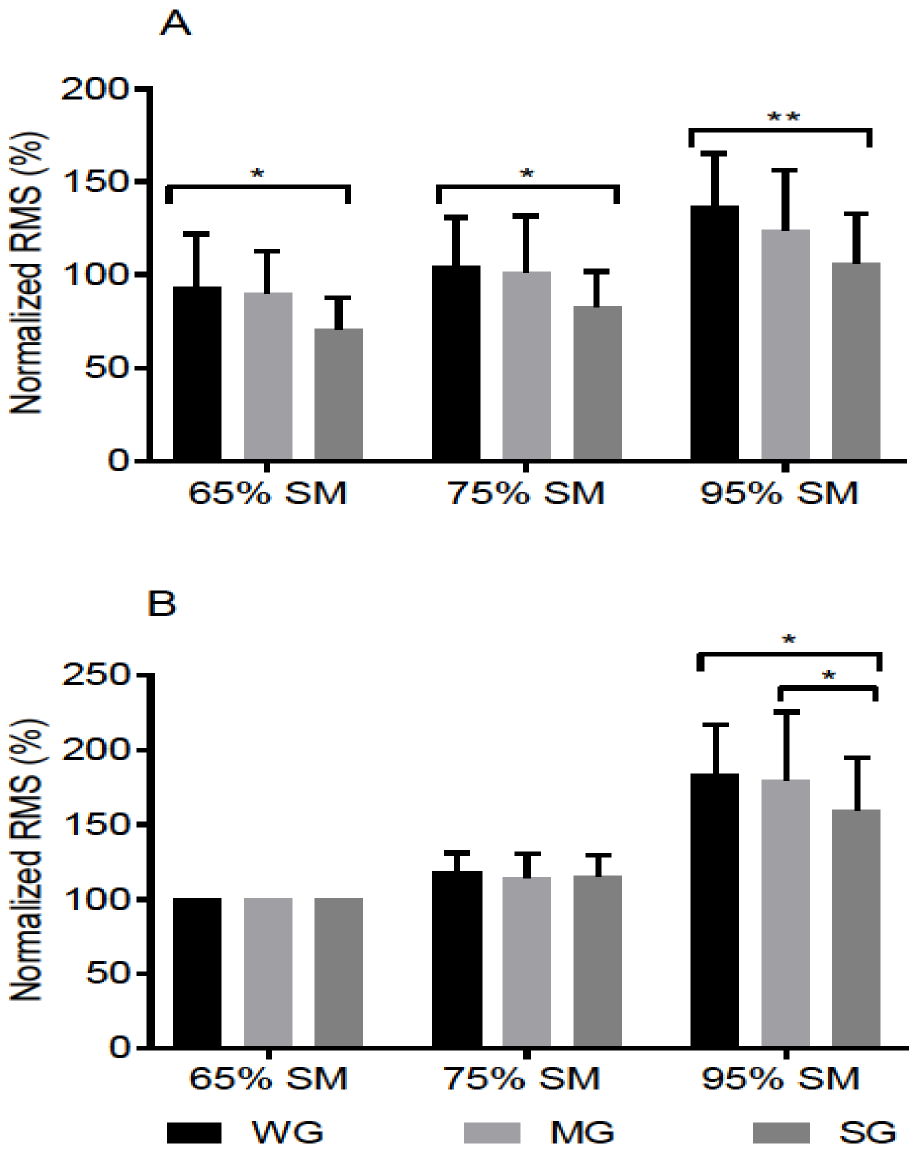


Figure 1. Normalized back squat mean RMS percentage for 3 groups (WG, MG and SG), 3 test loads; 65, 75 and 95% SM in the; A. Eccentric phase and B. Concentric phase. Significant differences: * $p<0.05$, ** $p<0.01$. (SM – system mass max, WG - weak group, MG – middle group, SG – strong group, SM – system mass max and RMS – root mean squared)

Group analysis by tertile. RMS is lower in SG compared to WG in all 3 eccentric tertiles for all loads apart from tertile E-1 at 95% SM ($F_{5, 235} = 35.6$, $p<0.0001$) (Figure 2). In tertile C-1 at 95% SM, both the WG and MG have higher activation than SG ($F_{2, 47} = 3.6$, $p<0.01$). There were no differences between groups in all concentric tertiles at 65 and 75% SM and for C-2 and C-3 at 95% SM.

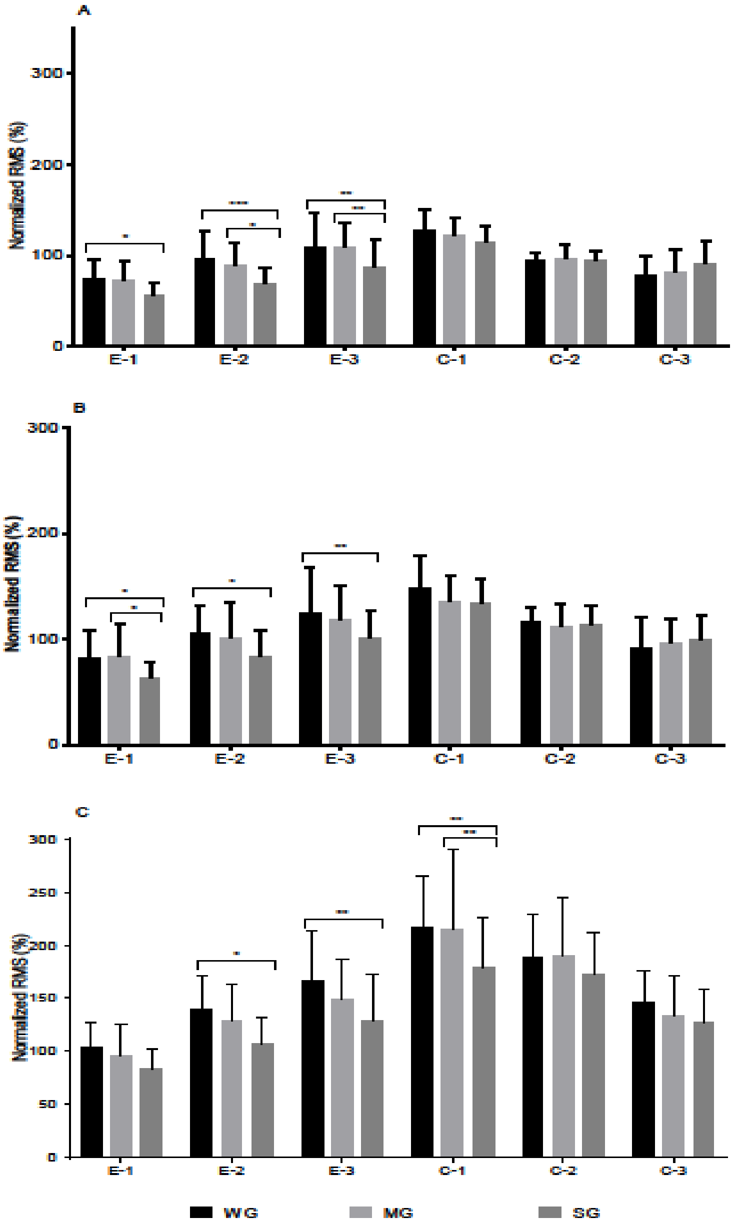


Figure 2. Normalized back squat mean RMS percentage for 3 groups in 3 tertiles for eccentric and concentric phases. Test loads: A. 65% SM, B. 75% SM and C. 95% SM. Significant differences: * $p<0.05$, ** $p<0.01$, *** $p<0.001$. (E – eccentric, C – concentric, WG - weak group, MG – middle group, SG – strong group and RMS – root mean squared)

All participant analysis by tertile. RMS increased significantly ($F_{5, 245} = 77.2$, *p*<0.0001) for each descending tertile (E1, E2 and E3) of the squat and peaked during the first part of the upward movement (Tertile C1) for each test load (Figure 3). RMS in the final two concentric tertiles (C2 and C3) decreased as ascent progressed to full extension of hips and knees. There was a significant load effect ($F_{2, 98} = 209.8$, p<0.0001) and interaction ($F_{10, 490} = 29.3$, *p*<0.0001) by tertile across the three test loads.

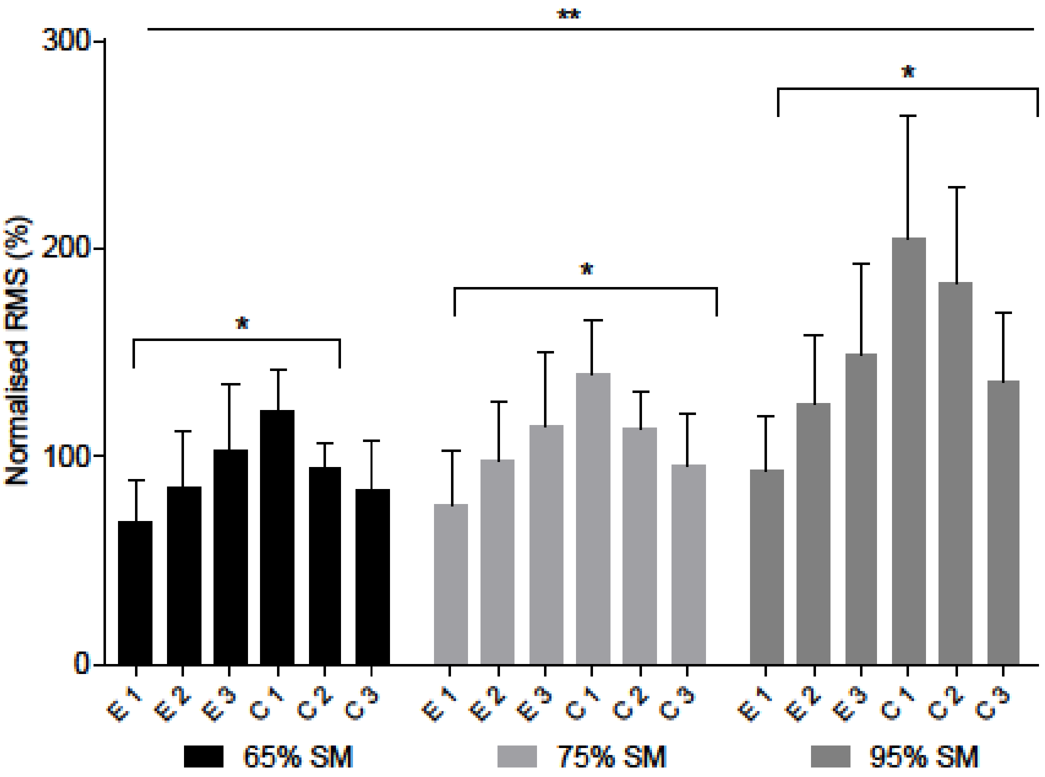


Figure 3. Normalized back squat mean RMS percentage for all participants, 3 tertiles for each phase, eccentric and concentric and 3 test loads; 65, 75 and 95% SM. * Significant differences between tertiles within each load, *p*<0.0001, ** Significant load effect *p*<0.0001. (SM – system mass max, E – eccentric and C – concentric, RMS – root mean squared)

Neuromuscular analysis: SJ and CMJ

Group analysis. In concentric and flight phase of both jumps, activation is significantly lower in SG compared to WG ($F_{2, 144} = 54.5$, *p* <0.05) (Figure 4). MG activation was higher than SG in concentric phase of SJ ($F_{2, 144} = 54.5$, *p* <0.05).

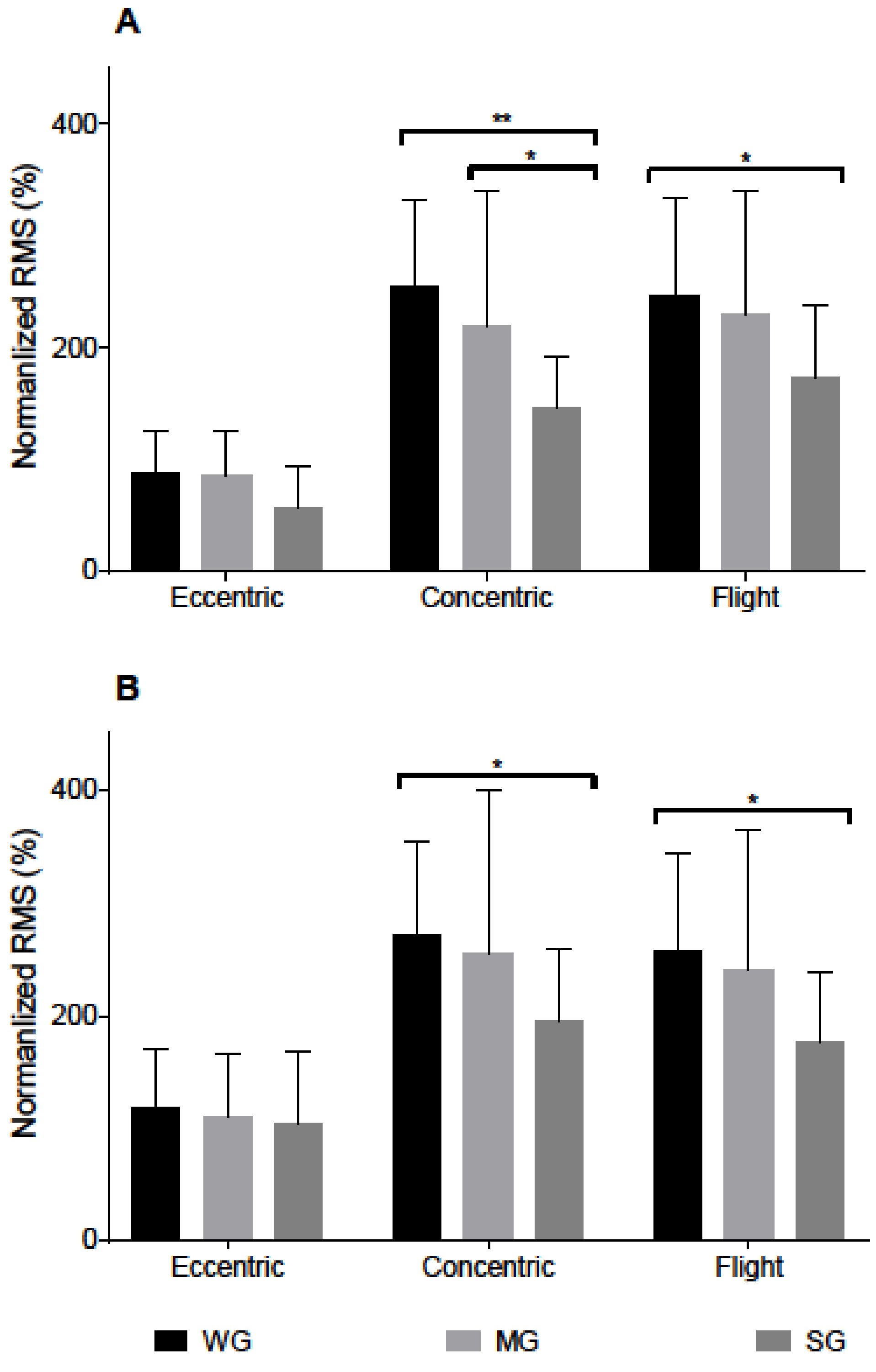


Figure 4. Normalized SJ and CMJ mean RMS percentage for 3 groups in eccentric, concentric and flight phase; A. SJ and B. CMJ. Significant difference between groups * $p<0.05$, ** $p<0.001$. (SJ - squat jump and CMJ – countermovement jump, WG - weak group, MG – middle group, SG – strong group and RMS – root mean squared)

All participant analysis. RMS was higher in concentric and flight phase compared to eccentric phase for both jumps ($F_{2, 300} = 75.8$, $p<0.0001$) (Figure 5). There were no differences in RMS between concentric and flight phase for both jumps (SJ: $p<0.91$, CMJ: $p<0.41$) nor between the SJ and CMJ for all phases (Ecc: $p<0.14$, Con: $p<0.13$, Flight: $p<0.99$).

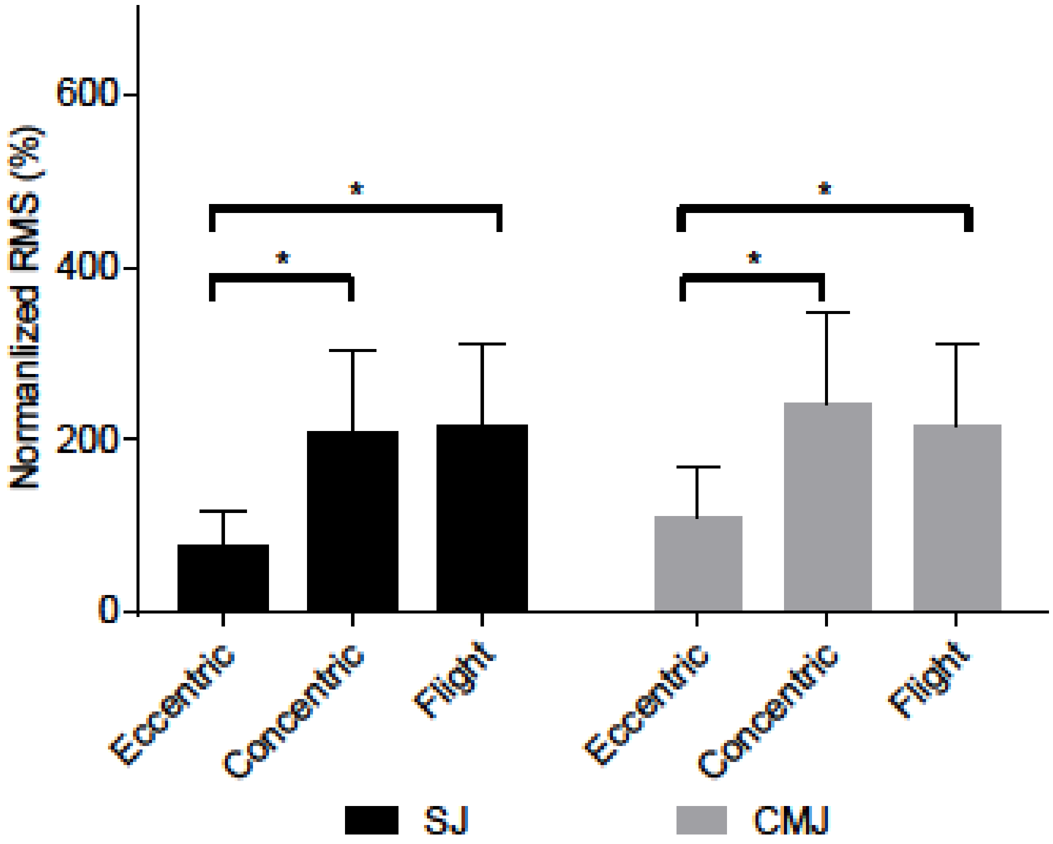


Figure 5. Normalized SJ and CMJ mean RMS percentage for all participants in eccentric, concentric and flight phase. * Significant difference between phases $p<0.0001$. (SM – system mass max, SJ - squat jump, CMJ – countermovement jump and RMS – root mean squared)

DISCUSSION

This study measured the impact of back squat training status on trunk muscle activation in squats, SJ and CMJ. High levels of back squat strength resulted in lower trunk muscle activation in squats at moderate to heavy loads and in bodyweight jumps. Detailed trunk muscle activation changes through the descent and ascent of the back squat were analyzed in order to understand the impact of squat depth on activation of trunk stabilizers. Trunk muscle activation was highest in the lowest segments of both descent and ascent of the parallel squat for all test loads. According to absolute and relative maximal back squat 1RM strength, participants represented a wide range of strength levels, confirmed by normal distribution and significant between group differences in 1RM test scores.

Evidence that higher back squat strength results in greater lower limb agonist EMG response to maximal strength effort guided our hypothesis for trunk muscle activation (21,22). We found against this hypothesis. Higher levels of back squat strength resulted in lower levels of trunk muscle activation in eccentric back squat at moderate and high loads and in the concentric phase, at high loads. The lower levels of trunk muscle activation in SG compared to WG occurred at the same relative back squat loads, which in fact were higher absolute loads. Hence, higher absolute and relative back squat strength from regular progressive squat training is achieved, in part through the development of more effective and efficient trunk stabilization. This is reflected in tertile analysis of RMS by group. Trunk muscle activation in SG was significantly lower than WG for first 3 tertiles of eccentric phase and in heaviest load, 95% SM, in first concentric tertile. This finding indicated that for phases of the squat that placed highest demand on trunk stabilizers, the strongest participants had lower trunk muscle activation than weaker participants. Repeated activation of these stabilizers through regular progressive loaded squat training to this depth results in important neuromuscular adaptations for athletic performance.

The current study confirmed our previous findings that increases in external load result in increased trunk muscle activation for both eccentric and concentric phases of back squat (1,2). We also showed higher activation in concentric compared to eccentric phase for all participants combined. There is evidence that lower limb muscle activation is highest in the final part of the eccentric and first part of concentric phase of the parallel squat (19,20). Parallel squats have been shown to be safe (37) and result in a number of positive physiological and performance adaptations (38). Higher trunk muscle activation in the current study occurred in the final 30° of the descent and first 30° of ascent, reinforcing the importance of squatting to parallel to maximize trunk muscle activation. This data confirms the importance of squatting to this depth to elicit optimal activation and ensure neuromuscular adaptations that underpin efficient performance.

Higher levels of back squat strength translated to lower trunk muscle activation in concentric and flight phase of SJ and CMJ. The SG, with higher jump heights in SJ and CMJ, had significantly lower trunk muscle activation than the WG. This demonstrates transfer of adaptation from a high load, low velocity, compound exercise, the back squat, to a dynamic and power oriented movement, central to performance in many sports and athletics activities. It also suggests that improved efficiency in dynamic trunk stabilization mechanisms from heavy squat training may transfer positively to other dynamic athletic activities.

This study is the first to report trunk muscle activation in squat jump and countermovement jump. McBride et al (2008) found concentric iEMG of the agonists (vastus lateralis and vastus medialis) did not differ between SJ and CMJ, despite higher CMJ concentric peak force, velocity and jump height (39). This is likely due to high levels of agonist pre-activity and eccentric muscle activity in CMJ and stationary SJ start position (39). Similarly, our study found no difference between SJ and CMJ trunk muscle activity in eccentric, concentric and flight phase despite marginally higher CMJ jump height. For both jumps, concentric and flight phase trunk muscle activation was higher than eccentric activation. Mean normalized RMS percentage in concentric and flight phase was approximately 200% indicating high demand for trunk stabilization during these two phases. Particularly interesting is the high activation of the trunk stabilizers during the flight phase where there is no ground contact and therefore direct force application. Our data provides evidence that there is a high demand on trunk stability in the flight phase of jumps. Importantly, we show that back squat training results in neuromuscular adaptations that improve efficiency of the trunk stabilizers under these conditions.

It has been suggested that improved jump performance as a result of back squat training is possibly due to biomechanical similarity in body position of the squat exercise and jump tests (40). Consequently, it may be that morphological training adaptations to squat training develops trunk muscle strength which increases trunk stiffness resulting in greater transfer of ground reaction force and therefore jump height (40). We have previously demonstrated that loaded free barbell back squat resulted in higher trunk muscle activation than the more supported machine based hack squat for the same relative, moderate to heavy loads (2). Our interpretation was that back squat was characterized by greater range of movement, faster descent and importantly, control of unsupported external load through full kinetic chain, including lower limbs, hips, pelvis and trunk. The current study therefore confirms previous speculation (40) that trunk muscle adaptations to back squat training results in improved trunk stability and stiffness. These adaptations facilitate greater resistance and transfer of force and power in both the ground based concentric and flight phases of explosive, dynamic activities.

While this was not a training study the most important findings have direct applied relevance. Firstly, a relative back squat 1RM of greater than 170% body mass acquired through squat training resulted in lower activation of the trunk stabilizers in the eccentric and concentric phase of heavy load squats compared to participants with a relative squat 1RM of less than 140% of body mass. This neuromuscular adaptation corresponds with increased load carrying capacity through the full kinetic chain, including the trunk. In the applied setting, it is accepted that the most important early adaptation to progressive loaded back squat training is primarily development of the trunk's capacity to maintain the position of the external load over the center of the base of support in the sagittal plane. The implication of this finding for strength and conditioning confirms and describes the neuromuscular adaptation in trunk stabilizers central to improved overall squat performance through training. The biomechanical and neuromuscular analysis of this adaptation, capacity to resist forward trunk flexion, within the context of overall squat training adaptation, is an important area for future research.

Secondly, the most relevant finding for athletic performance, is that the adaptation to squat training translates directly to dynamic body weight jumps, where trunk stabilization in higher jumps require lower muscle activation. Specifically, this neuromuscular adaptation in trunk stabilizers becomes significant in squat trained participants in the concentric phase, once downward force is applied, and subsequent flight phase occurs. Arguably, this greater stability presents as greater trunk stiffness in dynamic jumps, facilitating more effective resistance and transfer of force generated by lower limb in the concentric phase of the jump. Trunk muscle activation in the flight phase is not significantly different to that in concentric phase within each group. This highlights the importance of trunk stability and stiffness required for effective kinetic control while not in contact with the ground. Which explains, in part, the greater ground reaction force and higher jump heights in squat training studies (25,40–42). Furthermore, this greater trunk stiffness contributing to more effective transfer of concentric force into jump performance, comes at relatively lower activation of the trunk stabilizers. Logically, it would be expected that this adaptation translates to trunk control in other dynamic ground based athletic activities, such as sprinting, and represents an area for future research. This evidence means that loaded barbell squat can be regarded as an effective method of developing efficient trunk stability for performance in a range of dynamic sporting activities.

The third key finding with applied relevance is impact of squat depth on activation of trunk stabilizers. Knee and hip angle specific strength training to improve running and sprinting performance has been debated for some time in the literature (37). The case has been made to avoid extreme range of movement, characteristic of parallel and full squats in favour of quarter squats. Primarily, this is due to knee and hip actions that are similar to the target activity (37,43–46). This finding, higher trunk muscle activation in the deepest part of the squat, ensures that where development of trunk stability is a training aim, parallel or full squats are more effective than partial squats.

The use of surface electromyography to measure neuromuscular electrical activity reliably during dynamic, loaded activities has been challenged in the literature. However, we previously demonstrated reliability and sensitivity of sEMG in measuring lower limb (33,47) and trunk (1) muscle activity in back squat. Furthermore, to increase accuracy of our neuromuscular data, we followed a dynamic method of normalization to a submaximal reference point which has been found more accurate in analysis of trunk muscle RMS data in back squat (1).

CONCLUSION

This is the first study to show that back squat strength training adaptations reduce trunk muscle activation in squat jump, countermovement jump and moderate to heavy squats. Trunk muscle activation in back squat at moderate to heavy loads is highest in the last third of squat descent and first third of ascent, confirming importance of parallel squats for optimal trunk stability development. Trunk muscle activation in these demanding segments of the parallel back squat was significantly lower in trained participants compared to those less trained. Hence, the loaded parallel barbell back squat is an effective training method for development of efficient trunk stability. Furthermore, back squat training adaptations are effective in developing stability in dynamic jump performance.

These findings highlight possible future research areas including; 1) the impact of squat training adaptation on trunk muscle and whole body performance in acceleration, sprint and change of direction speed, and 2) impact of squat strength, acquired through chronic training, on acute fatigue of the trunk stabilizers in dynamic athletic activities.

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Thesis Conclusion

Thesis summary

The first study, the literature review confirmed that *trunk muscle activation increased with squat load and was greater in the concentric phase compared to the descent*. Furthermore, the review revealed the need to *establish reliability and sensitivity of surface EMG in measuring back squat trunk muscle activation* (Clark, Lambert and Hunter, 2012).

The survey found *that the majority of respondents were aligned to the more integrated definition of core anatomy and function, and hence supported a functional approach to exercise selection for developing trunk stability*. Despite this clear alignment with the scientific literature, *several participants remained in support of an isolated approach for the development of core stability.*

In the third study, we *demonstrated acceptable interday reliability and sensitivity of sEMG in measuring trunk muscle activation in barbell loaded back squat*. Importantly, we confirmed that in a well trained group, *activation increased significantly in response to increases in load equivalent to 10 percent of squat 1RM.* This increase occurred in all four trunk muscles (RA, EO, LSES and ULES) in the eccentric phase and all muscles (EO, LSES and ULES) apart from RA in concentric phase.

In the fourth study, *we demonstrated that despite significantly higher external loads in the hack squat, trunk muscle activation was greater in free barbell back squat at equivalent relative loads.* We determined that the back squat kinematic characteristics explaining these differences were*; greater range of movement, faster descent and the requirement to control the unsupported external load through the full movement, engaging the entire kinetic chain.*

The most important and novel finding in the final study was that *trunk muscle activation was lower in participants who were stronger in the squat compared to weaker subjects.* This was found for all three loads (65, 75 and 95% SM) in the eccentric phase and only the heaviest load (95% SM) in concentric phase. The second novel finding was that this *lower EMG response in the stronger group translated to all three phases of the SJ and CMJ but was only significant in concentric and flight phase.* This reduced activation in the strong group was also associated with significantly higher jump performances compared to the weak group. *In summary, adaptation to progressive squat training results in more efficient trunk stability in squat strength tasks and dynamic, bodyweight jump performance.* The final study demonstrated that *trunk muscle activation increased significantly in each 30^o segment of descent and reached the highest level in the first 30^o segment of ascent for all test loads. Hence, highest trunk muscle activation occurred in the two lowest 30^o segments of the parallel squat movement.*

Research design

In the 3 neuromuscular studies we used cross-sectional design to observe a single group (Study 3 and 4) and compare different groups, strong, middle and weak (Study 5), at a single time point. This is an observational approach where we did not intervene and influence participant status. As a result, the study 5 findings require ratification in a training study in order to facilitate further effective translation to applied practice. Specifically, a well structured randomized controlled trial (RCT) measuring effects of increased squat strength on trunk stability while performing dynamic athletic actions (Hecksteden *et al.*, 2018). The study should be conducted on a representative and large enough cohort of participants to ensure appropriate statistical power (Hopkins, Schabort and Hawley, 2001). Current applied training guidelines for development of trunk stability in healthy and athletic participants lack scientific foundation. Our findings partly fill this void by addressing the fact; loaded compound exercises are an effective method of developing dynamic trunk stability. Further investigation and clarification through effective RCT's are required to challenge and replace the many un-scientific core stability training practices.

Reliability and comparison studies on a small number of participants using multiple comparisons have been questioned. In study 3 & 4, we had a relatively small number of participants and analysed a fairly large number of data points, including kinematic and RMS measures. Hopkins et al (2001), in a review suggested that reliability of tests relating to physical performance were acceptable in a lower number cohort where the group is homogenous in the key area of competence (Hopkins, Schabort and Hawley, 2001). In these two studies, the 10 participants were competent in the barbell back squat evidenced by a mean relative 1RM of 165% body mass. Given that barbell back squat was the exercise being studied, we concluded that our cohort size represented an acceptable number for the analyses that we performed. Nevertheless, we performed sample size calculation at 90% power using G*Power effect size difference (F test ANOVA RM) from RMS (ULES) from 75-95% SM from back squat reported in study 4. This confirmed that a minimum sample size of 8 is required when performing 3 repeated measurements.

In study 5 we had a total group of 50 participants, who demonstrated normal distribution for relative BS 1RM according to D'Agostino & Pearson normality test (alpha=*0.05*). This resulted in two groups of 17 (weak and middle groups) and one of 16 (strong group). Using the same effect size RMS as above but applied in a different G*Power model (F test ANOVA RM) using 3 independent groups at 90% power, we calculated that a minimum of 12 participants per group were required. Given we used substantially more than 12 in each group we are therefore confident that type I and type II errors were avoided.

Electromyography

The use of surface EMG to measure muscle activity has been used for over two centuries and the refinement of procedures have improved accuracy and reliability. Electrode placement for effective measurement of trunk muscle EMG has been researched resulting in published guidelines (Hermens *et al.*, 2000; Huebner *et al.*, 2014). Normalization of trunk stabilizer EMG data captured in dynamic exercise has been extensively researched and discussed at length earlier in this thesis (Introduction and Introduction to Paper 3). The neuromuscular studies began with an investigation that demonstrated acceptable reliability and sensitivity for our method of analysis of trunk muscle activation in the back squat. Specifically, the research studies 4 and 5 were designed to ensure neuromuscular data was captured in a single test session, thereby avoiding re-application of electrodes and associated loss of accuracy. While there are recorded limitations associated with surface EMG assessment of muscle function and specifically in trunk stabilizers, it remains the most effective method of measuring this in loaded dynamic exercise. The accuracy of our finding, that increases in squat load produced greater trunk muscle activation in all muscles tested, improved over the 3 neuromuscular studies. This strongly suggests that this is a valid and reliable finding. This consolidates previous studies which reported trunk muscle load effect for the squat using diverse methodology as reported in our review (Clark, Lambert and Hunter, 2012).

Surface EMG measurement of rectus abdominus activation response to load increases in the dynamic back squat has proven inconsistent (Hamlyn, Behm and Young, 2007).

Huebner et al (2014) investigated surface EMG amplitude in 5 trunk muscles for 4 different electrode placement positions and found RA EMG amplitude varied the most, regardless of electrode position (Huebner *et al.*, 2014). We argued (study 3) that the variance in RA sEMG measurement may result from folds in the skin adjacent to RA during deep flexion of the hip. In study 3 we reported unacceptable absolute reliability (CV%) but fair relative reliability (ICC) for RA RMS and found a tendency for load effect in this muscle (Clark, Lambert and Hunter, 2016). In the second neuromuscular study we found a significant load effect for RA for all loads in both phases of the back and hack squat (Clark, Lambert and Hunter, 2017). We repeated this finding in the final study (n-50) where RMS in RA increased by 18 and 45% in the eccentric and concentric phase respectively, in response to 20% increase in external load. It appears that through improved and consistent electrode placement, better management of clothing and artefact, we were able to capture RA sEMG more accurately as we progressed through the 3 neuromuscular studies. Importantly this confirmed that RA activation is sensitive to increases in squat load.

EMG measurement on one side of the body is an established method for bilateral standing and jumping exercises performed in sagittal where the load is carried in the midline (Seroussi and Pope, 1987; Sihvonen, Partanen and Hanninen, 1988; Vakos *et al.*, 1994). Bilateral symmetry of EMG signal has been demonstrated in this category of movement, which means that EMG captured on the right-hand side accurately reflects bilateral trunk muscle activation in this category of exercise.

Kinematics

In the three neuromuscular studies, we used a linear transducer (Celesco, PT5A, California, USA) to measure barbell displacement and in study 5, jump kinematics. This is a highly effective method for analysis of squat kinematics where participants remain on the ground, with flat feet throughout the movement. Furthermore, there is evidence that analysis of jumps by linear transducer is valid and reliable in comparison to force platform tests (Harris *et al.*, 2010; Hansen, Cronin and Newton, 2011). Linear transducer data can determine the start and end of the squat and separate eccentric and concentric phases accurately. However, when identifying phases of jumps where participants leave the ground in the flight phase, this technology has limitations. The end of the concentric phase determined by linear transducer is the point where concentric displacement matches the start position or upright standing. Therefore, flight phase starts once displacement proceeds beyond this point. The start of the flight phase in kinematic analysis of jumps using a force plate would be at ‘toe off’ after full plantar flexion (Linthorne, 2001). This means that in study 5, the segment from flat foot upright standing to toe off is included in flight phase, despite still being in contact with the ground. What has not been reported, in the authors view, is relative contribution of this plantar flexion segment of the concentric phase to force application and jump performance.

In study 5 we found that trunk muscle activation was significantly higher in concentric phase compared to eccentric phase and importantly remained at the same level during flight phase. The question therefore arises, does the ground contact portion (plantar flexion) included in the flight phase influence this activation? Farris at al (2016), in an *in vivo* analysis of plantar flexor muscle-tendon interaction during vertical jumping, concluded that this muscle group makes their greatest contribution in early to middle portion of the concentric phase (Farris *et al.*, 2016). The authors suggest this is primarily to stabilize and facilitate power generated by knee extensors. They concluded there was no evidence that stored elastic energy in the plantar flexors contributed to force production in the final stage of concentric phase prior to toe off. Our interpretation therefore, is that high muscle activation in the flight phase is to stabilize the trunk for effective performance and in preparation for landing. This is supported by our finding that in the strong squat group activation in both concentric, and specifically the flight phase, was significantly lower than in the weak group.

In studies 3 and 4 we found that eccentric displacement decreased significantly with each 10% increase in load, while concentric displacement remained unchanged for all loads. The absence of knee and hip angle measurements in these two studies meant that we were not able to explain this with any certainty. We presented two possible explanations; increases in load resulted in subconscious proprioceptive inhibition preventing full knee and or hip flexion to avoid the final, most challenging segment of the parallel squat. We also suggested spinal shrinkage due to incremental compressive forces might be a factor. Wisleder et al (2001) found spine shrinkage of -3.9 ± 1.2 mm resulting from spinal bending, rotation and pure compression in response to a load equal to body mass (Wisleder *et al.*, 2001). We found a mean reduction in displacement of 56 mm (Range 22-86 mm) for loads equivalent to 77, 100, 127 and 150% body mass. Even at loads greater than body mass, it is unlikely that spine shrinkage would account in full for this reduction in eccentric displacement. Hence, we propose that reduced displacement is the result of lower squat start position due to spine compression and protective inhibition preventing full depth squats at heavier loads. Our explanation for the absence of a reduction in concentric displacement in studies 3 and 4 remains applicable. The instruction to complete this phase as explosively as possible means that velocity increases as mechanical load is overcome, and peaks in the final stage of hip and knee extension. This means that the final height of displacement exceeds the (compressed) start point, which is clearly visible when observing moderate to heavy load squats performed explosively.

Impact: Applied strength and conditioning

The back squat is an established method of developing lower limb for performance in sports where strength, power, acceleration and speed are important (Seitz *et al.*, 2014). There is also a growing appreciation of the role of trunk stability in effective function of the kinetic chain in dynamic athletic performance. Specifically, the role of the trunk anatomical region in resisting, coordinating, transferring and optimizing forces, torques and power generated by the limbs within the full, integrated kinetic chain. However, there is little scientific evidence on how this should be trained in a healthy and athletic population.

Trunk stability for sports performance is integrated within the full kinetic chain and therefore must tolerate force, torque and velocities characteristic of such movement. Logically therefore, a training intervention to develop trunk stability must adhere to two key training principles, specificity and overload. To be specific, exercises should be dynamic and similar to the movement for which trunk stability is required. Overload is achieved by executing selected movements under conditions of greater force, torque or velocity of movement.

The findings of this suite of studies confirm that loaded free barbell squat is effective in activating the trunk stabilizers. Importantly, this stimulus is integrated within a compound, whole body movement, which addresses the first limitation of traditional CST, specificity. Our research has confirmed that loaded back squats represent a training overload for the trunk stabilizers, thereby addressing the second limitation of traditional CST. While there has been progress in scientific challenge to traditional core stability, research has failed to give clear direction on the most effective training methods to develop dynamic trunk stability for athletic performance. Our research provides that direction with scientific evidence;

- *Trunk stabilizers are sensitive to typical training load manipulation in the back squat.*
- *Trunk muscle activation is highest in the final 30^o of squat descent to parallel and first 30^o of ascent regardless of load.*
- *Adaptations to loaded back squat training improve efficacy of trunk stability mechanisms under load.*
- *Trunk muscle adaptations to loaded squat training transfer to powerful, dynamic bodyweight jumps, characteristic of many sports.*

The loaded parallel back squat can be viewed with confidence as an effective method of activating and training the trunk stabilizers. Squat training with loads ranging from 65 to 95% of 1 repetition maximum will activate trunk stabilizers effectively. This will increase dynamic trunk stability required to withstand increasing loads in the squat and over time will do this more efficiently, at lower levels of trunk muscle activation. Furthermore, training to parallel depth or lower will optimize activation during training and develop stability in the deepest, most challenging phase of squat descent and ascent.

The development of squat strength through continuous, progressive training will transfer to SJ and CMJ performance (Wirth, Keiner, *et al.*, 2016). We showed that squat strength was also associated with lower trunk muscle activation in concentric and flight phase of SJ and CMJ. Therefore, squat training adaptation results in greater jump performance at lower levels of activation of the trunk stabilizers. Arguably, this adaptation to squat training will transfer to a range of dynamic athletic actions.

Recommendations for strength training programmes for development of strength, power and performance in athletic activities:

- Include the loaded free barbell back squat at progressive loads ranging from 65 to 95% 1RM for the development of trunk stability.
- Ensure correct and safe technique, specifically squatting to a minimum of parallel depth.

Use the following criteria to monitor progress in the development of trunk stability:

- In *compound loaded exercises*, observe and monitor the ability to *adhere to correct technique under load and progression of load*. Specifically, the capacity to manage load through the trunk; *evidenced by monitoring the ability to keep the path of the load over the base of support through the full range of movement.*
- In *technical sporting movements*, observe and monitor the ability to *execute technical movements at high force, velocity and torque effectively and correctly in situ*.
- In both of the above contexts, *monitor and assess acute change within a training session in response to fatigue to assess short-term endurance*.
- In both of the above contexts, *monitor and record development across a longitudinal training period to assess progress and chronic adaptation.*

Impact: Future research

Reviews on the application of CST for sports performance and proxies thereof report a number of commons research flaws (Reed *et al.*, 2012; Silfies *et al.*, 2015; Prieske, Muehlbauer and Granacher, 2016; Wirth, Hartmann, *et al.*, 2016), which should guide and inform future research. There is agreement that absence of a valid test of core stability has undermined progress in scientific understanding of core stability training effects (Hibbs *et al.*, 2008; Prieske, Muehlbauer and Granacher, 2016). Isolating and measuring trunk muscle strength as a component of trunk stability is in itself flawed (Okada, Huxel and Nesser, 2010). Furthermore, current reliance on tests that resemble common exercises used in CST means that results are biased in favour of CST. Complex neuromuscular interactions that underpin dynamic trunk stability cannot be described by a single measure of strength, especially strength measured by an isometric test. Current scientific methodology is not capable of isolating and measuring these factors within athletic activity to inform training manipulations. Nor is methodology available to measure acute exercise response or chronic training adaptations in selected trunk stability exercise interventions. *Which means that testing proxies of sports performance is currently the most effective method of assessing transfer of trunk stability training interventions.*

Our novel findings that improved trunk stability developed through loaded barbell squat training make a significant contribution to:

1. Increased load carrying capacity in loaded barbell squat by enhancing stability at more efficient levels of trunk muscle activation.

2. Increased body weight jump performance by increasing trunk stability at more efficient levels of trunk activation.

The importance of these findings for applied strength and conditioning practice have been described above. With reference to the first point, there would be value in better understanding how other commonly used loaded compound exercises, engage and develop the trunk stabilizers. Perhaps more meaningful and interesting, relating to point 2, is how do the adaptations to loaded squat training translate to trunk stability in other dynamic athletic activities? Research recommendations:

- Subject the *key findings of this research to further investigation in a randomized controlled training study.*
- Investigate *acute response and adaptation* of trunk muscles to exercise and training with *compound loaded exercises, including deadlift and Olympic weightlifting exercises.*
- Based on the principle that squat training impacts positively on stabilizer adaptation, *determine the impact of progressive squat training on neck strength as a protection against concussion*.
- Impact of *enhanced trunk stability from progressive squat training on dynamic trunk stability in performance of common sporting activities* such as; sprinting, cycling, reactive agility in racquet sports, canoe paddling and rowing.

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Appendices

Core Stability (copy)

Page 1: Page 1

There have been a number of recent reviews highlighting the lack of consensus around the topic of core stability. There is however agreement that core stability is important in everyday life and dynamic sporting activity. The confusion lies in the most effective manner of developing and measuring core stability.

This motivated me to embark on a PhD a few years ago looking at neuromuscular function of the trunk in the loaded squat in the hope of shedding some light on this approach for developing core stability.

An obvious related question is: How do people working in sport view core stability and its development for dynamic athletic performance?

I appreciate your time in completing this short survey (<15 min) which will focus on ***core stability for dynamic athletic performance***.

Page 2

Section 1: Demographics

1. What is your ***Primary*** discipline? ✱ *Required*

○ Sports Medicine Practitioner
○ Sports Physiotherapist
○ Masseur / Soft Tissue Therapist
○ Strength and Conditioning Coach
○ Sports Physiologist
○ Sports Psychologist
○ Performance Nutritionist
○ Biomechanist
○ Performance Analyst
○ Sports Coach
○ Athlete / Player
○ Sports management
○ Other

1.a. If you selected Other, please specify:

☐

2. What area of sport are you involved in? ✱ *Required*

○ Professional, full-time paid position working with full-time paid atheletes
○ Semi-professional, paid part-time position
○ Elite professional, full-time paid position working with funded and amateur athletes

(Institute)

○ Elite non-professional, part-time working with regional or national selected athletes
○ Volunteer in recreational club sport
○ Academic, university or school sport role
○ Other

2.a. If you selected Other, please specify:

☐

3. Please indicate below which describes most accurately where you do most of your work. ✱ *Required*

○ Team sport
○ Individual athletes
○ Combination of team and individual athletes

4. What is your highest academic qualification? ✱ *Required*

○ PhD
○ MSc or Masters
○ Degree or Honours degree
○ Diploma
○ Other

4.a. If you selected Other, please specify:

☐

4.b. Do you have a professional qualification linked to your discipline? ✱ *Required*

○ Yes
○ No

4.c. How many years have you been working in your current discipline? ✱ *Required*

Section 2: Core Stability

5. Which statement below **do you think** best describes what constitutes ***the core*** most accurately? ✱ *Required*

○ The spine and the associated muscles and nerves
○ The lumbar spine, pelvic and hip joints and associated muscles and nerves
○ The region between and including the pelvic and shoulder girdles and associated muscles and nerves
○ The region between and diaphragm and pelvic floor and associated muscles and nerves
○ Other

5.a. If you selected Other, please specify:

6. What term do you believe best describes the anatomical region that this survey is dealing with? ✱ *Required*

○ Torso

○ Trunk

○ Core

○ Upper limb

○ Other

6.a. If you selected Other, please specify:

☐

7. Please rate how strongly you agree or disagree with the following statements.

	✱ *Required*				
	1 Strongly agree	2 Agree	3 Neither agree nor disagree	4 Disagree	5 Strongly disagree
Core strength is required for core stability	○	○	○	○	○
Core strength and core stability are separate attributes	○	○	○	○	○
Core strength is required for dynamic athletic performance but not everyday life	○	○	○	○	○
Core stability is dependent on neural timing and muscular coordination rather than core strength	○	○	○	○	○

8. Do you believe that trunk muscle activation measured by ***surface electromyography*** is ***reflective of performance*** of the core stabilization system? ✱ *Required*

○ Yes

○ No

○ Don't know

9. Please rate the following ***categories of exercise*** on their ***effectiveness*** in developing core stability ***for dynamic athletic performance***?

	✱ *Required*				
	1 Very effective	2	3	4	5 Not effective at all
Isolated abdominal bracing	○	○	○	○	○
Isometric held exercises such the plank	○	○	○	○	○
Dynamic abdominal exercises such as sit-ups	○	○	○	○	○
Dynamic inverted exercises such as hanging leg raise	○	○	○	○	○
Suspended compound exercises using systems such as the TRX	○	○	○	○	○
Instability abdominal exercises performed on a Swiss ball	○	○	○	○	○
Functional exercises such as farmers walk	○	○	○	○	○
Loaded free barbell exercises such as Squats and Olympic lifts	○	○	○	○	○

10. Please rate how strongly you agree or disagree with the following statements as they relate to determining ***exercise selection*** for the development of core stability for ***dynamic athletic performance***.

✱ *Required*

	1 Strongly agree	2 Agree	3 Neither agree nor disagree	4 Disagree	5 Strongly Disagree
The exercise must subject the athlete to forces equal to or greater than expected in the sport or event	○	○	○	○	○
The exercise must emphasize correct movement pattern above all else	○	○	○	○	○
The exercise must subject the athlete to velocity of movement equal to or greater than expected in the sport or event.	○	○	○	○	○
The exercise must develop capacity for sustained isometric contraction	○	○	○	○	○

11. Please rate how strongly you agree or diagree with the following statements as they relate specifically to ***ground based loaded free barbell exercises*** (Squats and Olympic weightlifting exercises).

	✱ *Required*				
	1 Strongly agree	2 Agree	3 Neither agree nor disagree	4 Disagree	5 Strongly disagree
Trunk muscle activation will increase with increases in velocity of movement.	○	○	○	○	○
Trunk muscle activation is dependent on correct postural control	○	○	○	○	○

Trunk muscle activation is enhanced by slow controlled movement	○	○	○	○	○
Trunk muscle activation will increase with increases in external load	○	○	○	○	○

12. What is the most effective method of measuring core stability in a healthy, un-injured person? ✱ *Required*

☐

13. Do you think it is necessary to include specific exercises to train core stability in a healthy, uninjured ***non athlete's*** exercise programme? ✱ *Required*

○ Yes
○ No
○ Don't know

14. Do you think it is necessary to include specific exercises to train core stability in a healthy, uninjured ***athlete's*** exercise programme? ✱ *Required*

○ Yes
○ No
○ Don't know

15. Do you think that the development of core stability can ***prevent back pain***? ✱ *Required*

○ Yes
○ No
○ Don't know

16. Do you think that certain ***lower limb overuse injuries*** are caused by ***poor or under developed*** core stability? ✱ *Required*

○ Yes
○ No
○ Don't know

17. Do you think it is ***possible*** to isolate and train the core stabilization system? ✱ *Required*

○ Yes
○ No
○ Don't know

18. Do you think it is ***effective*** to isolate and train the core stabilization system? ✱ *Required*

○ Yes
○ No
○ Don't know

19. Do you think that the core stability is ***automaticall***y developed during normal, ***progressive*** exercise training? ✱ *Required*

○ Yes
○ No
○ Don't know

Page 3

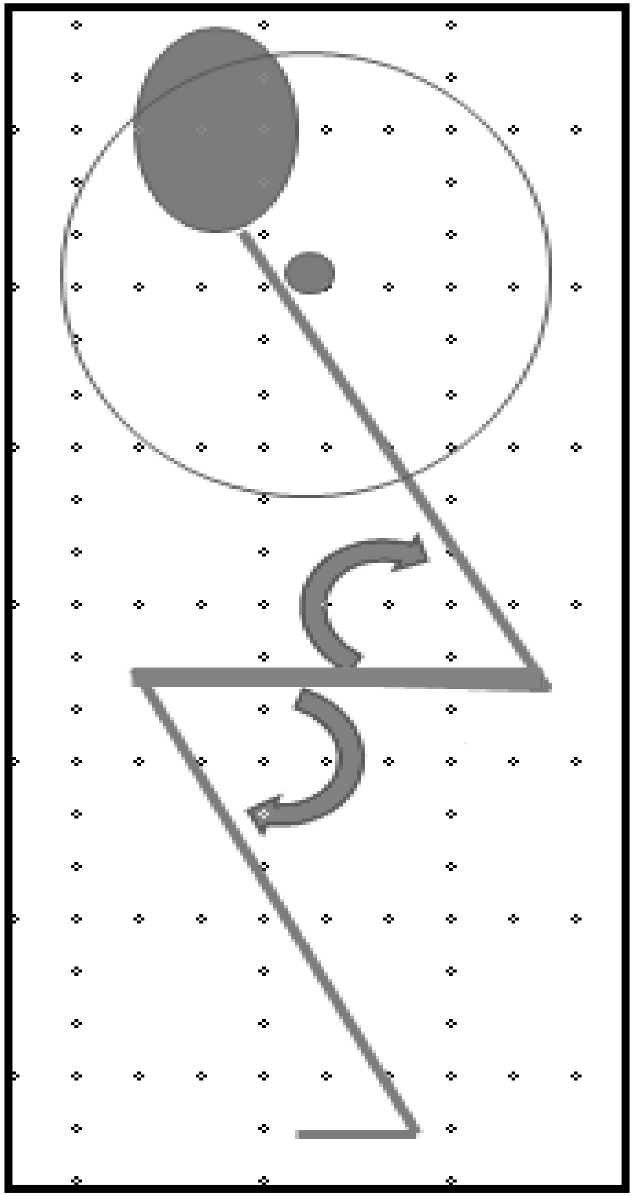
Thank you for completing the survey.

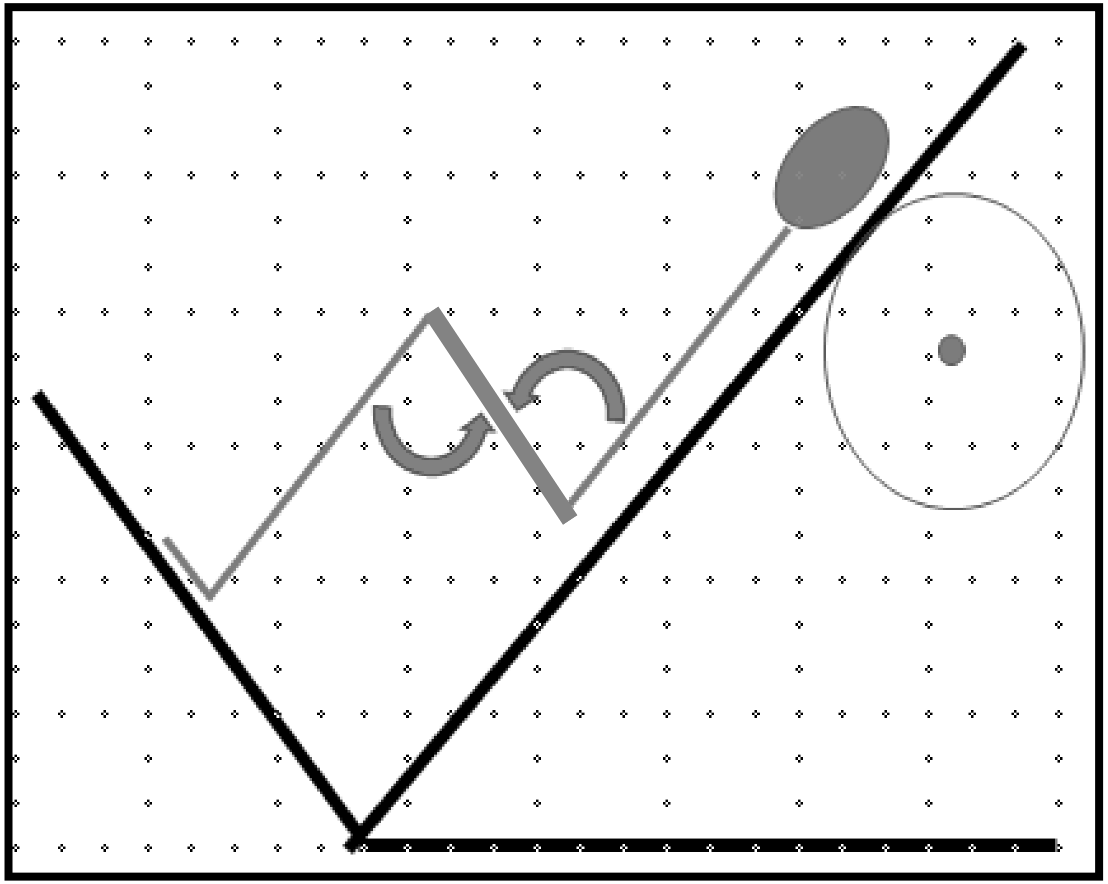
Appendix 2

Back squat and hack squat set-up

Appendix 2

Back squat and hack schematic





Muscle Sites:

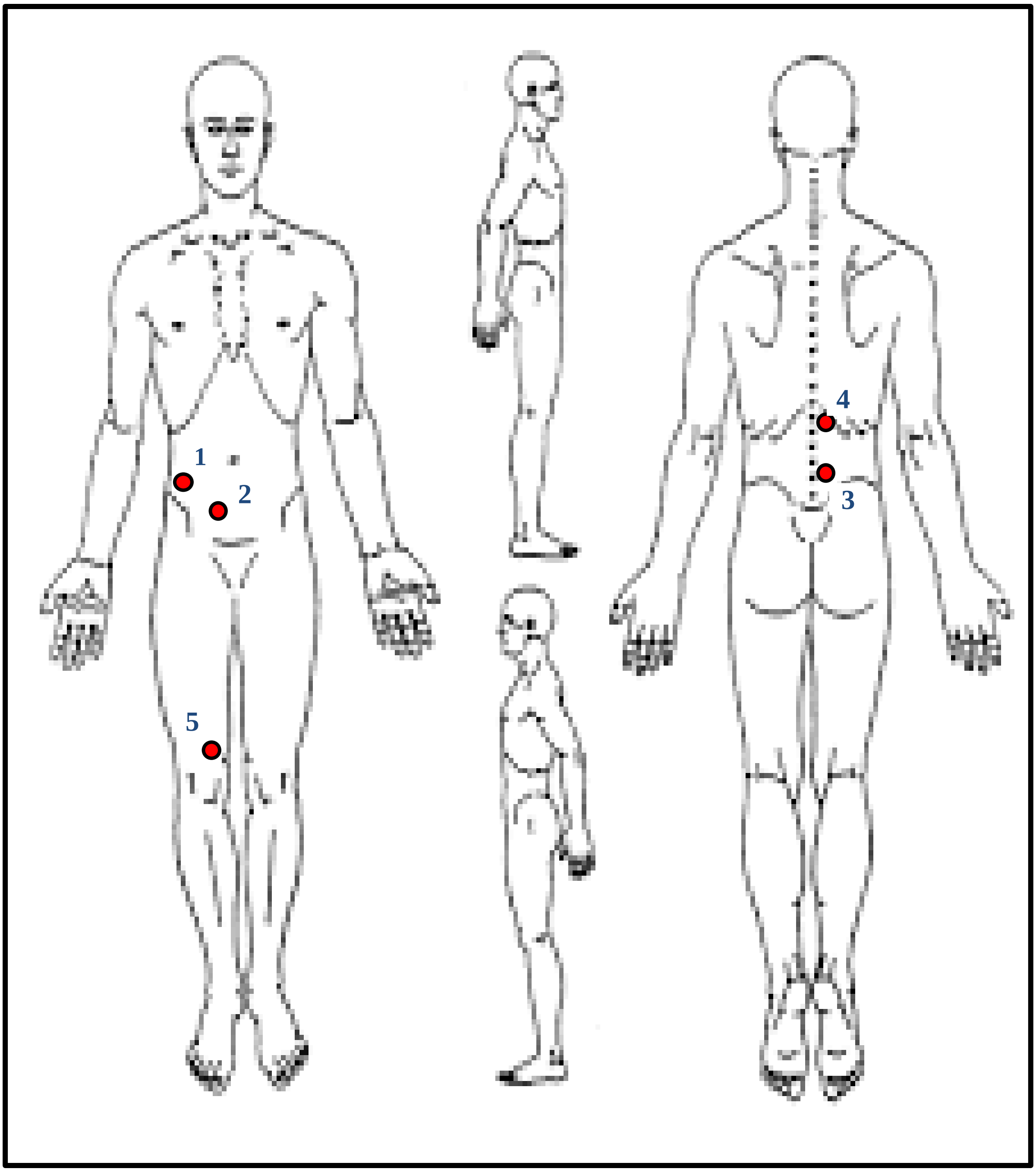
1 – Exernal oblique (EO)

2 – Rectus abdominus (RA)

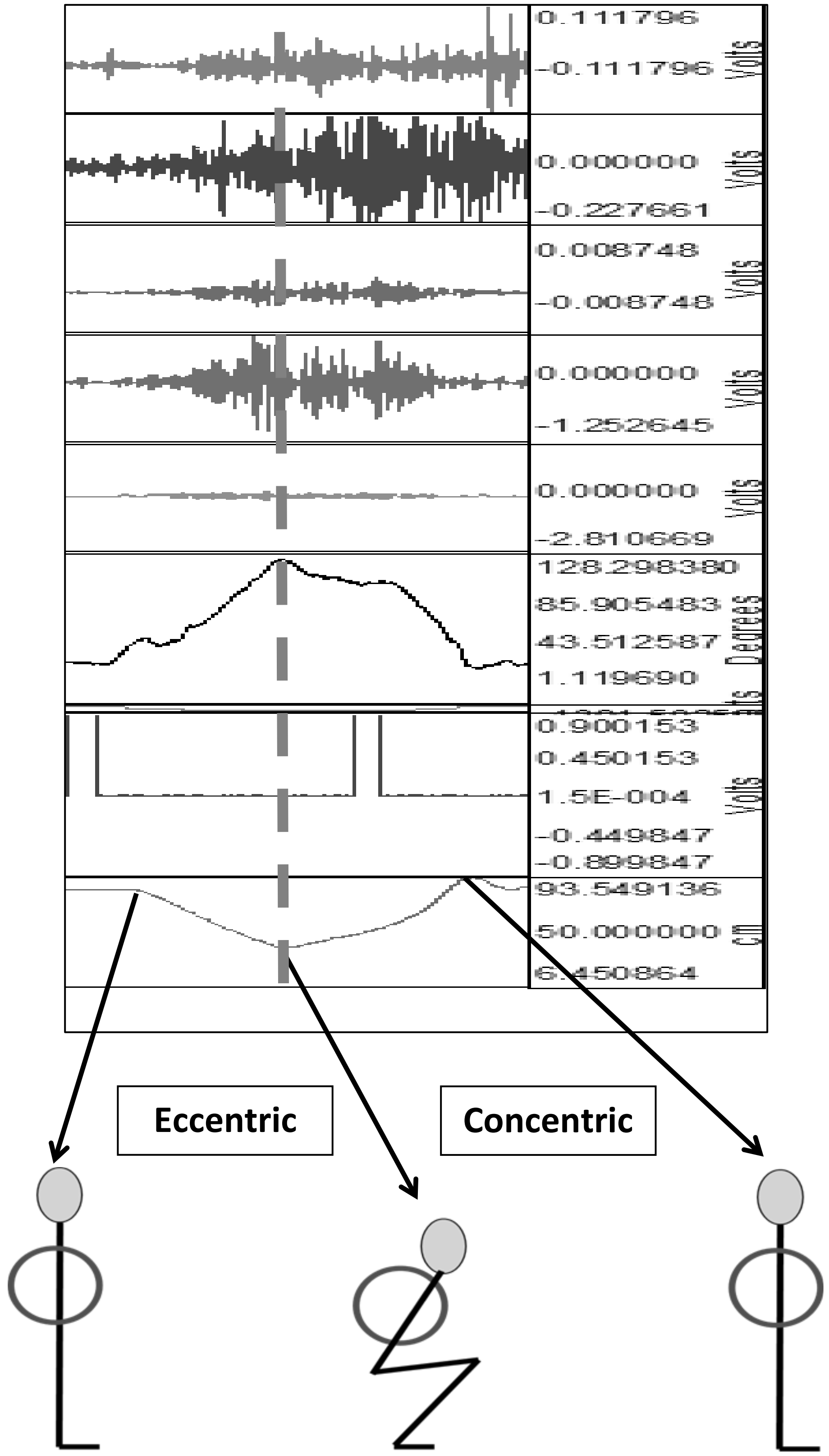
3 – Lumbar sacral erector spinae (LSES)

4 – Upper lumbar acral erector spinae (ULES)

5 – Vastus lateralis (VL)



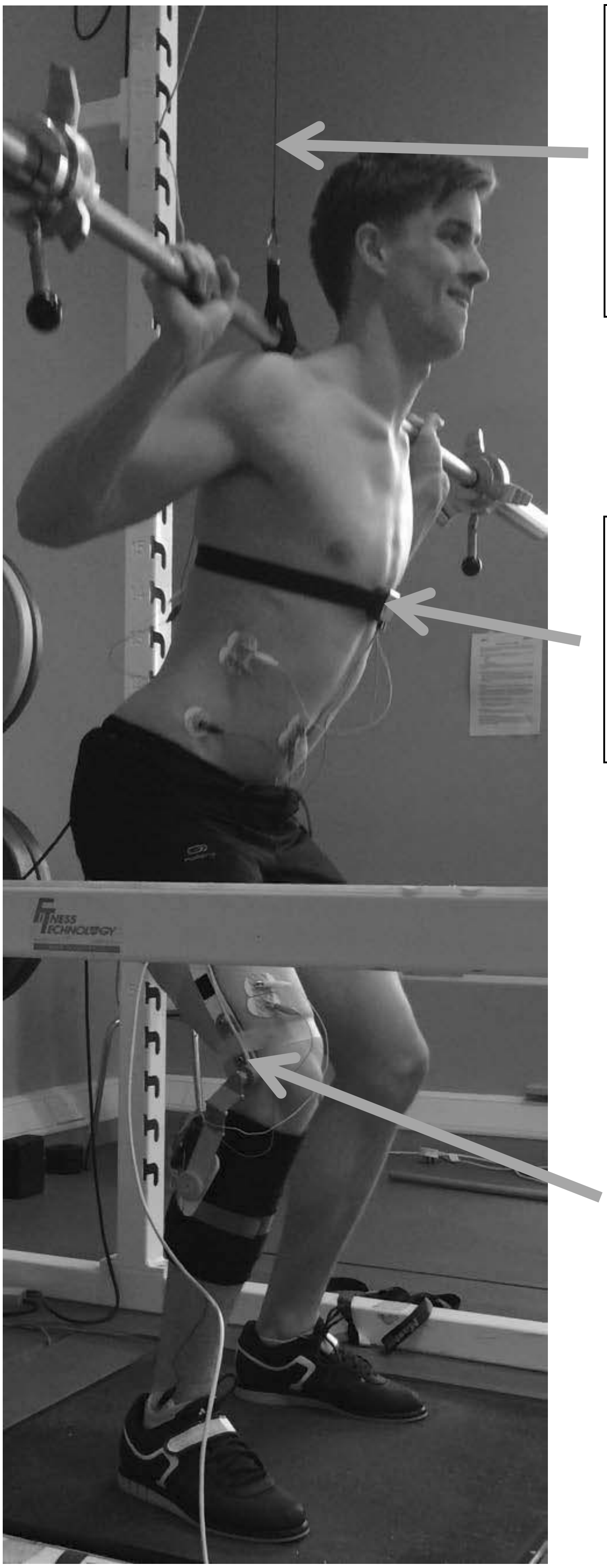
Appendix 4



Eccentric and concentric phase determined from displacement (cm) measured by linear encoder used to identify mean EMG for each phase.

Appendix 5

Kinematic set-up: Back squat



Linear encoder:

- Measures barbell displacement (cm)
- Determine eccentric and concentric phases of the back and hack squat and jumps
- Determine flight time for jumps

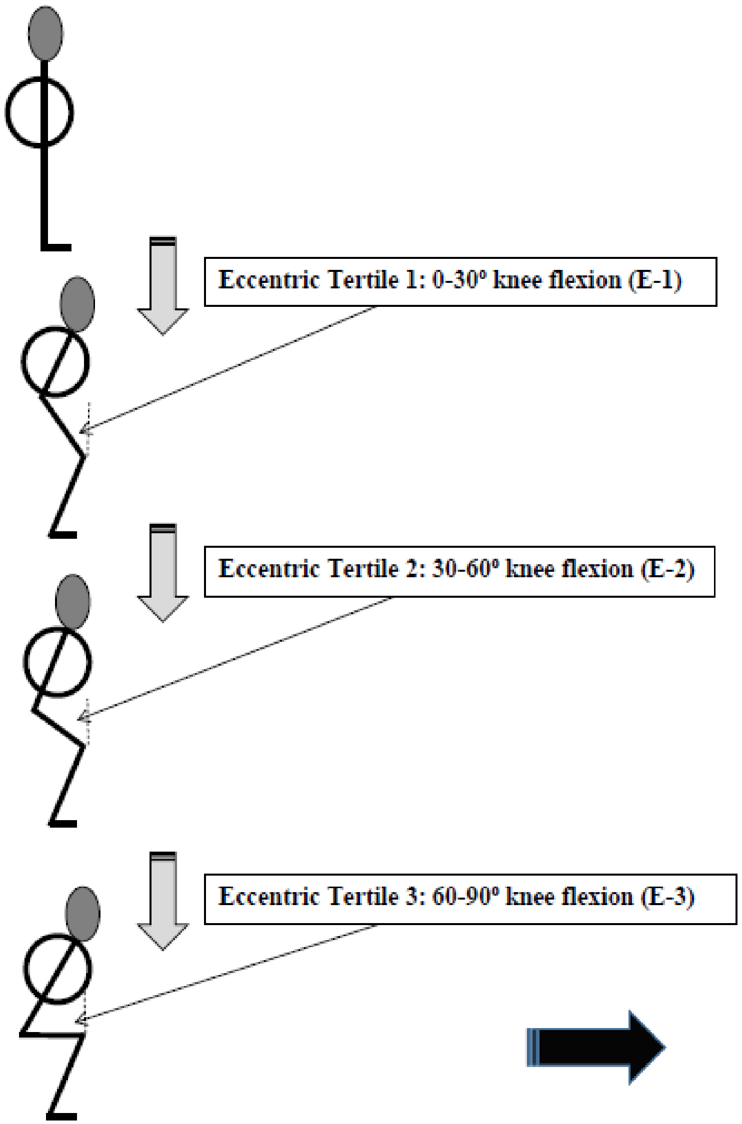
BioNomadix 2 Ch. EMG transmitter:

- Each transmits high resolution sEMG signal at a rate of 2000Hz from two muscle sites to the Biopac MP150 receiver unit

Electromechanical goniometer:

- Measure knee flexion in eccentric phase and extension in concentric phase.
- Signal used to determine three 30^{o} tertiles in each phase; eccentric and concentric.

Eccentric Phase Divided into Tertiles (30°)



Concentric Phase Divided into Tertiles (30°)