CLARK, D., LAMBERT, M.I. and HUNTER, A.M. 2016. Reliability of trunk muscle electromyography in the loaded back squat exercise. *International journal of sports medicine* [online], 37(6), pages 448-456. Available from: https://doi.org/10.1055/s-0035-1569366

Reliability of trunk muscle electromyography in the loaded back squat exercise.

CLARK, D., LAMBERT, M.I. and HUNTER, A.M.

2016

Reliability of trunk muscle electromyography in the loaded back squat exercise

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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.

Keywords: Back squat; Neuromuscular; Trunk muscles; Electromyography; Electrogoniometry

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 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.

Method

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 (See **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, 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.

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

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 $k = (\text{number of observations} - \text{number of tests})/(\text{number of participants} - 1)$

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

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 ($F_{(3,27)} = 3.06$, $p < 0.05$) (See **Fig. 1b**). There was also a significant increase in time as a result of increases in load for the eccentric phase ($F_{(3,27)} = 7.49$, $p < 0.05$) (See **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 (See **Fig. 1c**). Eccentric power increased significantly ($p < 0.05$) alongside load (See **Fig. 1d**). No significant differences or interactions were observed between test days for any of the eccentric kinematic variables (See **Table 2**).

Concentric displacement of the barbell was not different ($p > 0.05$) between test sessions or across loads (See **Table 2**). Mean concentric squat duration (See **Fig. 2a**) increased significantly ($F(3,27) = 34.21$, $p < 0.001$) with each test load increment (See **Fig. 2b**), whereas velocity decreased significantly ($F(3,27) = 45.68$, $p < 0.001$) (See **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 (See **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 (See **Table 2**).

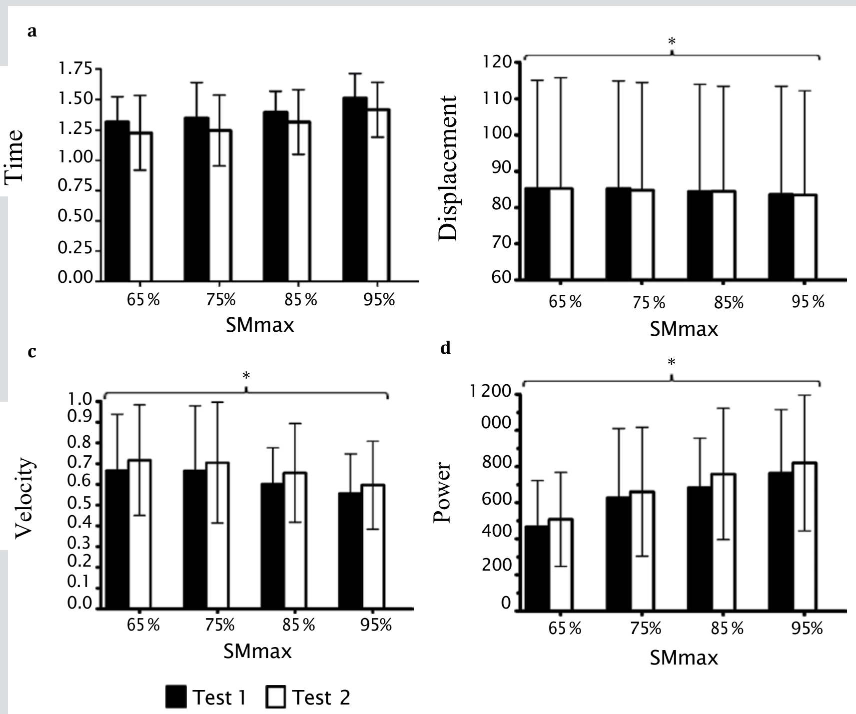

Fig. 1 Eccentric kinematic data for test 1 and 2 for all test loads: Time **a**, Displacement **b**, Velocity **c** and Power **d**. * Denotes significant load effect ($p < 0.05$).

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

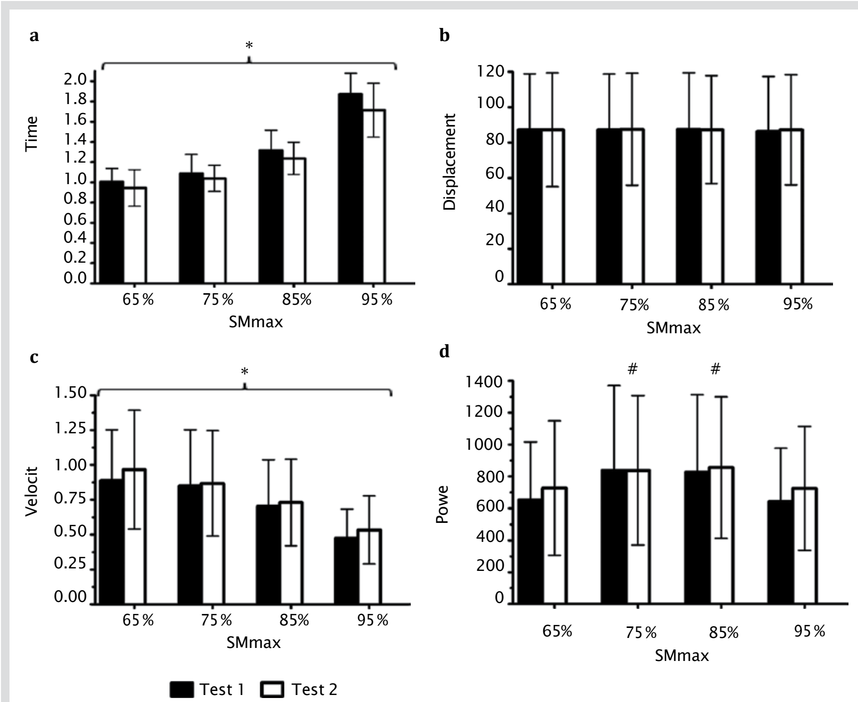

Fig. 2 Concentric kinematic data for test 1 and 2 for all test loads: Time **a**, Displacement **b**, Velocity **c** and Power **d**. * Denotes significant load effect ($p < 0.05$). # Denotes significant difference from data at 65 and 95 % SMmax ($p < 0.01$).

Reliability/repeatability

Kinematic mean differences and intra-participant CV % for test 1 to test 2 for each test load is presented on See **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 (See **Fig. 3**). Mean RMS increased significantly ($p < 0.05$) for all muscle sites by load in the eccentric phase (See **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 (See **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 % (See **Table 3**). Test-to-retest intra-participant CV % for all RMS data ranged from 12.6 ± 7.2 % to 29.4 ± 1.12 % (See **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 See **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.

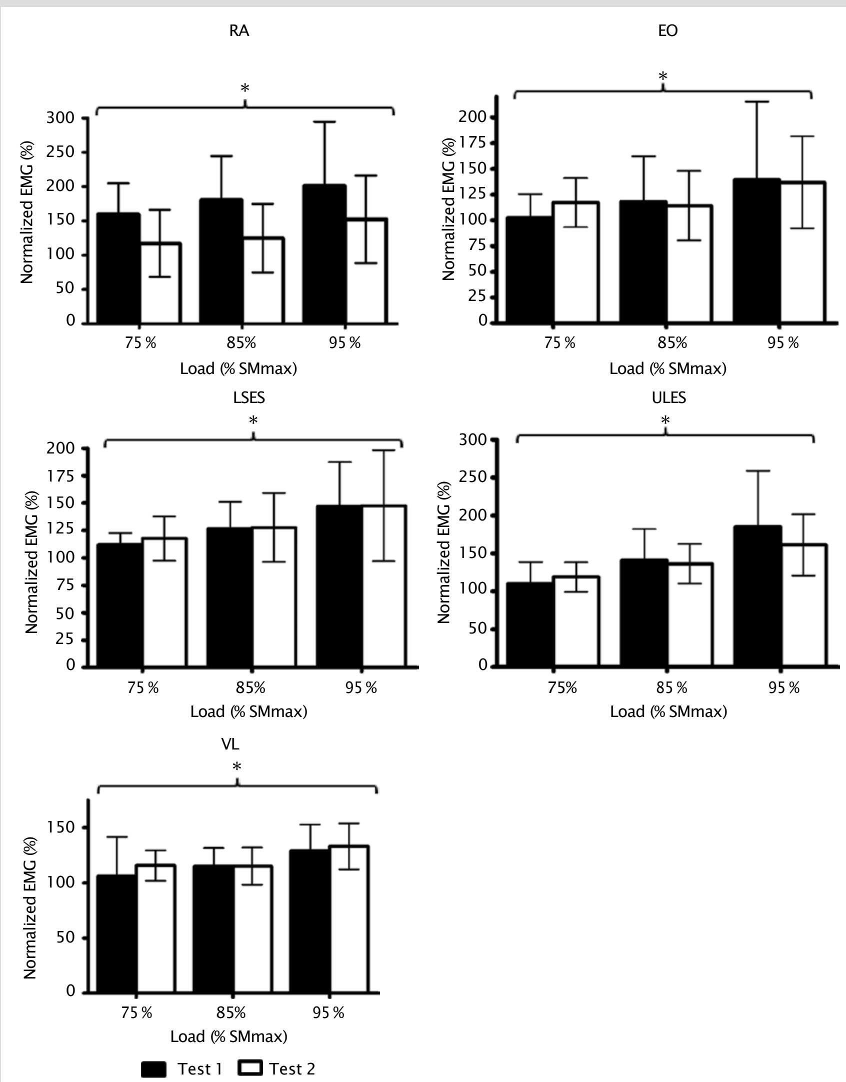

Fig. 3 Eccentric mean RMS for 3 test loads Normalized to 65% SMmax for 5 muscle sites: RA, EO, LSES, ULES and VL. * Denotes significant load effect $p < 0.001$.

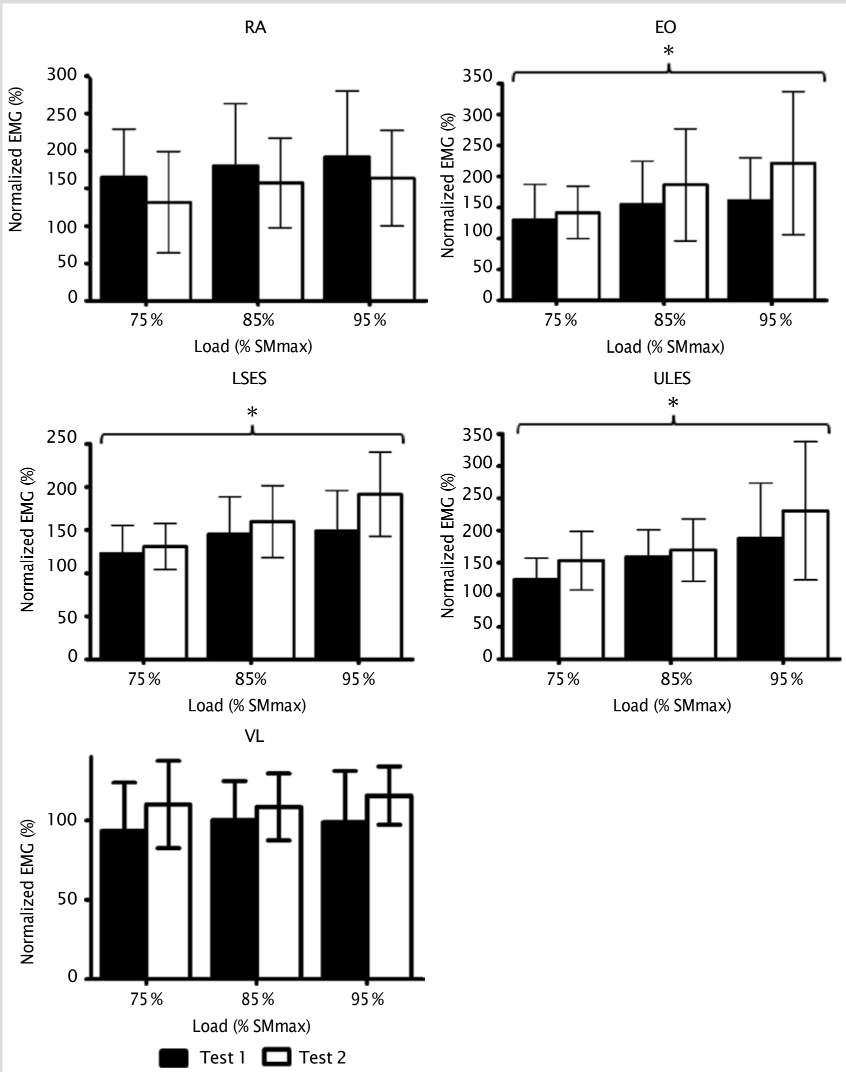

Fig. 4 Concentric mean RMS for 3 test loads normalized to 65 % SMmax for the 5 muscle sites: RA, EO, LSES, ULES and VL. * Denotes significant load effect $p < 0.001$.

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

Muscle site	Test load	Eccentric Interclass Correlation Mean (LCI–UCI) *	Inter-subject CV % Mean ± SD	Concentric 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

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.

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, ex pert 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 (See **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 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 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 (See **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 (See **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 time-frame. 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.

Acknowledgements

The authors wish thank Dr Tom Balshaw and Dr Michael Newton for their technical assistance in testing and Matlab data analysis respectively. Also Frank Kelly for his general technical assistance.

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