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ORIGINAL RESEARCH

STATIC LOADS ON THE LOWER BACK FOR TWO MODALITIES OF THE ISOMETRIC SMITH SQUAT

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ABSTRACT

Introduction: The squat is one of the most effective exercises in athletic training. However, there is a scarcity of research that reports the muscular and joint loads in the lumbar region incurred when performing the high bar and the low bar isometric squat modalities in a Smith machine. Therefore, this study aims to determine the muscle force of the lower back extensors, and the compressive ($\mathbf{R}_c$) and shear ($\mathbf{R}_s$) forces at the lumbo-sacral joint for the one repetition maximum (1RM) high bar and low bar isometric parallel-depth Smith squats.

Methods: Eight healthy male well-trained 400-m sprinters participated in the study. The athletes performed the two modalities of the isometric squat on a 7° backward-inclined Smith machine using a mean ± SD 1RM external resistance of 100.3 ± 7.2 kg. During the squat, the participants paused for 2-3 s at the bottom of the squat, corresponding to a position in which the thighs are parallel to the ground. This was, therefore, considered a static position for the calculation of isometric muscle forces and joint loads using static mechanical analysis. Moment arms, and joint and segmental angles were calculated from video images of the squatting performance. Internal forces were computed using a geometrical model of the trunk and lower limb.

Results: Spinal extensor muscular forces and lumbo-sacral joint forces were higher when using the low bar technique; with the exception of $\mathbf{R}_s$ which was approximately equal. The mean $\mathbf{R}_c$ were 10.2 body weights (BW) or 8,014 N (high bar) and 11.1 BW or 8,729 N (low bar).

Discussion: The low bar technique yields higher $\mathbf{R}_c$ and may therefore be avoided in the rehabilitation of spinal injuries. Increased bone mineral density and well-developed trunk musculature due to long term squat training can provide protection against passive spinal tissue failure. Therefore, the $\mathbf{R}_c$ found for the 1RM isometric parallel-depth Smith squat do not appear excessive for healthy well-trained athletes. The presence of $\mathbf{R}_s$ at the lumbo-sacral joint in both squat modalities suggests potential for damage to the intervertebral disc. The findings provide an in-depth understanding of the two squat modalities in isometric conditions for the prevention of lower back injury and the design of rehabilitation programs.

Keywords: Bone mineral density, intervertebral disc, lower back, lumbo-sacral joint, Smith squat, Statics.

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INTRODUCTION

The biomechanical and neuromuscular similarity to sportive movements, such as running and jumping, explains the widespread use of the squat exercise in strength training programs and rehabilitation interventions [1]. There are three fundamental variations of the Olympic barbell squat: front squat, high bar back squat, and low bar back squat [2]. In the high bar squat, the barbell is positioned at the base of the neck and above the posterior deltoid muscles. The low-bar technique involves holding the bar across the posterior deltoids and at the middle of the trapezius muscles [3]. Both the free squat and the Smith machine squat (Smith squat) can be conducted in diverse forms, including varied squat depths, foot posture and stance width [3]. The Smith squat is used by athletes complementary to the unconstrained free-barbell squat. By reducing balancing demands, the Smith squat allows athletes to use supramaximal loads in a partial range of motion, and therefore implement specialized, targeted training [3]. In addition, squat motion constrained to a single degree of freedom provides a more controlled, safer environment for resisted strength training [3]. The following is an overview of the loads on the vertebral structures imposed by the squat, and the mechanics, effectiveness and safety of the free squat and Smith squat. The merits of the isometric Smith squat and the high bar and low bar squat techniques are described previously [4].

Loads on the vertebral structures imposed by the squat

In the squat, compression of the vertebral bodies and intradiscal pressure augment linearly with increased external resistance [5]. Heavy resistances require greater torso stabilisation to counteract excessive spinal shear forces [6]. It has been suggested that deep squats may not pose a high risk of lumbar spine injury, as previously thought [3,6]. In particular, with increasing knee and hip flexion, tighter contact between tendons and skeletal structures (‘wrapping effect’) contributes to improved load distribution and better force transfer [3,6]. In fact, the influence of wrapping effects, functional adaptations of the passive tissue instigated by regular strength training practice, and soft tissue contact between the trunk and the upper surface of the thigh in deep squats can reduce the risk of injury. Thus, a lower risk of injury may be expected. Nonetheless, there is a dearth of research that quantifies the spinal loads incurred in the one repetition maximum (1RM) parallel-depth squat, in which the bottom of the squat corresponds to a position where the thighs are parallel with the ground.

MECHANICS OF THE FREE SQUAT AND SMITH SQUAT

Escamilla and co-workers [7] explained that in the free squat technique, the modulation of joint torques, muscle activity, and joint reaction forces is limited; since the line of gravity of the centre of mass of the athlete plus barbell must fall between the forefoot and heel. Such restriction imposed by the necessity to maintain balance limits the selective loading of different muscle and joint structures [3,7]. These limitations are overcome when the back is supported by a wall (wall squat) and when using the Smith squat where the exercise can be adapted to modulate the distribution of muscle activity and joint loading [7,8]. Positioning the feet forward in front of the knees and displacing the body and barbell towards the forefoot reduces knee torque and the compressive tibiofemoral and patellofemoral forces [3,8], however such actions emphasise hip and lumbosacral torques [3]. Previous literature has described the biomechanical characteristics of the squat movement [9,10,11]. However, there is a dearth of research that examines the lower back joint torques, and the spinal shear and compressive joint reaction forces that occur during the execution of the isometric parallel-depth Smith squat [3,6-8].

EFFECTIVENESS AND SAFETY OF THE FREE SQUAT AND SMITH SQUAT

In the contexts of athletic training and rehabilitation there is support for the use of the free barbell squat which may be considered more effective than the Smith squat [1-3]. The free squat is generally preferred by well-trained athletes because the requirement for balancing the barbell implicates large recruitment of trunk and lower limb musculature [12]. In agreement, Schwanbeck *et al.* [13] reported greater electromyographycal activity of the lower limb muscles when performing free squats compared to squatting in a Smith machine. There is, therefore, controversy regarding whether the Smith squat could be detrimental due to the unnatural path of the external load [1-3]. Nonetheless, several researchers have identified the Smith squat as a safe mode of strength training that helps beginners familiarise themselves with the squat movement, serves to occasionally change the routine and increase the lifted load in experienced athletes, accommodates the loads in individuals with injuries who may experience pain when executing the squat, and provides a safe form of closed kinetic-chain exercise for rehabilitation [11,14-17]. Because a Smith machine allows the manipulation of foot placement, this type of equipment provides some control over the compressive and shear forces acting on the lower back [15,18]. In particular, shear forces can expose the athlete to disc or facet injury. To eliminate shear forces at the lumbo-sacral joint, the spine must remain neutral throughout the squatting movement [14-16]. However, shear forces are difficult to reduce in a free squat [15]. It is, therefore, important to gain a deeper understanding of the spinal loads associated with the Smith squat.

Restricting anterior movement of the knee in an attempt to reduce the internal forces exerted on the knee joint may disproportionately augment musculo-skeletal forces on the hips and lumbar spine [2,15,17,18]. Similarly, changing from the high bar and the low bar squat techniques is thought to cause a redistribution of loads on the lower back and knee joints. However, to the best of the authors' knowledge, there is no evaluation of the compression and shear loads on the lumbar spine that compares the high bar and low bar isometric Smith squats with maximal external resistance. Therefore, this study aims to determine the internal forces incurred in the 1RM high bar and low bar isometric parallel-depth Smith squats. The findings of this biomechanical evaluation can be used to guide the use of the isometric parallel-depth Smith squat in strength training, so that the training is safe and prevents lower back injury. The findings have also applications for the design of injury rehabilitation programs in which it is desirable to manipulate muscle recruitment and joint loads.

METHOD

Study Design

In this study we present part two of a larger study on the biomechanics of the parallel squat. Part one reported musculoskeletal loads on the knee and ankle [4], whereas part two reports loads on the lower back. The same methodology was used for both parts of the study, therefore the reader is advised to read both papers in conjunction with one another. An experimental repeated measures design was employed in which the same participants performed both the high bar and low bar isometric parallel-depth Smith squats. Data collection was carried out under laboratory conditions; whereby the warm up protocols, position of the feet directly below the barbell at set-up, and the definition of the bottom of the squat were monitored closely in the controlled lab environment. The foot positioning at set-up permits standardisation of the initial squatting position, which is also a natural set-up position for the athlete [1,2].

Participants

A group of 8 healthy male 400-m sprinters of club level took part in the study (mean ± SD age = 22.3 ± 1.4 years, height = 178.9 ± 10.2 cm, mass = 80.0 ± 12.6 kg, 400-m performance = 51 ± 1.5 s). This study was approved by the Institution's Research Ethics Committee.

Execution of the Squatting Movements

The technical model illustrated by Baechle and Earle [2] was adopted in the present study to standardise the squatting technique of the participants [4]. The high bar and low bar squats were performed on a LifeFitness Hammer Strength 7° backward-inclined Smith machine. The backward-inclined type of Smith machine was selected due to being a ubiquitous system in many strength training venues [7,12,13,18]. The participants' 1RM was established two days prior to the test using the high bar free squat technique, since the participants stated that they were accustomed to establishing their 1 RM using the high bar free squat. The 1RM was determined using the *direct method* [4,19]. Specifically, the *direct method* protocol involved 2 to 3 warm-up sets and then 2 to 3 working sets [5,13,20]. The working sets consisted of choosing a weight that the participants thought they could lift for 1RM [21], whereby the chosen weight was adjusted throughout these sets to determine the maximum lifting capability. The mean ± SD 1RM load was 100.3 ± 7.2 kg (1.25 ± 0.09 BW). The warm up protocol and execution of the squat movements on the day of data collection are reported previously [4].

Figure 1: Definition of the 'bottom of the squat' with the thighs approximately parallel to the ground

During the squat, the participants paused for 2-3 s at the bottom of the squat (Fig. 1); this quasi-static position was used for the application of static mechanical analysis [18,22].

Videoing and Digitisation

Two-dimensional video recording of the Smith squat movements and subsequent manual digitisation of video images were carried for the construction of a stick figure that consisted of trunk, thigh, shank, and foot segments, as described previously [4]. Data obtained from digitisation was used for the calculation of moment arms, joint and segmental angles, and internal forces by applying principles of Statics [2,18,22-24].

LUMBO-SACRAL JOINT

Calculation of weight over the lumbo-sacral joint

The weight over the lumbo-sacral joint (W) was obtained using *eq.* 1.

$$\boldsymbol{W} = \left[\left(m_a \times m_{ub}\right) + m_b\right] g \tag{1}$$

where, m_a = mass of the athlete, m_{ub} = ratio of upper body mass to total body mass, m_b = mass of the barbell, and g = gravitational acceleration [22,24].

Determination of lower back extensor muscles force

The lower back extensor muscles force (M) was obtained using the 2nd condition of equilibrium (ΣMoments = 0); *eq.* 2.

$$\left(\boldsymbol{W} \times d_w\right) + \left(\boldsymbol{M} \times d_m\right) = 0 \tag{2}$$

where, d_w = moment arm of the weight over the lumbo-sacral joint and d_m = moment arm of the lower back extensor muscles force held constant at 0.05 m (Fig. 2, in which θ = angle of trunk inclination to the right horizontal) [22,24]. In Fig. 2, clockwise moments are positive and anti-clockwise moments are negative.

The horizontal (F_h) and vertical (F_v) force components, resultant joint reaction force (R), and angle of the resultant joint reaction force (θ_1) were determined using *eqs.* 3-6, respectively.

$$F_h = M\cos\theta \tag{3}$$

$$F_v = M\sin\theta \tag{4}$$

$$R = \sqrt{F_h^{\,2} + F_v^{\,2}} \tag{5}$$

$$\tan\theta_1 = F_v / F_h \tag{6}$$

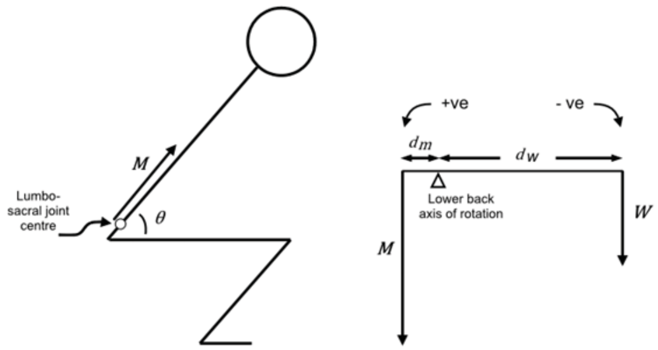

Figure 2: Angle θ and M (left) and moments about the lumbo-sacral joint (right) [2,21,23]

M = lower back extensor muscles force; θ = angle of trunk inclination to the right horizontal; W = weight over the lumbo-sacral joint; d_w = moment arm of W about the lumbo-sacral joint axis of rotation; d_m = moment arm of M.

Calculation of compressive and shear force

The compressive ($\mathbf{R}_c$) and shear ($\mathbf{R}_s$) forces were resolved from **R** (*eqs.* 7 & 8; where, θ_2 is the angle of the compressive force) [22,24]; Fig. 3.

$$R_c = R\cos\theta_2 \tag{7}$$

$$R_s = R\sin\theta_2 \tag{8}$$

The data were diagnosed for normality of distribution using Kolmogorov-Smirnov and Shapiro-Wilk normality tests. The data met the assumptions of normality ($p \le 0.200$). Paired *t*-tests were used to evaluate the differences in musculoskeletal forces between the high and low bar techniques. The significance level was set at a Bonferroni-corrected $p < 0.0125$. Effect size (η^2) and statistical power were obtained.

RESULTS

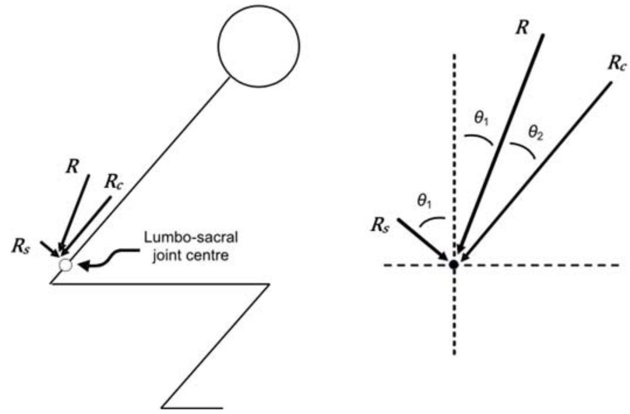

Figure 3: Computation of Rc and Rs at the knee joint [2,21,23]

R = resultant joint reaction force; R_c = compressive force at the lumbo-sacral joint; R_s = shear force at the lumbo-sacral joint; θ_1 = angle of the resultant joint reaction force; θ_2 = angle of the compressive force.

When using the high bar technique, the athlete's trunk was in a more vertical position (θ = 68.3° ± 4.6° for the 8 participants); thus, the moment arm about the lumbo-sacral joint was shorter (0.25 ± 0.04 m) than in the low bar technique (θ = 63.1° ± 7.1°; 0.27 ± 0.04 m), whereby the hip was posterior to the bar. The counterbalancing torque generated by the spinal extensor muscles was 372.8 ± 43.0 Nm (high bar) and 402.6 ± 50.0 Nm (low bar). Spinal extensor muscular forces and lumbo-sacral joint forces were higher when using the low bar technique (Fig. 4); with the exception of $\mathbf{R}_s$ which was approximately equal. The participants yielded higher inter-participant variation (SDs) in the low bar technique. When using the low bar technique, **R** and $\mathbf{R}_c$ reached forces relative to body weight (BW) of 11.3 ± 1.7 BW and 11.1 ± 1.5 BW, respectively. The angles of the resultant joint reaction force were 80.2° ± 2.5° (high bar) and 81.3° ± 2.3° (low bar). The differences between the high bar and low bar techniques were statistically significant for all variables, with the exception of $\mathbf{R}_s$ (Table 1).

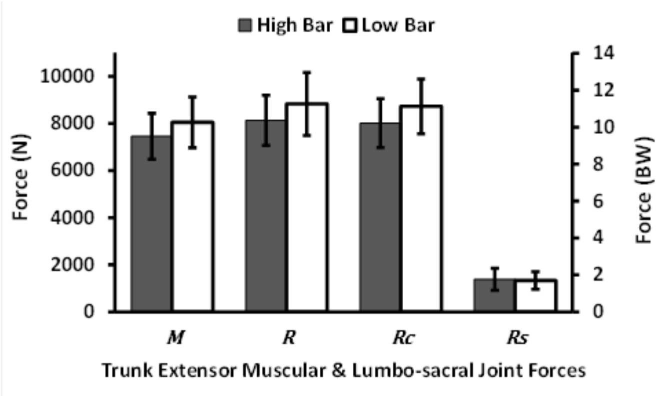

Figure 4: Mean ± SD muscular and joint forces at the lower back.

Table 1: Results of the paired t-tests.

Variable	t value	df	sig.	η^2	power
M	-12.13	7	0.001	0.96	1.00
R	-6.94	7	0.001	0.87	1.00
R_c	-9.24	7	0.001	0.92	1.00
R_s	1.34	7	0.223*	0.20	0.21

* Non-significant (Bonferroni-corrected p < 0.0125).

M = lower back extensor muscles force; R = resultant joint reaction force at the lumbo-sacral joint; R_c = compressive force at the lumbo-sacral joint; R_s = shear force at the lumbo-sacral joint.

DISCUSSION

The $\mathbf{R_c}$ calculated in the present study were 8,014 ± 1,035 N or 10.2 ± 1.3 BW (high bar) and 8,729 ± 1,161 N or 11.1 ± 1.5 BW (low bar), based on an external resistance of 1.25 ± 0.09 BW used in the mathematical model (Fig. 4). In previous research, compression loads on the L3–L4 segment during the half and quarter back squats using external resistances between 0.8 and 1.6 BW were between 6 to 10 BW (3,100 – 7,324 N) at the turning point of the squat [6]. Thus, the findings suggest that the parallel squat depth causes higher loading of the passive spinal tissues than the half and quarter back squats [6]. It is known that intervertebral discs prolapse in a load combination of high axial compressive and shear forces in ventral flexion [6,14]. Therefore, the results suggest that when using the low bar technique, in which the axial compression is higher, there is a greater risk of passive tissue failure than in the high bar technique [5,14,19,22]. In axial compression, the vertebral body is the weakest link and is the initial structure that fails due to compression failure and fracture of the endplate [6,14,22]. However, long term strength training that exposes the spine to high axial compression loads can result in functional adaptations of the vertebral bodies, manifested as increased bone mineral density (BMD) and therefore enhanced compression tolerance [5,6,14,20,22]. Compressive strength can be defined as the ultimate load a vertebral body can tolerate under axial compression prior to failure [5,22]. In ex-vivo measurements, a positive and linear correlation of $r = 0.82$ ($p = 0.00001$) has been reported between BMD and compressive strength of the vertebral bodies (L3, n = 101) [14]. Compressive strength of a vertebral body has been calculated as 11,000 N in previous work [6], which is higher than the mean $\mathbf{R_c}$ values obtained in the present study of 8,014 N ± 1,035 N (high bar) and 8,729 N 1,161 N (low bar) indicating, therefore, a low likelihood of acute vertebral body failure when performing the 1RM squat [5,6,14,20,22].

The present study includes male athletes only. However, the use of maximum resistances may yield excessive spinal compressive stress and pose a risk of injury to female athletes [14]. Hartmann *et al.* [6] reported the compressive strength (ex-vivo) of an L4/L5 vertebral segment for a 22-year-old man (8,800 N), that of an L3/L4 vertebral segment for a 22-year-old woman (6,200 N), and of a 39-year-old man (8,200 N). Thus, analysis of compressive strength suggests that females are at higher risk [6,14] due to a smaller end-plate cross-sectional area of the vertebral body than in males [14-16]. Consequently, the biomechanical analysis of high bar and low bar isometric Smith squats should be extended to the assessment of spinal loads in females. Nonetheless, past research suggests that increased BMD and well-developed trunk musculature can provide protection against injury in well trained athletes when using heavy resistance [5,6,14,20]. In the study of Lavallee and Balam [14], elite athletes of mean age 17.4 years and mean training experience of 2.5 years possessed 133% higher BMD of the L2–L4 vertebrae than a control group. The BMD of these young athletes also exceeded the normative values of 400 adult men of age between 20 and 39 years by 113% [14]. Furthermore, Hartmann *et al.* [6] found a positive linear correlation between the BMD of lumbar vertebrae (in vitro) and both the tensile strength ($r = 0.84$, $p < 0.05$) and the stiffness of the anterior longitudinal ligament ($r = 0.78$, $p < 0.05$). Both tensile strength and stiffness promote greater passive stability of the vertebral segments in vivo. Therefore, the spinal loads calculated for the isometric parallel-depth Smith squat in the present study do not appear excessive for well-trained male athletes [3,6,14,20,22], however whether the 1RM low bar technique is adequate for female athletes should be investigated in future research.

In the present study, the mathematical modelling of lumbar spine muscular and joint forces is limited to the bottom of the squat position. Further work may include analysis of the loads at various depths of the isometric Smith squat [1,3,6,8,18,23]. Future research may evaluate the combined effect of using the low bar technique and restricted anterior movement of the knee [7,17]. Such combinations attempt to reduce the forces exerted on the knee but may considerably increase forces exerted on the lumbar spine [17]. It has been suggested that deep squats may not necessarily cause an increased injury risk of the lumbar spine due to wrapping effect, functional adaptations, and soft tissue contact between the trunk and the front surface of the thigh [2,6]. Therefore, future static and dynamic models may incorporate these inherently protective factors to more accurately estimate musculo-skeletal injury risk during the squat. More comprehensive mathematical models for the calculation of spinal forces may also incorporate three-dimensional motion analysis to explore the effects of hyperextension of the lumbar spine and sacrum movements during the squat exercise [15,16]. The static biomechanical evaluation provides an in-depth understanding of the two back squat modalities in isometric conditions and provides a basis for the development of dynamic models of the high bar and low bar squat that incorporate accelerations of the lifted weight, as well as accelerations and inertial effects of body segments [9,10,15,22-24].

CONCLUSION

The high bar technique involves a more vertical position and a mean 29.8 Nm lower counterbalancing torque generated by the spinal extensor muscles. Spinal extensor muscle forces and lumbo-sacral joint forces were therefore higher when using the low bar technique; with the exception of $\mathbf{R}_s$ which was approximately equal. Thus, the low bar technique may be avoided in therapeutic interventions as it poses a greater risk of passive tissue failure. The presence of shear forces at the lumbo-sacral joint of approximately 2 BW suggest potential for damage to the intervertebral disc, as intervertebral discs are known to prolapse in a load combination of high axial compressive and shear forces in ventral flexion. The biomechanical assessment of high bar and low bar isometric parallel-depth Smith squats should be extended to female athletes, in whom for a given external resistance the lumbar vertebral body is exposed to higher axial compressive stress than in males. There is evidence that suggests that the human intervertebral discs are responsive to training stimuli and the compression tolerance of the discs increases with long term squat training. Increased BMD and well-developed trunk musculature through accumulated squat training also provide protection against injury in well trained athletes. Therefore, the spinal loads found in the present study do not appear excessive for well-trained athletes. However, the choice of squat modality may be important in the rehabilitation of spinal injuries.

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