EFFECTS OF INTERNAL KINETICS AND MUSCLE ACTIVITY DURING THE WIDE AND NARROW BARBELL BACK SQUAT

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ABSTRACT

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Introduction. The barbell back squat (BBS) is a commonly utilized strength exercise to support general preparedness for the demands in multiple sports. Recently, studies have shown various strength training exercises have the potential to recruit the regions of the hamstrings differently. Specifically, it is unknown if changing stance width under conditions where forward knee movement is restricted engages the hamstring regions differently and how this corresponds with measured 3-D net joint moments (NJM) demands on the prime extensor joints. Therefore, the goal of this thesis is to investigate the acute effects of utilizing the wide barbell back squat (WBBS), and the narrow barbell back squat (NBBS) to femur parallel depth on the hip and knee musculature in a population of intermediate athletes.

Methods.14 amateur rugby players (6 males, 8 females, age 27.36 ± 3.71 years; height: 174.1 ± 10.4 cm, body mass 81 ± 21.86 kg, squatting experience: 3.93 ±1.77 years) completed a 3-week familiarization period to learn controlled versions of femur parallel depth WBBS and NBBS. On the 4th week all subjects completed a strict technical 1 RM test protocol for both widths on separate days. On the 5th week, all subjects performed WBBS and NBBS with 70% and 85% of 1-RM loads (shoeless, tempo 3-0-0), where biceps femoris long head (BFLH) and semitendinosus (ST) activity were recorded with 15-channel high-density electromyography (HD-EMG) with both overall and 5 channel regional divide (distal, medial proximal), 3-D net joint moments (NJM) of lower lumbar, hip and knee, and bipolar sEMG from gluteus maximus (GM) and vastus lateralis (VL). All sEMG was normalized and analysed from the ascent phase.

Results. The WBBS had higher hip flexion (85% load), hip abduction, and hip internal rotation ($p<0.05$). The NBBS had higher knee flexion and dorsiflexion ($p<0.05$). Ascent time changed with load but not descent time and no effects of width were noted ($p>0.05$). WBBS had higher activity in BLFH and ST overall, and in medial (85% load) and proximal (both loads) regions ($p<0.05$). Hamstring activity increased with load in the NBBS 85% BFLH proximal region, WBBS 85% BFLH overall, distal, ST overall, medial, and proximal regions ($p<0.05$). No hamstring regional interaction differences were found in all loads ($p>0.05$). At both loads, the WBBS had higher 3-D hip-to-knee NJM and hip-to-knee extensor NJM ratios ($p<0.05$). Hip-to-knee ratios did not increase with load ($p>0.05$). The WBBS had higher 3-D hip NJM, hip sagittal NJM, hip frontal NJM, hip transverse NJM, and knee frontal NJM ($p<0.05$). The NBBS had higher 3-D L5/SI NJM (85% load), 3-D knee NJM (85% load) and knee sagittal NJM ($p<0.05$). Both heavier loads of WBBS and NBBS had higher 3-D L5/SI NJM, 3-D hip NJM, hip sagittal NJM, and 3-D knee NJM ($p>0.05$). In the WBBS (70% load) GM activity was higher compared to the NBBS ($p<0.05$). Between the NBBS loads, GM activity increased with heavier load ($p>0.05$). VL activity did not change significantly between widths or load ($p>0.05$).

Discussion and practical applications. Although reaching statistical significance, WBBS hamstrings activity was only at low levels, ranging with a mean of 26-38% of MVIC across loads. But considering the combined effect of the biomechanical differences, the WBBS around femur parallel depth might provide different long-term benefits compared to the NBBS. Specifically, sports that have high 3-D hip demands and knee stability demands, such as team or individual sports that involve multiple change of directions, should potentially vary stance width combined with control of forward knee movement in the barbell back squat to possibly gain more functionality in strengthening the lower limbs while not increasing the load on the lumbar. Long-term studies are needed to further confirm practical relevance.

Keywords. Sports science, strength training, back squat, stance width

LIST OF COMMON ABBREVATIONS

AL: Adductor longus

AM: Adductor magnus

APT: Anterior pelvic tilt

BBS: Barbell back squat

BFLH: Biceps femoris long head

COM: Centre of mass

COP: Centre of pressure

sEMG: Surface electromyography

ES: Effect size

GM: Gluteus maximus

GRF: Ground reaction force

GT: Greater trochanter

HD-sEMG: High density surface electromyography

L5/SI: The 5th lumbar vertebrae and the Sarcroiliac-joint.

NBBS: Narrow barbell back squat

NJM: Net joint moment

PCSA: Physiological cross-sectional area

PPT: Posterior pelvic tilt

RF: Rectus femoris

SD: Standard devation

SEM: Semimembranosus

ST: Semitendinosus

TUT: Time under tension

VI: Vastus intermedius

VL: Vastus lateralis

VM: Vastus medialis

WBBS: Wide barbell back squat

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1 INTRODUCTION

Many practitioners and scientists consider the barbell squat to be the most effective exercise for developing lower body strength (Wretenberg et al. 1996, McCaw & Melrose. 1999, Escamilla et al. 2001, Fry et al. 2003, Gullett et al. 2008, Paoli et a. 2009, Pereira et al. 2010, Schoenfeld. 2010, Bryanton et al. 2012, Clark et al. 2012, Swinton et al. 2012, Aspe & Swinton. 2014, Hooper et al. 2014, Chiu et al. 2016, Contreras et al. 2016, Hammond et al. 2016, Slater & Hart. 2017). Its wide use for different populations further increases its popularity. The barbell squat has been used successfully for sports performance (Styles et al. 2016), for sedentary populations (Bloomquist et al. 2013), in rehabilitative settings (Neitzel & Davies. 2000), for adolescents (Myer et al. 2005), for the elderly (Hagerman et al. 1999) and it is used as an actual competition exercise (or it is a part of a larger movement pattern) in sports such as powerlifting, Crossfit and Olympic weightlifting (Escamilla et al. 2001b, Ho et al. 2011, Hooper et al. 2014). Due to its applicability in such a vast array of the population, multiple variations have been developed and utilized in practice. Of these squat variations, many have been objected to biomechanical research. Variations that have been researched include and are not limited to; different widths (McCaw & Melrose. 1999, Paoli et al. 2009, Swinton et al. 2012), unilateral barbell back squat vs. barbell back squat (BBS) performed bilaterally (McCurdy et al. 2010), BBS vs. barbell front squat (Yavuz et al. 2015), BBS vs. overhead squat (Aspe & Swinton. 2014), changes in degree of stability (Scwankbeck et al. 2009), different depths (Contreras et al. 2016), and restricted vs. non-restricted BBS (Fry et al. 2003, Chiu et al. 2016). Biomechanical disciplines employed include kinematics, kinetics, and surface electromyography (sEMG) with varying approaches and limitations. In addition, several meta-analysis and reviews have been conducted on the squat exercise (Schoenfeld. 2010, Clark et al. 2012, Seitz et al. 2014). The BBS is debatably the most studied version of squatting within sports science, but countless unanswered questions concerning its evidence based utilization in practice still exist. Therefore, due to its vast universal applicability within many sports, more understanding of its complexity via in depth research built on what is already more or less known should further add value to its use.

2 THE GENRAL ROLE OF THE BARBELL BACK SQUAT IN ATHLETIC DEVELOPMENT

Strength training movements are chosen based on their capability to generate general neuromuscular and physiological adaptations, but also for more specific movement pattern and metabolic reasons according to the individual athlete’s needs. As a general “non-specific” exercise, the barbell squat offers superior neuromuscular and morphological adaptations to the lower-body’s musculature and even specific trunk musculature (i.e. erector spinae) compared to many other strength training exercises (Hamylin et al. 2007, Schoenfeld. 2010, Bompa & Buzzichelli. 2015. p. 139). The prime movers in barbell squat come from the both the hip and knee extensor group, which debatably means that any athlete that has value of strengthening the hip and knee extensors for their sport could value of implementing barbell squat into the program at least at some point of the yearly training program (Schoenfeld.2010, Bompa & Buzzichelli. 2015. p. 139). Even though the barbell squat for most athletes is not directly a sports specific stimulus, it has the potential to support the strengthening of something called proximal-to-distal kinetic energy sequence (Robertson et al. 2008). The scientific theory of proximal-to-distal energy sequence states in basic terms that in athletic movements such as sprinting & jumping, kinetic energy is optimally transferred in the body if it moves from the hip to the knee and then to the ankle (Bobbert & Schenau. 1988, Jacobs & Schenau. 1992a), which is also called triple extension mechanics. Squatting can be considered a highly suitable candidate for strengthening such coordinative energy transfer (Robertson et al. 2008).

Also, the relative contribution of net joint moments (NJM) to athletic movement, especially between the hip and the knee, is something that the barbell squat might be able to influence positively. In this context and simplified, NJM are the product of net muscular torque of the agonist and antagonist group at a specific joint and the moment arm length in a specific biomechanical plane, measured in Newton meters (Nm) (Beardsley & Contreras. 2014). Some authors have categorized compound lower-body movements as either “knee dominant” or “hip dominant”. This is usually calculated by taking the peak hip and knee extensor NJM and creating the following ratio; if the hip-to-knee extensor NJM is less than 1 the movement is categorized as knee dominant and if it is greater than 1 it is hip dominant (Riemann et al. 2012, Beardsley & Contreras. 2014). Some studies that has observed how NJM at the hip and knee behave in athletic tasks such as specific jumping tasks, have demonstrated that the NJM can significantly shift between the hip and knee when effort is added. Higher hip-to-knee extensor NJM have been noted with increasing running speed (Schache et al. 2011), increased vertical jump height (Lees et al. 2012) and shifting from a vertical jump to a horizontal jump (Sugisaki et al. 2014). All three studies noted significant hip dominance at the highest intensities, but what was also interesting is that Schache et al. (2011) and Lee et al. (2012) noted that at lower intensities there was a presence of knee dominance. Based on these studies, it seems that the role of the hip extensors should be potentially prioritized in strength training for most sports that involve jumping and/or sprinting, at least in the horizontal direction, but without underestimating the role of the knee extensors. Similarly, the BBS ratio of hip-to-knee extensor NJM seem to increase with increasing load (Bryanton et al. 2012. Vigotsky & Bryanton 2016). But categorizing different BBS forms (i.e. wide vs. narrow) to either hip dominant or knee dominant based on the ratio of hip-to-knee extension NJM is slightly less clear. Beardsley & Contreras (2014) used Bryanton (2011) data and found that the narrow barbell back squat (NBBS) to thigh parallel depth became significantly more hip dominant at higher loads with 50% of 1 RM showing a ratio of 1.12:1 and 90% of 1 RM showing a ratio of 1.49:1. But according to Bryanton's (2011) data, this significant shift was largely due to that peak knee extensor NJM did not increase with increasing load, whereas in other BBS studies it has been found to increase (Cotter et al. 2013).

There is consensus within the sports science community in that increased strength to certain extent in the BBS has good carryover to sport performance in most athletic populations (Cronin et al. 2007, Schoenfeld 2010). This is due to that moderate to large correlations have been found between maximal BBS strength and acceleration capability, change of direction, and contact strength (Baker & Nance. 1999, Wisløff et al. 2004, McBride 2009, Comfort et al.2012, Speranza et al. 2016). The degree of carryover depends on such details as the athlete's level (Cronin et al. 2007), but also in how the BBS is performed. For example, long-term strength training studies have been performed on different depths of BBS and unilateral vs bilateral BBS (Hartmann et al. 2012, Bloomqvist et al. 2013, Speirs et al. 2015, Rhea et al. 2016). These studies demonstrate that when deciding to utilize BBS in athletic development settings, one of the first considerations of the practitioner should be what technical variation it is utilized. On this note, there is still no long-term strength training study that has compared different squatting widths effect on performance outcomes or even acute correlations to performance markers. This is possibly because proper interest has not been "woken" for the topic within the sports science community.

3 INTERNAL KINETICS AND EXTERNAL KIMEMATICS OF THE SPINE, HIP AND KNEE IN THE BARBELL BACK SQUAT AT NARROW AND WIDE WIDTHS

The differences between NBBS and wide barbell back squat (WBBS) can be simplistically distinguished by how wide the legs are placed from each other and where the barbell is placed. In terms of stance width, according to some sources NBBS is defined as a width equal to the distance between the greater trochanters (Paoli et al. 2009) or around shoulder width (Escamilla et al. 2001b, Myer et al. 2014) and a WBBS is defined as 1,5+ times the width of the greater trochanter (Paoli et al. 2009) or around 150%+ wider than the shoulders (Escamilla et al. 2001b). Width does though slightly vary due to inevitable differences in anthropometry of the lifter. A potentially easier – more anthropometrically friendly – approach to define the differences in the NBBS and WBBS is an explanation that takes into consideration the primary kinematic goals. Potentially more evident in populations outside of powerlifting, one common kinematic goal in a wider stance is to restrict the anterior movement of the knee-joint without causing excessive lean of the trunk and therefore potentially create more torque through the hip-joint compared to the knee-joint. In a narrower stance, the knees are in general “allowed” to travel more freely in the anterior direction, therefore there is even slightly more flexion from the knee-joint relative to the hip-joint, which usually allows a slightly more upright trunk (Swinton et al. 2012).

The utilized barbell position on the back varies between practitioners. In a NBBS, the barbell is usually placed on the top of the trapezius near the 7th cervical vertebrae (Wretenberg et al. 1996, Fry et al. 2003), also called “high bar” (Goodin. 2015). In the WBBS the barbell is usually placed slightly lower. This varies in literature and stereotypically is connected to the term “low bar” position, which is usually defined as placing the barbell two inches below the superior aspect of the shoulders (Goodin. 2015). But a slightly higher position is also accepted in literature, that could be considered “mid-bar” positioning (term not utilized in literature), due to being placed between the typical high bar and low bar position. This would be on the posterior deltoids and inferior or across the scapular spine (Wretenberg et al. 1996, Fry et al. 2003).

In terms of kinetics, BBS studies that have compared different widths have both examined differences in internal and external kinetics. Because external kinetics such as peak power, peak force, peak velocity, and rate of force development do not seem to differ significantly between the two width variations (Swinton et al. 2012), this literature review will focus on the differences in the internal kinetics among the more detailed kinematic differences. To comprehend and shed light to the complexity of the topic, each adjacent joint that is significantly loaded and are considered main movers in the BBS will be examined separately.

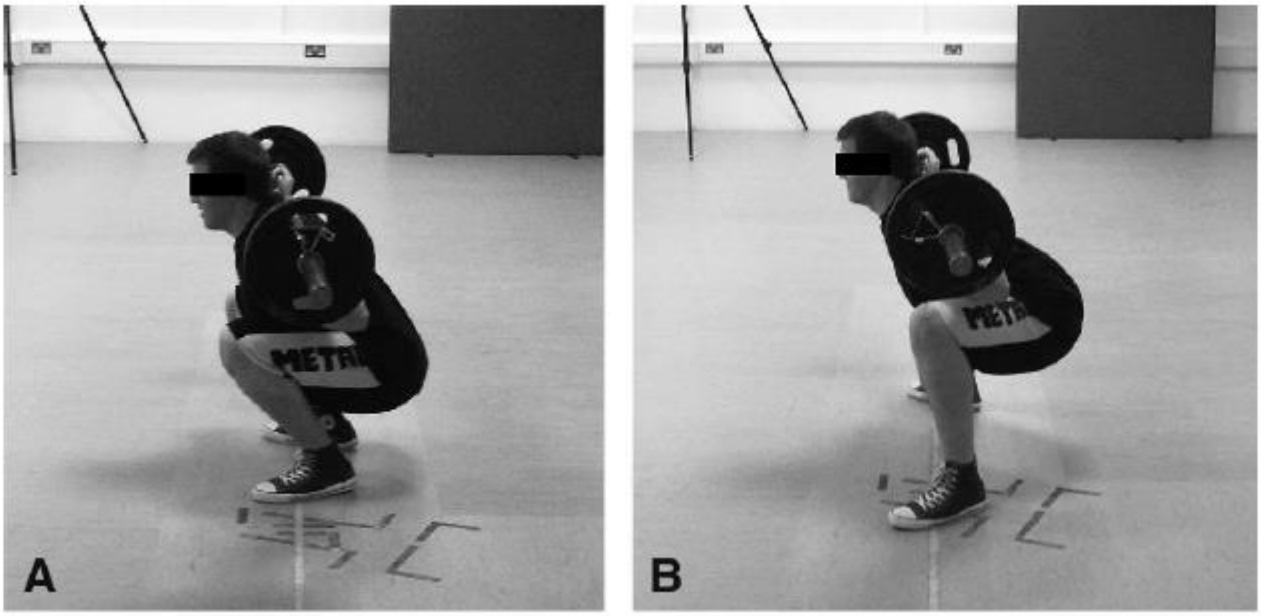


FIGURE 1. The NBBS (A) and WBBS (B) (Swinton et al. 2012).

3.1 The spine and the pelvis

3.1.1 The spine

The spine is comprised of 24 mobile vertebral segments, each displaying 6 degrees of freedom. Also, there are five fused vertebrae that create one big segments called the sacroiliac (SI) joint. Individually and as a unit, the spine is capable of flexion and extension in the sagittal plane, lateral flexion in the frontal plane, and rotation in the transverse plane (Schoenfeld. 2010). An array of muscles supports the spine, also known as the “core” muscles (Key. 2013), which is a substantially complex topic on its own. Many of these muscles have a task to isometrically stabilize the spine in dynamic lower and upper limb movement (Lee & McGill. 2015). In terms of the barbell squat, the spinal stabilizers ensure that a stable, upright posture is maintained throughout the movement (Schoenfeld. 2010). There are at least a few studies using a squatting movement patterns that have observed spinal movement with 3D motion capture (Walsh et al. 2007, Kingma et al. 2010, McKean et al. 2010). Studies such as McKean et al. (2010) reported significant lumbar flexion in the deepest positions of the NBBS and WBBS. What was also interesting is that subjects tended to flex the lumbar spine directly when the load was placed on their back in the starting position in both narrow and wide widths (McKean et al. 2010). Even though some similarities can be found, spinal segment movement does seem to differ between studies. McKean et al. (2010) reported increased lumbar flexion (hypolordotic) at the bottom of both widths, while Walsh et al. (2007) reported a more hyperextended (hyperlordotic) lumbar position on the bottom position of a NBBS.

Because the barbell is placed on the upper or lower deltoids in the BBS depending on the technique used (low-, mid-, high bar), it creates a significant moment arm between the barbell and the lumbar spine (Swinton et al. 2012). This large moment arm to a significant extent explains the relatively high surface electromyohraphic (sEMG) activity found in the lumbar region when performing a BBS compared to other lower back exercises (Hamylin et al. 2007). The lumbar erector spinae possibly contribute the most to spinal stabilization in the BBS by helping resist vertebral shear forces and maintain anteroposterior spinal integrity (Delitto & Rose. 1992. Hamylin et al. 2007. Schoenfeld. 2010). Increasing this external moment arm via excessive trunk lean has been proposed as a predecessor for forces on the spine that might lead to injuries (Fry et al. 2003. Chiu et al. 2016). As demonstrated by Fry et al. (2003), the moment arm on the lumbar further increased when restricting anterior knee movement. This led the authors to conclude that restricting anterior movement of the knees can possibly lead at some point to vertebrae injury, most prominently in the lumbar spine region. Also, a recent study by Chiu et al. (2016) restricted knee movement and named the phenomenon anterior knee rotation movement. This kinematic structure among other variables Chiu et al. (2016) measured (Figure 2), provides significantly more complexity to the squatting movement pattern from a sagittal plane viewpoint.

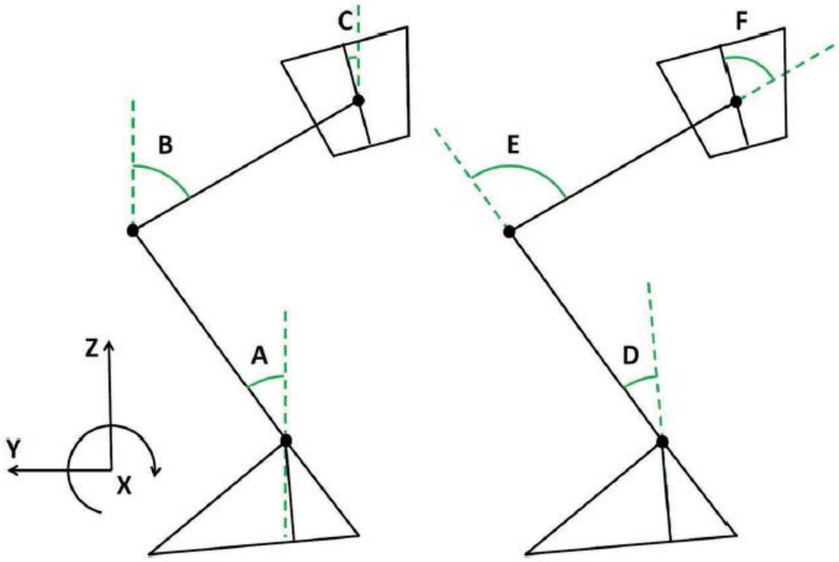


FIGURE 2. Illustration of sagittal plane (X-axis) segment and joint angles. A - anterior leg rotation; B - posterior thigh rotation; C - pelvic anterversion; D - ankle dorsiflexion; E - knee flexion; F - hip flexion (Chiu et al. 2016)

Although valuable information was attained, Fry et al (2003) and Chiu et al. (2016) restricted knee movement at the same stance width and did not even discuss the possible limitations. Swinton et al (2012) study showed that increasing stance width from a NBBS into a knee movement restricted WBBS allowed restriction of anterior knee movement without increasing the moment arm on the lumbar in the sagittal plane (compared at thigh parallel depth). In fact, not only was the moment arm at the lumbar spine very similar between the WBBS and the NBBS, but the NJM was found to be lower at the lower lumbar for the anterior knee movement restricted WBBS at the same absolute load (Swinton et al. 2012). Escamilla et al. (2001a) also found that in the BBS trunk lean did not increase with less anterior knee movement if the squat stance was widened. This phenomenon is explained by the increased hip abduction in a WBBS, which shortens the distance between the knee and the hip in the horizontal plane. This leads potentially to the femurs “pushing” the lifter less back when sitting on the hip, which leads to less distance between the lumbar and the barbell. Another important method that can be used in any form of barbell squat is to increase spinal integrity via creating Intra-Abdominal Pressure (IAP). With the help of trapping air into the body via the Valsalva maneuverer, IAP is established and has been found to significantly increase the support around the lumbar in a squat (McGill et al. 1999). Also, it has been shown that a downward gaze increases trunk flexion by 4.5° and hip flexion by approximately 8° compared with a straight ahead or upward gaze (Donnelly et al. 2006).

3.1.2 The pelvis

The spine should not be seen in just isolation. The sacrum attaches to the pelvis, creating the Sarcroiliac-joint (SI – joint) (Sturesson et al. 2000). The actual SI-joint itself has only around 0.3 mm of movement and worsens with age (Sturesson et al. 2000), meaning if the SI-joint visually moves, then both the lumbar and pelvis have probably moved. The pelvis is the origin/insertion of quite many important trunk- and lower body muscles for dynamic movement, therefore movement of the pelvis will affect not only directly the passive structures of the lumbar but the moment arm of the attached muscles and consequently potentially their activation intensity and patterns (Delitto & Rose. 1992, Hogervorst & Vereecke. 2015). This has been indirectly shown via sEMG research, where activity of the erector spinae significantly changed when the pelvis was actively manipulated into different starting positions in the initiation of the ascent phase of a bodyweight squat (unfortunately, no lower body muscles were observed) (Delitto & Rose. 1992). In the Delitto & Rose (1992) study subjects were asked to pick a box from the ground in a squat position from either a posterior pelvic tilt (PPT) position or an anterior pelvic tilt (APT) position. They showed that the sEMG activity of the erector spinae muscles was greater when subjects maintained an APT position instead of a PPT position. The oblique abdominals behaved activity wise the same in both conditions. The authors suggested that the greater trunk muscle activity occurring with the APT position may ensure optimal muscular support for the spine while handling loads, thereby reducing risk of back injury (Delitto & Rose. 1992). It has also been shown that in an experienced Olympic weightlifter, actively increasing APT in the starting position was in important variable for successful lifts (Ho et al. 2011). Although this being the case, it is probably wider to state that uncontrolled pelvic movement in any direction will not have the same trunk muscle support and may affect performance and at some point, lead to excessive shear forces on the lumbar that potentially lead to damage of the passive structures or even indirectly to other joint injuries around the body (Chaudhari et al. 2014). Quite logically, pelvic movement seems to be harder to control the more hip- and knee flexion there is in a squat (Schoenfeld. 2010. Nielsen. 2015), at least in untrained populations. Also, if the knees are restricted in a NBBS causing a large moment arm at the lumbar spine, there seems to be significantly more movement of the pelvis the deeper the squat goes (Chiu et al. 2016). Nielsen (2015) found in his master's thesis research that a wider foot placement with feet externally rotated decreased PPT movement at 70 degrees of knee flexion, which would imply potentially more lumbar control in terms of avoiding flexion in wider stances, at least if compared to a restricted NBBS. Research still is unclear on what causes the pelvis to excessively tilt and what the benefits and setbacks are. Some ideas have been proposed for the uncontrolled movement including weakness or unbalance of the stabilizing lumbar musculature, individual hip structure and the less accepted theory; “tight” hamstrings (Nielsen. 2015). In conclusion, stability around the spine and the pelvis should ideally be given considerable attention when aiming to quantify differences in compound lower-body exercises.

3.2 The hip-joint

The hip-joint, also referred to as the acetabulafemoral joint, is a ball-and-socket joint between the femur and acetabulum of the pelvis. The ball-and-socket joint format makes it freely mobile in all 3 biomechanical planes of movement, with flexion and extension in the sagittal plane, abduction & adduction in the frontal plane, and internal & external rotation in the transverse plane (Schoenfeld. 2010). The primary hip muscles involved in the barbell squat include the hip extensors gluteus maximus (GM), hamstrings (semitendinosus (ST), semimembranosus (SEM), biceps femoris long head (BFLH) and the adductor magnus (AM) (Schoenfeld. 2010, Vigotsky & Bryanton. 2016)

3.2.1 Hip – joint kinematics

Hemmerich et al. (2006) reported that 95° (SD ± 26) of hip flexion was required to reach maximal depths in a bodyweight squat. The standard deviation reported by Hemmerich et al. (2006) is quite large, so clearly there are different strategies of reaching full depth in a multi-joint movement such as the squat. A lack of hip flexion means potentially compensating with more trunk flexion to reach depth (Kim et al. 2015), which consequently as mentioned before can be harmful for the spine (Fry et al. 2003, Schoenfeld. 2010, Chiu et al. 2016). Wretenberg et al. (1996) compared the kinematics of powerlifters utilizing their version of the BBS to Olympic lifters utilizing their Olympic-style BBS. In the powerlifting BBS, hip flexion was reported to be 132° (SD ± 8) and for the Olympic squat 111° (SD ± 4) at thigh parallel. Although stereotypically a powerlifting BBS represents usually a wider stance width and an Olympic style BBS in general represents a narrower stance width (Swinton et al. 2012) this cannot be confirmed due to that Wretenberg et al. (1996) did not report specific stance widths. They only reported that the force plate was 60 cm wide and none of the subjects felt restricted in choosing freely their stance width within this space. Escamilla et al. (2001b) measured 39 powerlifting competition lifts that included both narrower stance styles (around shoulder width) and wide stance styles (~70% wider than shoulder width). Reported hip flexion angles to slightly below parallel were 107° (SD ± 10) and 110° (SD ± 7) for NBBS and WBBS, respectively. Very similar results to Escamilla et al. (2001b) were reported by Swinton et al. (2012), who measured three-dimensional (3-D) kinematics to parallel depth. The quite significant reported differences in hip flexion between Wretenberg et al. (1996) and Escamilla et al. (2001b) can be partly explained by the technology used to measure kinematics. Wretenberg et al. (1996) used two-dimensional (2-D) motion capture whereas Escamilla et al. (2001b) used 3-D and 2-D motion capture. Escamilla et al (2001b) results showed that 2-D motion capture has its setbacks for measuring biomechanical data, especially for wider stance squats. Based on the information presented by Escamilla et al (2001b) and Swinton et al. (2012), it seems that maximal hip flexion to reach parallel in both a WBBS and a NBBS differ minimally with slightly more hip flexion required in a wide stance.

In terms of the hip internal and external rotation, there seems to be lack of ROM guidelines for a full depth squat. Kim et al. (2015) reported in their deep squat ROM study that a lack of hip internal rotation ROM can cause difficultly to reach depth. Female lifters tend to have larger ROM in hip internal rotation compared to their male counterparts (Kim et al. 2015). This might be connected to anecdotal evidence of experienced female lifters possessing in general more squatting mobility than male lifters (Contreras et al. 2016). But forcing more internal rotation for some can be on a structural level damaging. High degrees of hip flexion with a combination of internal rotation might lead to hip pain for some due to for example how their femoral head is shaped, which if frequently irritated may lead to an acetabular labral tear (Lewis & Sahrmann. 2006). Anthropometric differences in any joint including the hip joint structure is a complex subject influenced by gender, ethnicity, heritage, exposure, and age (Loder & Skopelja. 2011), and it is slightly out of the scope of this thesis. Although this being the case, it is good for the practitioner and researcher to be aware of possible significant differences in hip joint structures. These differences will inevitably cause different squatting movement patterns due a variation of limiting range of motion factors (Lamontagne et al. 2011). There are quite a few important studies showing significant variation between males and females in different age groups from different ethnic backgrounds and even individual side-to-side differences in femoral neck structure and acetabulum shape (Fabry et al. 1975, Maruyama et al. 2001, Zalawadia et al. 2010). Currently, there is plenty of debate within the strength & conditioning training community on how to approach individual differences in hip structure when learning to squat. More research is warranted on this topic. It seems for now that there is no evidence against the assumption that the largest part of the population is anatomically capable of varying their squatting stance width to some extent and at least reaching depths around femur parallel without compromising joint integrity if coached properly.

A detailed comparison between the NBBS and the WBBS to thigh parallel depth was made by Swinton et al. (2012), where hip joint kinematics and kinetics were taken from all 3 biomechanical planes. Hip flexion and extension kinematics were slightly different between the NBBS and WBBS (Figure 3).

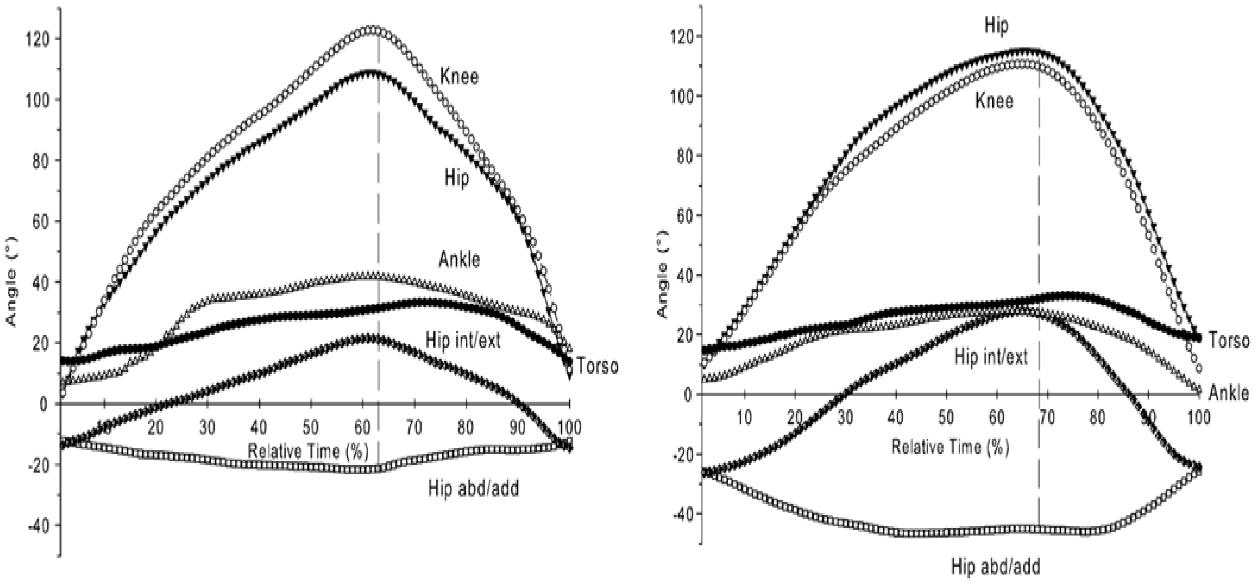


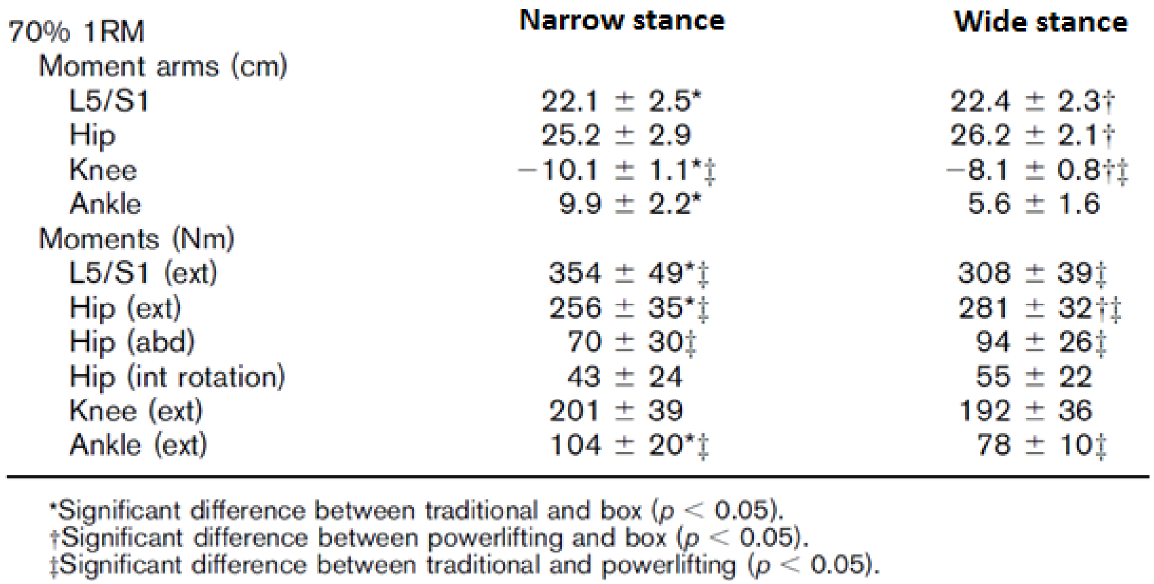
FIGURE 3. Joint angle-time curve observed during the NBBS (left) and WBBS (right). Dashed line indicates transition from the descent phase to the ascent phase (Swinton et al. 2012).

The largest differences in kinematics between back squatting widths could be observed in abduction & adduction and internal & external rotation mechanics. As we can see from figure 3, the WBBS (right figure) kinematics require significantly more hip adduction and hip external rotation in the starting position compared to the NBBS (left figure). Once in motion, adduction to abduction movement can also be observed in both techniques but more so in the WBBS.

3.2.2 Hip-joint internal kinetics

In Swinton et al. (2012) study, although similar sagittal plane hip external moment arm lengths at parallel were observed between the NBBS and WBBS, significantly higher hip extensor NJM were observed in the WBBS (Table 1).

TABLE 1. Joint moment arms and joint moments (Nm) in the NBBS and WBBS (Swinton et al. 2012).



70% 1RM	**Narrow stance**	**Wide stance**
Moment arms (cm)		
L5/S1	22.1 ± 2.5*	22.4 ± 2.3†
Hip	25.2 ± 2.9	26.2 ± 2.1†
Knee	−10.1 ± 1.1*‡	−8.1 ± 0.8†‡
Ankle	9.9 ± 2.2*	5.6 ± 1.6
Moments (Nm)		
L5/S1 (ext)	354 ± 49*‡	308 ± 39‡
Hip (ext)	256 ± 35*‡	281 ± 32†‡
Hip (abd)	70 ± 30‡	94 ± 26‡
Hip (int rotation)	43 ± 24	55 ± 22
Knee (ext)	201 ± 39	192 ± 36
Ankle (ext)	104 ± 20*‡	78 ± 10‡

*Significant difference between traditional and box ($p < 0.05$).
†Significant difference between powerlifting and box ($p < 0.05$).
‡Significant difference between traditional and powerlifting ($p < 0.05$).

A shift towards larger hip NJM have been reported before in experienced lifters when switching from a narrow to a wider stance (Wretenberg et al. 1996). On the contrary, Escamilla et al. (2001b) reported no mean differences between squatting widths, but this might have been due to that data was taken from powerlifting competition lifts, where the movement patterns are not manipulated by coaching at this point anymore (Swinton et al. 2012). The authors from Swinton et al. (2012) discussed that the increased extensor NJM at the hip are probably due to that experienced lifters in the WBBS emphasize hip extensor torque more than knee extensor torque when moving through the motions and therefore recruited the muscles surrounding the hip to a larger extent. Unfortunately, sEMG was not taken but this claim can be partially supported by increased mean GM sEMG activity reported from previous studies that have observed the effects of widening the stance in a BBS (McCaw & Melrose. 1999, Paoli et al. 2009). Both McGaw et al. (1999) and Paoli et al. (2009) did not test 1 RM separately for both WBBS and NBBS, but instead tested in the subjects preferred stance, therefore WBBS and NBBS loads were not relative. It is also worth noting that both sEMG and NJM values in isolation do not always give us an accurate picture of the relative strength requirements of a muscle group in a specific task (Criswell. 2011. p. 30, Bryanton et al. 2012). Another alternative measure for observing torque requirements for muscle groups in a specific task would be via Relative Muscular Effort (RME) measurements. This is a relatively new approach, which is measured as the ratio of a muscle group's NJM during a task relative to the muscle group's NJM during a maximal voluntary isometric contraction (MVIC) test at different angles. Unfortunately, this has only been done for a NBBS position. Either way, the results were interesting and showed that hip extensor RME increased more than knee extensor RME with increased load (50% and 90% of 1 RM were used) and both hip and knee extensor RME increased with depth (Bryanton et al. 2012). Unfortunately, more detailed knee and hip kinematics were not reported but this shift towards more hip extensor RME compared to knee extensor RME could have probably been observed as a movement pattern shift with increasing load (pushing the hip more back etc.). This has been shown for example in a BBS fatigue study by Lander et al. (1992) where an increased trunk lean was reported towards the last repetitions, probably caused by compensation movement patterns between the prime movers in the squat (quadriceps and GM). Also, Andrews et al. (1983) reported that when their subjects lifted BBS loads of 40%, 60% and 80% of their 4-RM, the kinematics changed with increasing load by increased hip flexion, therefore the resultant muscle torque at the knee did not increase in proportion to the load. It is good to mention that based on anecdotal evidence from both the laboratory and practice, 1 RM tests are seldom without clear movement pattern shifts. There should be more effort to test a more stable "technical maximum", which in turn should increase the validity and reliability of the submaximal sets.

A follow up study that used Bryanton et al. (2012) data was done by Vigotsky & Bryanton (2016) and showed based on a complex musculoskeletal model to what extent which hip extensors were contributing to the task at a given depth. Their results showed that AM seems to substantially contribute to hip extensor NJM in the BBS, producing, on average, more than 50 % of the net hip extensor NJM, especially in deeper depths and lighter loads (Figure 4). This most likely is largely influenced by the hip extensors internal moment arms. Although the GM is a prime mover in the squat, it has been reported that increasing depth past 90° does not allow the muscle to produce large amounts of torque despite being in stretch, due to that the internal moment arm is significantly smaller at high hip flexion angles (Delp et al. 1990, Escamilla et al. 2001b). This phenomenon can be seen from Figure 4. The hamstrings musculature also significantly increased their role in supporting hip extensor torque when load was increased at deeper depths. Based on the data in

Figure 4 it seems that the hamstrings musculature started contributing significantly more once the AM could not produce enough hip extensor torque. As stated before, these shifts to larger hip extensor NJM with increasing load probably have a lot to do with changes in kinematics. Therefore, it would be interesting to see how the results compare if a technical 1 RM was used for both the NBBS and WBBS. Data from Bryanton et al. (2012) and Vigotsky & Bryanton (2016) were taken from angles above 119° of knee flexion, which in a narrow position corresponds to slightly under femur parallel depth (Swinton et al. 2012, Cotter et al. 2013).

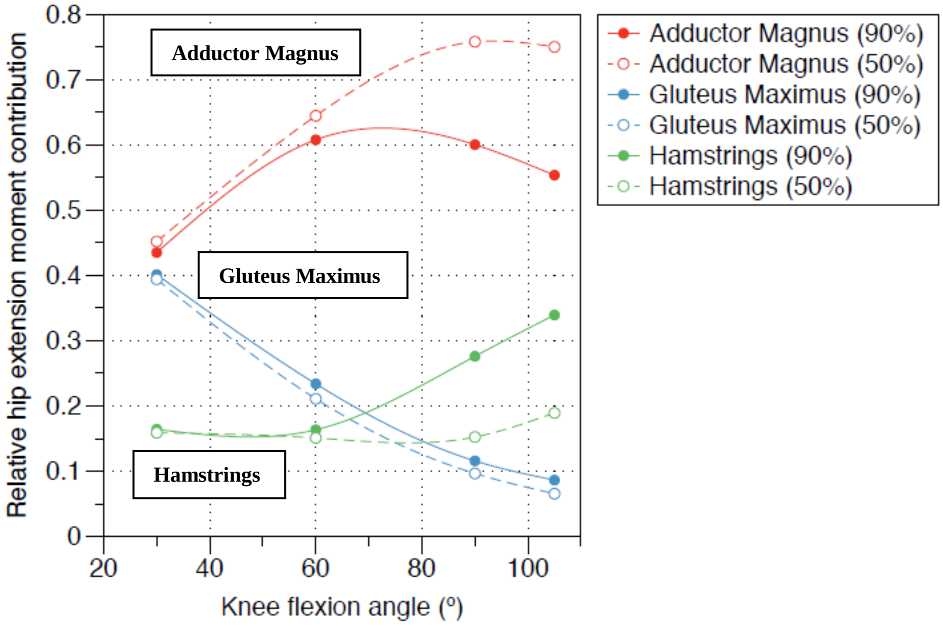


FIGURE 4. Relative muscle contribution to hip extension moment with respect to depth and barbell load (Vigotsky & Bryanton 2016).

Increasing depth from femur parallel increases hip extensor NJM in both the WBBS and NBBS (Wretenberg et al. 1996). Although Chiu et al. (2016) did not compare widths, they also showed that increasing depth in a NBBS from parallel depth to under parallel depth ("deep squatting") increased hip extensor NJM. Although this being the case, based on sEMG data increasing depth past parallel does not seem to further activate such hip extensors as BFLH and the GM to a significant extent when relative submaximal loads are used (Contreras et al. 2016). In terms of the GM this would make sense based on what was previously mentioned about the internal moment arms.

Swinton et al. (2012) also reported NJM for hip abduction and hip internal rotation, both of which were significantly higher for the WBBS compared to the NBBS (Table 1). The WBBS has been shown to activate the Adductor Longus (AL) significantly more than the NBBS (McCaw & Melrose. 1999. Pereira et al. 2010) but not the AM (Paoli et al. 2009). Paoli et al. (2009) study included a few methodological errors, including not normalizing any of the measured muscles, not mentioning where the electrodes were placed, and specific squatting depth was not mentioned in both sEMG and 1 RM testing. Also, whether external rotation could be matched in the NBBS to the same level as the WBBS if coached is unclear, but currently at least based on Swinton et al. (2012) data it seems that increased hip abduction consequently naturally increases space for more hip external rotation. This external rotation requirement seems to also cause more shifts between internal and external rotation during the WBBS, which is a detail that should be monitored. In conclusion, the hip-joint seems to be stimulated in all 3 planes of motion in the BBS, possibly more in the WBBS stance.

3.3. The hamstrings role in the squat

The hamstrings muscle group have been a subjected to a substantial amount of sport science research in the last decades. This can be attributed to multiple factors including their role in performance, their high rates of injury, and the challenges they present in quantifying their functional use via biomechanical research. From a neuromuscular standpoint, the hamstrings transmission of force has been both divided into intermuscular and intramuscular attributes, where both their timing of activation and absolute force production capability seem to be key components. From a structural and architectural standpoint, although potentially falling behind in physiological cross-sectional area (PCSA) compared to other prime movers such as the gluteals and quadriceps, their tendinous and variable pennate nature might potentially compensate well when mechanical peak power production is of concern (Woodley & Mercer. 2005, Hogervorst & Vereecke. 2015). This muscle design however seems to come at a cost. Because high forces at high speeds are distributed on the hamstrings in sport performance, particularly in sprinting, they also seem to be highly susceptible to injury (Opar et al. 2012, Edourard et al. 2016, Sugiura et al. 2017), which have been even reported to be increasing within the last decade (Ekstrand et al. 2014). Both acute and long-term studies have been completed in the effort to understand what strengthening exercises could potentially most optimally increase functional performance and lower the risk of injury in the hamstrings. In terms of acute studies, sEMG and functional Magnetic resonance imaging (fMRI) have been popular technological choices in quantifying hamstring utilization (Mendiguchia et al. 2013, Bourne et al. 2015, Schoenfeld et al. 2015, Bourne et al. 2016, Mendez-Villanueva et al. 2016). There has also been growing interest in understanding the different regional activity patterns of the hamstrings. This is due to that there is clear evidence of non-uniform distribution of recruitment patterns across the length of the hamstring, that can significantly change between movement task (Mendiguchia et al. 2013, Schoenfeld et al. 2015, Bourne et al. 2016, Mendez-Villanueva et al. 2016). Mendez-Villanueva et al. (2016) fMRI study compared exercises with substantial differences in hip and knee joint force utilization. The only hip isolated exercise “conic-pulley” showed higher T2 values in the proximal BFLH and ST region compared to the other movements (Nordic hamstring exercise, Russian belt deadlift, flywheel leg curl). Also, the conic-pulley showed unchanged in T2 readings for all other regions, emphasising the point that whether the chosen exercise is more or less hip dominant might have an effect on the regional activity, in this case specifically the proximal region. This improvement in research methods could have significant implications on exercises utilized for hamstrings injury prevention and rehabilitation.

Although lacking some attributes of fMRI, there are approaches to increase the validity of sEMG use. For example, traditional bipolar electrodes can be switched to high density electrodes (HD-sEMG), which cross a much larger surface area. These high-density electrodes provide multiple benefits, including observing accurately spatial distribution of recruitment patterns with a single electrode array (Stegeman et al. 2012). Also, due to the hamstring muscle boarders are so close to each other, avoiding cross-talk can be difficult when the aim is to distinguish activity between specific hamstrings muscles. This has led to many researchers using sEMG in hamstring studies to use the terms “lateral” and “medial” hamstrings instead of “biceps femoris long head” or “semitendinosus/semimembranosus” to avoid making type 1 error (Escamilla et al. 2001b, Schoenfeld et al. 2015). This quality leak to a large extent can be avoided using 2D-ultrasonography to find the hamstring muscles midlines, therefore hopefully providing more clarity in the results.

In terms of how the hamstrings behave in the squat, they seem to stay in a fairly isometric state during the squat due to being biarticular in nature and therefore more or less shortening at one end and lengthening at the other (Schoenfeld. 2010). This has been demonstrated indirectly with sEMG data, showing that hamstrings are only moderately active as hip extensors in both the WBBS and NBBS (Clark et al. 2012). It has been proposed that if the knee angle is restricted in some form in the BBS, posterior displacement of the hip (more hip flexion compared to knee flexion) could theoretically have some effect on the hamstrings force-length relationship (McCurdy et al. 2010), therefore effecting sEMG activity. This seems not to be the case based on studies that have compared a WBBS to a NBBS (McGaw et al, 1999, Escamilla et al. 2001b, Paoli et al. 2009), or studies that have compared increasing posterior displacement of the hips in the same width (Chiu et al. 2016).

Combined with 3-D kinematic data of the hip, measuring regional differences in sEMG activity may play a role in understanding how the hamstrings can behave in different squatting positions. Most BBS studies that have researched the hamstring musculatures sEMG activity have used the BFLH to represent the group (Clark et al. 2012, Aspe & Swinton. 2014, Chiu et al. 2016, Contreras et al. 2016. Slater & Hart. 2017). Based on the sEMG review provided by Clark et al. (2012) and most sEMG squat studies published after their paper to this day have followed SENIAM protocol (Aspe & Swinton. 2014, Chiu et al. 2016, Contreras et al. 2016. Slater & Hart. 2017), which is putting the electrodes on the medial portion of the hamstring musculature (Hermens et al. 1999). The middle belly has been chosen as a trusted site due to that it has typically the largest PCSA and also because it is not an innervation zone (Woodley & Mercer. 2005), therefore the chances of cross-talk and highly fluctuating sEMG amplitudes are minimized. As previously mentioned there are multiple studies that have shown regional differences in hamstrings activity behaviour (proximal vs. distal) depending on what type of movement is used; hip dominant or knee dominant (Mendiguchia et al. 2013, Schoenfeld et al. 2015, Villanueva et al. (2016). Therefore, it might be of value to further understand the role of the hamstrings musculature by exploring how the regional differences behave with shifting joint moments from the knee to the hip in the squat and vice versa.

3.3.1 The relationship between the quadriceps and the hamstrings muscle group in the barbell squat

The musculature that surrounds the knee further provides dynamic stabilization for the knee joint in quite a complex manner. These muscles include for the most part the quadriceps and the hamstrings group. During the squat, the primary muscles acting on the knee are the quadriceps muscles; vastus lateralis (VL), vastus medialis (VM), vastus intermedius (VI) and rectus femoris (RF), which carry out knee extension and resist knee flexion (Schoenfeld. 2010). The quadriceps are considered one of the prime movers in the squat and the squat is also considered one of the best exercises to activate and develop the quadriceps (Schoenfeld. 2010, Clark et al. 2012, Aspe & Swinton. 2014). The quadriceps muscle groups antagonist is the hamstrings muscle group. They form a complex relationship that helps stabilize the knee joint and as mentioned before; support optimal leg extension mechanics. Their relationship, and specifically their co-contraction has been of interest since the beginning of the 20th century when the famous researcher Warren Lombard wanted to understand why they co-contract during sit-to-stand motion (Lombard & Abbott. 1907), also known as Lombard's paradox (Gregor et al. 1985). Because muscles cannot develop different forces at different parts (i.e. a force produced by the hamstrings for hip extension will pull with equal force at the knee), the answer for the paradox of how a human can stand up sufficiently with the co-contraction had to lie somewhere else. The paradox was classically explained by noting the internal mechanical advantages at the hip or knee of the surrounding muscles. The Hamstrings musculature have a longer internal moment arm at the hip and a shorter one at the knee compared to the RF. Simplistically, this way although there is co-contraction present in a sit-to-stand movement pattern, the net moment should be an extensor moment at the hip and knee joint (Lombard & Abbott. 1907). In terms of knee stabilization, the co-contraction and strength balance between the quadriceps and the hamstrings seems to be quite essential for injury prevention at the knee (Kobayashi et al. 2010). The co-contraction seems to increase tibiofemoral and patellofemoral compression forces, which although possibly puts high demands on the meniscus depending on the load, it is speculated to be a protective function against shear forces on passive structures (Gullett et al. 2008. Slater & Hart. 2017). If strength levels are in balance (ratios differ in literature) and the muscles line of pull is put in an optimal angle, the posterior directed pull of the hamstrings musculature on the tibia help neutralize anterior shear forces (Escamilla et al. 2001a) and possibly even mediolateral shear forces (Palmieri-Smith et al. 2009, Slater & Hart. 2017) on the knee joint and thus alleviating stress from such structures as the ACL and MCL. This is logical due to that the hamstrings also assist external (BFLH) and internal (ST and SEM) rotation of the hip (Biel. 2010). For example, in open chain movements the sEMG activity of the BFLH can be increased by laterally rotating the tibia, while the activity of the ST and SEM can be increased by medially rotating the tibia (Mohamed et al. 2003, Jonasson et al. 2016). Unhealthy anterior and medial shear forces can be present in excessive anterior and medial knee movement relative to the foot (Escamilla et al. 2001c), which is possibly most evident when depth of a squat is forced to such an extent that the heel comes off the ground and/or signs of the tibia "collapsing in" causing pronation ankle mechanics to be more present (Bell et al. 2008, Toutoungi et al. 2000, Kim et al. 2015). Therefore, due to that the hamstrings musculature perform multiple tasks in dynamic movement, co-contraction or increased activation in specific rotated positions should be potentially seen as a more diverse phenomenon that varies in value in different situations. As mentioned before, these sensitive changes might better be picked up with more advanced technology.

3.4 The knee-joint

The knee-joint, that includes the tibiofemoral joint and the patellofemoral, is a synovial hinge joint formed between three bones: the femur, tibia, and patella. Its primary task is sagittal plane movement throughout a range of motion of 0 to approximately 160° of flexion. The knee-joint complex is considered a slightly modified hinge joint due to that it can complete a small amount of axial rotation during dynamic movement. This causes the instant centre of rotation at the knee to shift slightly throughout the squat (Schoenfeld. 2010). The knee joint also has an assistive joint called the patellofemoral joint, which encompasses the patella bone sliding over the surface of the femur during extension and flexion of the knee. The primary role of the patella is to improve quadriceps muscle group efficiency by increasing the angle of force application for the quadriceps tendon and therefore increasing the internal moment arm (Fox et al. 2012). The knee joints movement is also supported by a piece of cartilage named the meniscus and an array of ligaments, the most popular being the anterior cruciate ligament (ACL), medical collateral ligament (MCL), lateral collateral ligament (LCL), and posterior cruciate ligament (PCL).

3.3.1 Knee-joint kinematics

The kinematic requirements differ substantially between squatting styles. When comparing a parallel depth BBS at different widths, knee flexion angles have been reported to be around 120° in a NBBS (Bryanton et al. 2012, Swinton et al. 2012, Cotter et al. 2013) and in a WBBS around 110° (Escamilla et al. 2001b, Swinton et al. 2012). If anterior knee movement is not restricted, NBBS usually allows the lifter to reach something called “full depth” or a “deep squat” in a squat pattern (Chiu et al. 2016). This has been reportedly around 135° (Chiu et al. 2016)

3.3.3 Knee-joint internal kinetics

Although it is a positive phenomenon that the hamstrings support hip extension mechanics in the squat, as stated before the increasing demands of the hamstrings has direct implications on the quadriceps (Figure 5). The increased hip extensor NJM increases knee flexion NJM (Figure 5, green lines) and the quadriceps muscles, specifically the monoarticular quadriceps, have to counter this in order to produce sufficient net knee extensor NJM (Vigotsky & Bryanton 2016).

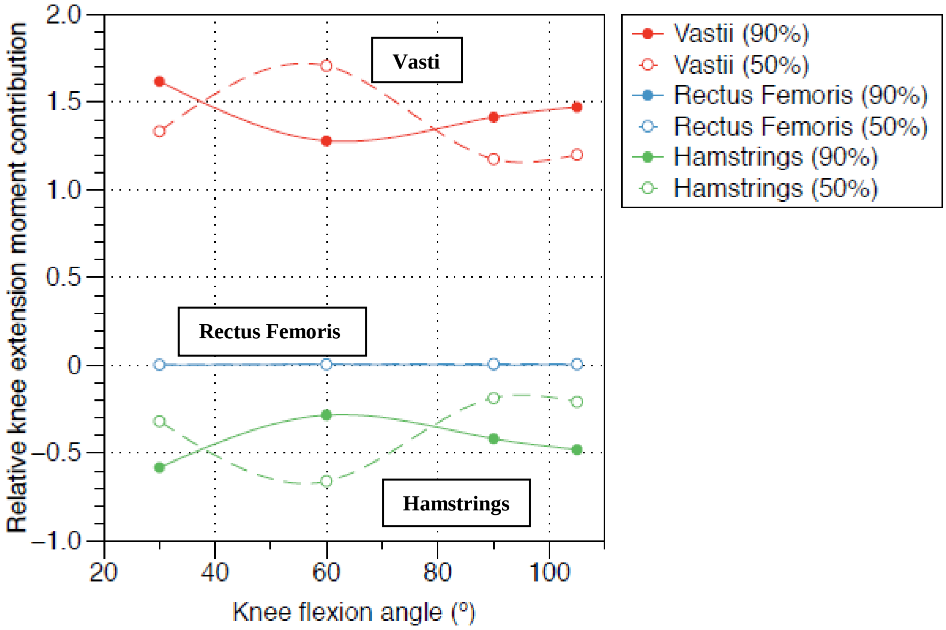


FIGURE 5. Relative muscle contribution to knee extension during squat with respect to depth and barbell load Vast represents the sum of the V.Lateralis, V.Medialis, V. Intermedis) (Vigotsky & Bryanton 2016).

However, this increased muscular effort of the quadriceps does not seem to be picked up by just measuring NJM. This is because NJM do not take into consideration co-contraction (Bryanton & Chiu 2014). This can be observed by comparing studies that have taken either NJM or sEMG or both. For example, it has been shown that knee extensor NJM can be reduced by manipulating the hips position more posteriorly in a NBBS (Wretenberg et al. 1996, Fry et al. 2003, Chiu et al. 2016). But Chiu et al. (2016) showed that sEMG activity at the quadriceps musculature stayed the same between the two conditions of NBBS even though there were significant shifts in knee NJM. This helps clarify the increased use of the hamstrings musculature in hip extension places more demands on the quadriceps musculature even though the NJM are reduced at the knee. It is good to clarify that even though initially one might assume that NJM at the knee behave similarly between an anterior knee movement restricted NBBS and a WBBS, this does not seem to be the case. They have been reported to stay fairly the same (Escamilla et al. 2001b), increase slightly in the NBBS (Swinton et al. 2012), or significantly (Wretenberg et al. 1996). Although not quantified in the same study, the sEMG of the quadriceps seems to behave the same in the WBBS as in the unrestricted or knee restricted NBBS position; it stays fairly the same (McCaw & Melrose 1999, Paoli et al. 2009, Chiu et al. 2016). There could be a case made for increased depth, where a WBBS movement pattern will not allow as much knee flexion as a NBBS (Wretenberg et al. 1996) and therefore activate the quadriceps musculature less. This though has yet to be properly substantiated and when NBBS to femur parallel depth has been compared to a deep NBBS with the same relative load there was no significant increase in the quadriceps sEMG activity (Hammond et al. 2016, Contreras et al. 2016). At this point it seems that to maximize quadriceps development, one needs to achieve a minimum depth of thigh parallel but not necessarily deeper. Whether that is via a NBBS or a more anterior knee movement restricted BBS does not seem to matter to a significant extent (Swinton et al. 2012, Chiu et al. 2016, Contreras et al. 2016, Vigotsky & Bryanton 2016). Therefore, when comparing the WBBS and NBBS, if sEMG was taken from the quadriceps and HD-sEMG activity was measured from the hamstrings while measuring hip and knee extensor NJM, there should be no significant change in quadriceps sEMG but potentially changes in the different regions of the hamstrings between the two conditions due to increased hip extensor demands and possibly a reduction in knee extensor NJM demands.

4 RESEARCH QUESTIONS

The goal of this thesis is to gain further insight into the biomechanical similarities and differences between properly standardised versions of the WBBS and NBBS in athletic populations and confirm previous observations. Specifically, the primary objective of this thesis is to explore the WBBS and NBBS under two relative loading conditions on overall and regional activity (HD-sEMG) in the hamstrings and how hip and knee NJM behave. The secondary objective is reporting lower lumbar (L5/SI) NJM and bipolar sEMG data from the GM and VL. This data will be taken to support the interpretation of the kinetic similarities and differences between the techniques.

Thesis questions and corresponding hypothesis:

1. Are there significant differences in overall hamstring HD-sEMG activity and different regional interactions in the WBBS and NBBS using relative loads?

There will be higher hamstring activity in the ascent phase in favour for the WBBS, mostly in the proximal region (Schoenfeld et al. 2015, Mendez-Villanueva et al. 2016).

2. Are there significant differences in measured 3-D plane moments (lower lumbar, hip and knee) and hip-to-knee moment ratios between the WBBS and NBBS at different loads?

The 3-D hip-to-knee moment ratio will be higher in the WBBS condition, due to higher 3-D hip moments and lower 3-D knee moments (Wretenberg et al. 1996, Swinton et al. 2012). The L5/SI region will be similar with a significant load interaction. All measured NJM will increase with load (Swinton et al. 2012).

3. Are there significant differences in gluteus maximus and vastus lateralis sEMG activity between the WBBS and NBBS using relative loads?

GM activity will be higher in the ascent phase of the WBBS, with VL activity only changing with loading condition (McGaw & Melrose 1999, Paoli et al. 2009).

5 METHODS

5.1 Subjects

All subjects were recruited from the Jyväskylä Rugby Club. In total 14 amateur rugby players (6 males, 8 females, mean ± SD, age 27.36 ± 3.71 years; height: 174.1 ± 10.4 cm, body mass 81 ± 21.86 kg, squatting experience: 3.93 ± 1.77 years) were recruited to the study. 6 of the subjects were a part of Finland’s national rugby team (for complete information on subjects see appendix D). Only athletes with a minimum of 1 year of active BBS experience, a WBBS or NBBS 1 repetition maximum to body mass ratio of at least 1.0, no health concerns and who completed all the required familiarization sessions could participate in the measurements. Written informed consent was obtained from all subjects on the first day of familiarization and approval was granted from the University of Jyväskylä Ethical Committee and was performed in the accordance with the Declaration of Helsinki (Appendix A).

5.2 Study design

A cross-sectional, repeated measures design was used to compare kinematic and kinetic performance measures of different versions of the BBS. This thesis had a primary focus of examining specific biomechanical characteristics between the WBBS and the NBBS within specific technical boundaries with both 70% and 85% of 1 RM loads. All subjects were familiar with BBS to parallel depth and completed 3 weeks of familiarization with all 4 squatting conditions. The fourth week was devoted to 1 RM testing for both the WBBS and NBBS squat on two separate days in a randomized order. Data was collected on week 5 with 1 testing session 5-7 days after the final 1 RM test (figure 6).

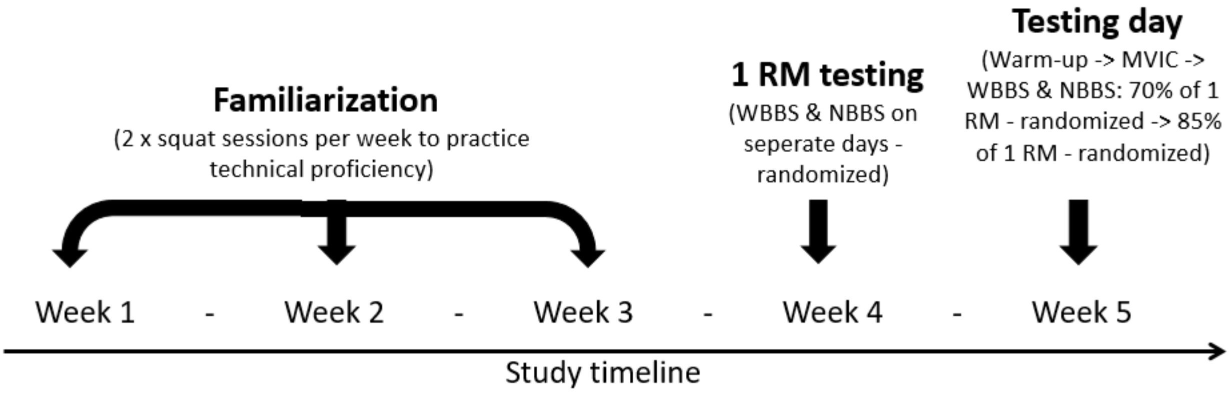


FIGURE 6. 5-week study timeline.

5.3 Familiarization

All familiarization sessions were conducted at the University of Jyväskylä neuromuscular research centres gym. Familiarization was in total 3 weeks including 6 sessions (2 per week) in total required to participate in the testing. The initial 18 participants were divided into groups of 4-5, who would train together for the rest of the familiarization. Out of the 18 subjects, there were 4 dropouts before testing (3 male, 1 female). Reasons included injury sustained in rugby practice and timetable issues. The sets and reps were kept similar the entire familiarization phase with a high focus on technique and a lower focus on overload. In the first week of familiarization sets could be increased and reached a total of 8-10. In week 2 and 3 most subjects started reaching basic technical proficiency and therefore weight could be increased and sets could be reduced to a total of 5-6. The subjects that were less familiar with one form of squatting where allowed 1-2 sets extra with either the

WBBS or NBBS. Repetitions were kept at 4-6 range depending on the weight used. The WBBS and the NBBS had a couple of specific details in common, and that was depth, tempo, bar positioning and footwear. We proposed that femur parallel depth would be a practical depth level to standardize due to that a.) the depth is commonly used in practice for multiple purposes b.) In terms of the hips external moment arm, peak distance is reached when the femur is horizontal, c.) Even though there would be clear differences between individual hip structures, femur parallel depth would a realistic to expect everyone to reach, therefore reducing the chance of dropouts d.) because this study was focused on the hamstrings, it was avoided to go to deeper depths where visual PPT could be observed as previously mentioned, which could significantly affect the internal moment arm of the hamstrings.

In the first familiarization session subjects were explained to standardize warm up for all familiarisation sessions and testing. The standardized protocol included 5 minutes on an ergo bike (Teambike, PRECOR, USA) followed by 5 minutes of dynamic warm-up used in their team practice. Also on the first session, subjects were screened by performing their current technique of BBS used in their current training program with a dowel while being filmed. There was a clear observed variation in individual squatting widths but none of the subjects were familiar with squatting in the wide position that was required in this study. Therefore, proper familiarization became essential to minimize error in the study. Because squatting mechanics would be measured without shoes to avoid any effects on movement patterns, familiarization was also completed without shoes with an exception made for minimalist shoes (figure 11).

Following the screening performance subjects were explained the kinematic positions sought after for analysis from the WBBS and NBBS and related to how they were currently moving. Before loading the squats, both the WBBS and NBBS were practiced with bodyweight. Wide squat positions were practiced with a wall drill. The wall was used as a coaching tool so that subjects could practice posterior displacement of the hips while keeping a trunk angle preferably around 50 degrees, similar or slightly higher to previously reported literature (Escamilla et al. 2001b, Hales et al. 2009). Width was increased until subjects could comfortably shift their weight towards their heels and achieve close to vertical shin positioning without falling backwards. External rotation of the feet was coached to be around 20-40 degrees, similar to previously reported literature (Paoli et al. 2009). External rotation was further increased or decreased based on observed individual range of motion patterns and communication with the subject. In general, subjects felt comfortable to reach femur parallel depth with around 30 degrees of foot external rotation with a width approximately 1.5 (1,52±0,07) of the distance between the greater trochanters (GT) after the dynamic warm up (figure 7, B). GT width was measured with measuring tape with the subject laying supine. Fingers were pushed into the skin so that they clearly were in contact with the GT.

The narrow squat position was practiced highly based on the recommendations of the National Strength and Conditioning Association (Myer et al. 2014). Exceptions included a slightly narrower stance by standing in GT width (0.99±0,04) instead of shoulder width (figure 7, A). Anterior knee displacement was promoted but restricted to the extent that the centre of pressure stayed around midfoot at parallel depth. This meant that there was some variation in how far the knees travelled in the anterior direction but in average the knees stayed very close to the toe line (Figure 11). In general, external rotation the toes were kept in the NBBS around 10-20 degrees, therefore only slightly less than the WBBS (Figure 7). For a couple of subjects the narrow width restricted them from comfortably reaching femur parallel. As with the WBBS, this was fixed successfully by exploring higher ranges of external rotation; around 30-40 degrees. Posterior hip displacement was still dominantly present in the narrow squat compared to the amount anterior knee displacement (figure 11), therefore creating a similar trunk lean as the WBBS.

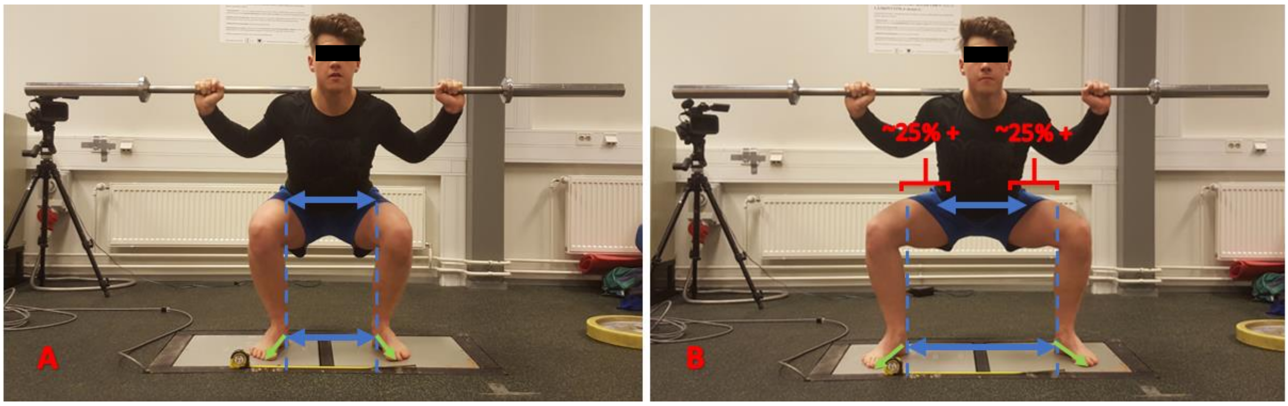


FIGURE 7. Frontal view of a typical width ratio and foot external rotation position in the study. Stance width was measured from heel to heel and compared to GT width. NBBS (A) was around GT width (0.99 ± 0,04) and WBBS (B) was around 150% of GT width (1.52 ± 0,07).

Once comfortable squatting widths had been established with bodyweight, the movement patterns could be loaded with the barbell. The barbell was placed at the same location for both NBBS and WBBS in the effort to increase biomechanical similarities. Specifically, the barbell was placed on the top of the posterior deltoids similar to previous literature (Hatfield et al. 1981, Fry et al. 1993), which as stated earlier, could be considered a position between a high-bar squat and a low-bar squat (Goodin. 2015), or as stated earlier; a mid-bar position. A high bar position was avoided to increase the ease of posterior hip displacement and a low bar position was avoided to increase the similarities of hip flexion and trunk lean for both NBBS and WBBS conditions. Loading the athlete also further increased their capability to displace their hips posteriorly, especially in the WBBS, because the barbell functions as a counter weight. At the end of the first session, squat widths for both wide and narrow positions were taken by measuring the distance between left and right legs medial boarder of the calcaneus (Appendix D). These distances were used in every preceding familiarization session and in the testing session.

It was essential to standardize cueing as much as possible. The subjects were first presented with both internal and external cueing and towards the end of the familiarization cueing was kept entirely external (Table 2). This was done to avoid the potential significant influence of internal cueing on sEMG activity in the testing session reported previously in literature (Wulf et al. 2010). The external cues used in familiarisation were also used on testing day. Internal cues were more individualized than the external cues based on individual movement pattern issues determined by the practitioner.

TABLE 2. Coaching cues used in study

Cueing	**All squats**	**NBBS**	**WBBS**
Internal focus	“Externally rotate the legs without the big toe leaving the ground or the toes scraping the ground” “Tilt the pelvis up/down (depending on the issue)” “Fill your entire trunk with air like a balloon” “Pull the barbell down towards the hip”	“Initiate movement from both the knees and the hips” “Keep your weight midfoot/towards the heel” “Let the knees travel freely forward while maintaining a tall posture”	“Only move via the hip” “Push the hips back and down while keeping a tall posture” “Weight on the heels” “Feel the tension in the posterior muscles and inner thighs by sitting on them and using them as a sling”
External focus	“Screw the legs into the ground” “Gaze towards something slightly below your line of sight” “Brace your trunk as you would when taking a punch (not effecting posture)” “Push the ground down”		

Descent, amortization, and ascent phase tempo was also practiced in the familiarization sessions. This was introduced on week 2 to avoid overloading the subject with information. A tempo of 3-0-XX was used, where “3” is the seconds of the descent phase, “0” represents the amortization phase and, “XX” represents the ascent phase. “XX” means that the phase had to be as fast as possible, while maintaining form. Although the amortization phase was kept at 0 seconds, we did not want to observe any bouncing, therefore the subject was told to stop quickly for ~0.5 seconds and then initiate the ascent phase. A tempo was played to the subjects via a metronome application (Pro Metronome, EUMLab, Xanin Technology, Germany). Depth to femur parallel was visually controlled via the verbal feedback of the experienced practitioner. In the WBBS, forward knee movement was either controlled by oral feedback or sometimes with the help of such tools as a dowel placed in front of the knees.

Subjects were also familiarized to the positions and contractions required for hamstring and GM normalization on the last week of squat technique training. For the hamstrings, this was done by completing 3 maximal isometric contractions for both knee flexion and combined hip extension and knee flexion in the dynamometer used on measurement day. For the GM, this was done by practicing the standing “glute squeeze” MVIC task 3 times (Contreras et al. 2015). There was no MVIC test for the quadriceps, but rather they were normalized on testing day by comparing to the peak mean sEMG value of each subject.

5.4 1-RM testing

After 3 weeks of familiarization, two extra sessions were devoted to test 1 repetition maximum in both the WBBS and NBBS. The 1 RM test order was randomized for all subjects. 1 RM testing was done for both the WBBS and NBBS, due to that based on anecdotal evidence they can be significantly different from each other. The 1 RM protocol followed to a large extent a procedure described by Kreamer and Fry (1995), which seems to be a common procedure in acute squat studies (Yavuz et al. 2015, Chiu et al. 2016). Specifically, after completing the same general warm up that was used in familiarization, subjects completed an incremental loading protocol of around 4-6 sets before reaching their 1 RM. The first set was completed by performing multiple repetitions (4-6) with the barbell, then after a short break an in equal amount of repetitions with a load assumed to be approximately 50% of 1 RM was completed. Following this, repetitions were significantly reduced to around 2-3 and loads were increased with about 15-25% (depending on the level of the athlete) for the next two sets. All sets above 50% of 1 RM had between set breaks of 3-4 minutes. The goal was to achieve around a 90 % of 1 RM mark by the fourth set for most of the subjects. Based on the subjects RPE score for the estimated 90% of 1 RM weight set a realistic estimation could be made of what might be a technical 1 RM load. After this consecutive 1 RM trials were made until any unwanted technical alterations were visual, such as;

- A change in the synchronization of hip and knee movement in the ascent phase. This is typically observed by observing that the movement is clearly initiated at the knees before the hips, therefore the hip is pushed up and the trunk starts to lean forward.
- Clear valgus collapse, where the patella is clearly not tracking the toes.
- Any clear deviations in the spine
- Centre of pressure (COP) shifts. For example, in the wide squat, when the shin angle was clearly moving away from a vertical position at parallel depth, shifting the COP towards the midfoot. Similarly, in the narrow squat, when COP could move too far forward, but now it was from the midfoot towards the forefoot, usually visual by the heel coming slightly off the ground.

Because all subjects could probably lift significantly more weight without these restrictions the 1 RM testing referred to as “technical 1 RM testing”.

5.5 Testing day summary

Subjects arrived to the neuromuscular research centre in Jyväskylä where all data was collected. HD-sEMG electrodes were placed on the ST and BFLH. Bipolar sEMG electrodes were placed on the VL and GM. Electrode placement was followed by normalization for the hamstrings using MVIC tests. 20 markers were placed on the lower and upper body for motion analysis. A 10-minute warm up was completed before commencing measurements of 6 different back squat conditions (WBBS + NBBS at 70 + 85% loads, WBBSF and NBBSF at 70% load) on two force plates surrounded by 7 motion capture cameras in randomized order.

5.5.1 Surface electromyography

Surface electromyography (sEMG) electrode placement protocols were initiated by placing all electrodes on the dominant leg (all subjects were reportedly right leg dominant). Hamstring muscle borders were marked with the help of B-mode 2-D ultrasonography (Aloka α10, Tokyo, Japan) (Figure 8). Following this, markings were made on the location recommended by SENIAM for bipolar electrode placement on both hamstrings. Specifically, SENIAM recommends for the BFLH that the bipolar electrodes are placed 50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia. For the ST, the recommendation is 50% on the line between the ischial tuberosity and the medial epicondyle of the tibia (Hermens. 2012).

FIGURE 8. Finding the midline of the hamstrings with the help of 2-D ultrasonography.

GM and VL bipolar locations were also marked. Specifically, markings on the GM was placed 50% on the line between the sacral vertebrae and the greater trochanter in accordance with SENIAM. For the VL, markings were placed 2/3 on the line from the anterior spina iliaca superior to the lateral side of the patella in accordance with SENIAM. Before electrode placement, skin adhesion and impedance was improved with shaving the skin with a razor, followed by light treatment with sand paper and an alcohol swab.

FIGURE 9. HD-sEMG electrodes placed on the midline of BFLH and ST with the help of 2-D ultrasonography.

After skin preparation, 16-channel semi-disposable HD-sEMG arrays (ELSCHO16, OT Bioelettronica, Torino, Italy, 10 mm inter-electrode distance) were attached along the midline between muscle borders of BFLH and the ST using an adhesive foam, which was connected to the amplifier of the EMG system (EMG-USB, OT Bioelettronica). The high-density electrode consisted of 15 electrode pairs and 1 summoning pair. For all subjects, the electrode was placed with the effort to put the middle of the array (electrode pair 8) as close as possible to the location advised by SENIAM (Hermens et al. 1999), which was consistently either at or close to mid belly of the muscles. Due to individual muscle length differences and active tissue boarders, minor variations were present in electrode placement. For ST, array was attached below the proximal tendinous inscription (Woodley & Mercer 2005) of the muscle defined with 2-D ultrasonography. The SENIAM location was quite close to the centre of the measured medial region of both hamstrings with only slight variation. For all subject’s electrodes ended up being placed so that 5-7 electrode pairs were located proximal and 7-9 distal to the SENIAM area. Following electrode placement, the cavities of the electrode arrays were filled with 20 µl conducting gel for proper electrode-skin contact (Signa gel electrode gel, Parker Laboratories, New Jersey, USA). Following this electrode were secured to the skin with tape (Leukoplast, BSN Medical, Hamburg, Germany) (Figure 9). Reference electrode for the high-density array electrode system was placed over the wrist. Following hamstrings preparation, 2 circular pregelled electrodes with an electrode diameter of 95 mm (Ambu Blue Sensors N-10-A, Medicotest, Olstykke, Denmark) were placed with 20 mm interelectrode spacing on the right gluteus maximus and the right vastus lateralis with an effort to put the electrodes parallel to the orientation of the fibers. After placing electrodes on, Low impedance was further verified with an Ohm meter, measuring the Ohm – resistance between the electrode pair. Under 5 KOhm was considered acceptable (Konrad. 2006).

After preparation, signal quality was checked for the hamstrings, VL and GM with prone submaximal isometric knee flexion, knee extension and glute squeeze contractions. High density EMG data were collected at 2048 Hz, amplified by a factor of 1000 and converted to digital signal (EMG-USB 12-bit analog-to digital converter, OT Bioelectronica). 15 differential signals were recorded from each muscle during the tasks using the BioLab software (v. 3.1, OT Bioelectronica) with 10 mm interelectrode distance to minimise cross-talk (De Luca et al. 2012).

Bipolar sEMG data was collected at 10-1000 Hz, amplified by a factor of 1000 (model 16 - 2, EISA, Freiburg, Germany), converted to digital signal using a 32-bit A/D converter with a ± 2.5 V range (Cambridge Electronic Design, Cambridge, UK), and processed in Spike (Spike2, Cambridge Electronic Design, Cambridge, UK).

5.5.2 MVIC for hamstrings

For normalising EMG signals for the hamstrings, subjects performed maximal knee flexion and combined knee flexion and hip extension isometric contractions (MVICs) after specific warm-up including ten submaximal contractions with increasing intensity (from ~30 to ~90%). MVICs included holding a maximal contraction for 3 seconds a total of 3 times, separated by 1 minute rest. MVICs were performed in a custom-made dynamometer (UniDrive, University of Jyväskylä). Specifically, the hamstrings were isolated with the subject laying in a prone position with the dominant leg (right leg for all subjects) bent from at ~ 20° knee flexion in accordance with previous literature (Figure 10) (Konrad. 2006). The measured leg was attached by the ankle to the force transducer in form of an ankle brace placed 2 cm above the lateral malleolus. The hip was fastened securely with a belt so that hip flexion would be avoided in the unilateral contractions. Force data was measured with the bi-axial force transducer of the dynamometer and collected at 1000 Hz that was digitised using a 32-bit A/D converter (Cambridge Electronic Design, Cambridge, UK).

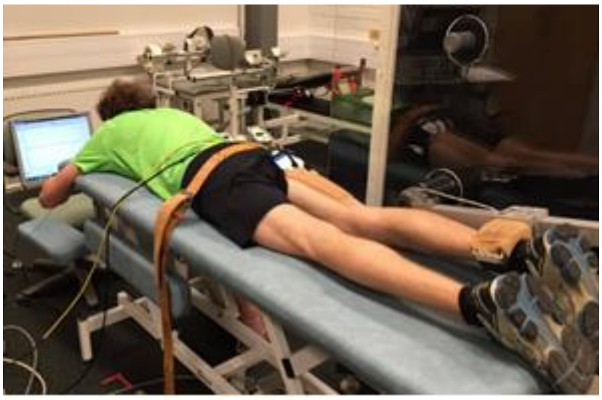


FIGURE 10. MVIC test position for hamstrings. Tests in included knee flexion and combined knee flexion and hip extension.

The data was visualized in real-time and recorded using Spike2 software (Cambridge Electronic Design). Force data was synchronized with sEMG data by sending a synchronization pulse from the Spike2 to the EMG software.

5.5.3 Normalization for gluteus maximus

The gluteus maximus muscle was normalized by a standing MVIC glute squeeze task previously used in literature (Contreras et al. 2015). This was done post warm-up by asking the subject to squeeze the gluteals while externally rotating the femurs as hard as possible for 3 seconds, which was repeated a total of 3 times with a 1 minute break.

5.5.4 Kinematics and kinetics

Before performing the warm-up for the squats, 14 mm diameter reflective markers were secured in the following locations, 4 cm above the C7 vertebrae (due to the barbell being so close to C7), at T10, the jugular notch, xiphoid process of the sternum, over the anterior and posterior superior iliac spine, lateral thigh, lateral epicondyle of the femur, lateral shank, lateral malleolus, calcaneus and second metatarsal head of each side following the full body Plug-in Gait Model in the Nexus Software (Vicon Motion Systems Inc., Oxford, UK), with excluding the arms. Further, anthropometric data was collected for the Nexus software. This included measuring ankle width, knee width, leg length and height. To determine three-dimensional (3-D) external force, L5/S1, hip and knee NJM and kinematics, 3-D marker displacements were recorded with 7-camera Vicon motion analysis system at 250 Hz sampling frequency (Vicon Motion Systems Inc., Oxford, UK) and 2 force plates (AMTI, Watertown, MA, USA) at a 1000 Hz sampling frequency using Nexus software. The origin of the global axes was set to the corner of the force plates. The X, Y, and Z axes were set to medial-lateral, anterior-posterior, and vertical directions, respectively.

5.5.5 Squat protocol

Before the measured squats were initiated, a 10-minute warm up was completed that included 5 minutes on a ergo bike, light dynamic stretches and warm up sets with 30 and 50% of 1 RM for both NBBS and WBBS. In total, 4 different back squatting conditions were measured in randomized order. Each condition had to include two technically accepted repetitions for analysis. Repetitions in a set were done one at a time with a intraset break of 30 seconds. Interset breaks were kept at 2-3 minutes. Tempo and depth was controlled according to the familiarization protocol via oral feedback from the practitioner.

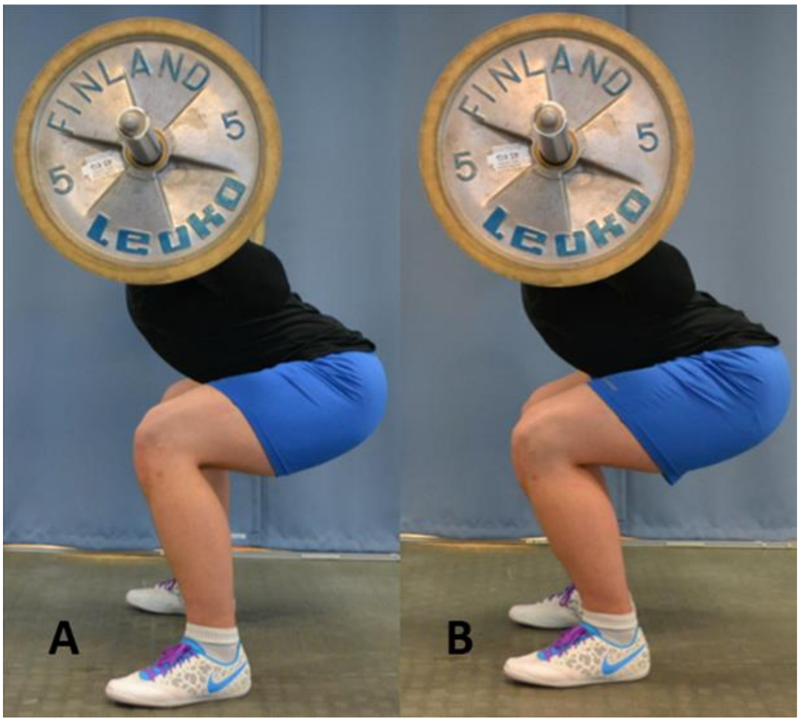


FIGURE 11. Side view of the WBBS and NBBS. In the pictures, we can see the standardized squatting femur parallel depth for both the WBBS (A) and the NBBS (B).

The squatting width that was determined in familiarization was used in testing by putting tape markers next to the force plate so the subject knew where to place their feet. There was a total of 4 conditions measured; WBBS and NBBS with 70% and 85% of technical 1 RM. First, repetitions were analysed for the WBBS and NBBS with a 70% of technical 1 RM load in randomized order. Following the first 2 conditions, the 85% of technical 1 RM load was completed for the NBBS and WBBS, also in randomized order.

5.6 Data analysis

sEMG activity from the right leg was determined for the ascent phase of each task. The right legs knee and the hip angle were used to determine the initiation of the squat, the start of the ascent phase, and the end of the movement. Hip kinematics in all 3 planes, knee flexion and ankle flexion were calculated in Nexus software based on the Plug-in Gait Model after smoothing marker trajectories with an 8 Hz low-pass Butterworth filter. Both array and bipolar electrode sEMG data were band-pass filtered using a 20-500 Hz fourth-order zero-phase Butterworth filter in Matlab (MathWorks Inc, Natick, MA, US). In MVICs for the hamstrings, sEMG data on a 1-second stable plateau was root-mean-squared (RMS) for each channel of the HD-sEMG array electrode. Highest RMS activity across MVIC tasks and repetitions for each channel was used to normalize hamstring sEMG activity for each squat repetition. Also for the GM, highest RMS activity across MVIC repetitions was used to normalise sEMG activity for each squat repetition. For the VL, the squat repetition out of the two recorded repetitions reaching the highest mean value in the ascent phase across both squat types for each subject was used as an individual normalization value. For each squat squat type, sEMG activity from all muscles were averaged for the 2 repetitions. For the hamstrings, EMG activity was expressed as a percentage of the highest mean RMS activity of the corresponding channel in MVIC (% of MVIC). Further, channels 1-5, 6-10, and 11-15 were averaged and assigned as activity in the distal, middle and proximal regions, respectively. Noisy channels were removed and if one channel was removed from one squat type then it was removed from all the other squat types for the same subject. Averaging sEMG activity from all 15 channels and the 5 channels for each region minimised the effects of muscle movement under the skin on the defined regional sEMG activity. For the GM and VL, sEMG activity was expressed as a percentage of the highest mean RMS in the reference contraction (%).

NJM were calculated by inverse-dynamics calculations in the Nexus software based on the full body Plug-in Gait model, using subject’s anthropometric data, GRF data and kinematic data. The NJM calculated in this study are expressed as the internal (muscles) net moments with respect to distal segment local coordinate system. Specifically, L5/SI, hip and knee NJM in all 3 planes, 3-D external forces in all 3 planes, and anterior-posterior centre of pressure (COP) data were analysed further after exporting all kinetic and kinematic data from the Nexus software after smoothing force plate data with an 8 Hz low-pass Butterworth filter. All kinematic and kinetic data from the force plates was exported to- and analysed in Microsoft Excel. Reported joint kinematics, NJM and external force data were summoned between legs and averaged between repetitions. NJM from all biomechanical planes were normalized to the subject’s body weight and expressed as Nm/kg. Following this, peak NJM were found for each plane for L5/SI, hip and knee. 3-D NJM was calculated by summoning the sagittal, frontal, and transverse plane NJM in each data frame (250 Hz) of each squat repetition using the following equation:

$$\sqrt{((sagittal\ NJM)^2 + (frontal\ NJM)^2 + \ (transverse\ NJM)^2)}$$

Following this, peak 3-D NJM were found for each repetition, where the cell with the highest resultant sum value was derived from the data set. External force data was presented by formatting the peak force value from the vertical, anterior-posterior, and medial-lateral plane into peak relative % contributions. COP was presented in the anterior-posterior plane in form of a cm/time axis. All charts were interpolated to a 0-100% format.

5.7 Statistical analysis

Intratrial reliability for each variable analyzed was assessed by intraclass correlation coefficient (ICC) and cross-validation (CV) using Hopkins (2015) spreadsheet. The data was checked to be normally distributed by using Shapiro-Wilk’s test of normality. Interactions between regions of the hamstrings between the WBBS and NBBS were analyzed with a 2x3 (squat type x region) ANOVA. In case of biased sphericity determined by Mauchly’s test, Greenhouse-Geisser correction was applied. Pairwise comparisons were conducted with Bonferroni correction. Potential differences in all measured kinematic and kinetic variables between WBBS and NBBS and between loads were analyzed using a paired samples t-test, where each load condition was compared separately between widths and between the same squat condition. In the effort to establish more practical significance to the results, effect size (ES) was run on all variables allowing interpretation our data against Cohen's benchmarks to assign small (>0.02), medium (>0.05) and large (>0.08) effects (Cohen. 1988). Descriptive data are mean ± standard deviation unless otherwise stated.

Alpha was set at $p<0.05$, and was further divided into very significant ($p<0.01$) and highly significant ($p<0.001$).

6 RESULTS

Out of 14 of the measured subjects, only 10 subjects' data could be used for NJM and kinematic analysis, 7 for bipolar sEMG and 8 for BLFH and 5 for ST HD-sEMG due to equipment malfunctions. Specifically, 4 subjects NJM data was not reliable due to force plate calibration issues, sEMG for 7 subjects was removed due to our sEMG device had to be replaced mid study due to malfunction and it was decided not to trust its produced data from the first 7 subjects, and a large part of the HD-sEMG array electrodes that were used were malfunctioning, even more so for ST. Subject data can be found in table 3, more descriptive subject data can be found in APPENDIX D.

TABLE 3. Subject characteristics after width and 1 RM testing. Both the wide and narrow conditions stance widths were divided by greater trochanter width (WBBS/NBBS/GT) and 1 RM testing to parallel depth was related to the subject's bodyweight (kg/kg). Specific weights are shown in APPENDIX D.

Subject	**Gender**	**Age**	**Height (cm)**	**Weight (kg)**	**Lifting Experience (years)**	**WBBS/GT width**	**NBBS/GT width**	**WBBS (kg/kg)**	**NBBS (kg/kg)**
1.	Female	26	159	53	4	1,5	0,97	1,46	1,56
2. (d)	Female	26	164	65	2	1,47	0,97	1,04	1
3. (b, c)	Female	25	178	68	3	1,52	0,97	1,29	1,25
4. (b-d)	Female	28	164	58	4	1,46	0,92	1,29	1,38
5. (a-d)	Female	27	170	64	5	1,5	0,97	1,6	1,6
6.	Female	28	167	70	4	1,5	0,94	1,43	1,5
7 (c, d)	Female	24	164	63	2	1,57	1,03	1,11	1,19
8 (a-c)	Female	24	166	67	1	1,47	1	1,01	1,01
9 (c, d)	Male	26	190	108	5	1,53	1	1,39	1,36
10 (a, c)	Male	31	178	107	7	1,59	1,03	1,45	1,59
11.	Male	35	184	102	4	1,5	0,97	1,2	1,23

12. (b-d)	Male	22	183	93	2	1,41	0,97	1,51	1,4
13. (d)	Male	27	184	115	6	1,62	1	1,37	1,28
14. (a-c)	Male	34	188	101	6	1,69	1,06	1,44	1,34
AVR		**27,36**	**174,18**	**81**	**3,93**	**1,52**	**0,99**	**1,33**	**1,33**
SD		**3,71**	**10,43**	**21,86**	**1,77**	**0,07**	**0,04**	**0,18**	**0,19**
🟩	= All data		🟨	= Missing data. Missing moment data (a), Missing BFLH (b), Missing ST (c), Missing bipolar EMG (d).					

All variables showed normality between subjects. For the 2 repetitions averaged for each squat condition, intratrial reliability (ICC) ranged from fair to excellent between all variables (Less than 0.40—poor, between 0.40 and 0.59—fair, between 0,60 and 0,74—Good. Between 0,75 and 1,00—Excellent). Specifically, all kinematics variables were >0,68, except for descent phase that ranged from 0,41 – 0,91 (Fair – Excellent). All NJM variables were >0,90 except for L5/SI frontal and transverse NJM that ranges between 0.42 – 0.85 (Fair – Excellent). All bipolar sEMG values ranged from 0,40 – 0,99 (Fair to Excellent). Detailed results can be found in appendix D, table 9.

6.1 Kinematics

All kinematic results are presented in APPENDIX D table 3 and data is visualized in figure 12. The WBBS reached moderate to large ES for higher hip flexion angles in the 85% load condition ($p<0.05$), abduction angles at both loads ($p<0.01$), and hip internal rotation angles at both loads ($p<0.001$). Across all loads, the NBBS reached a large ES for higher knee flexion angles and dorsiflexion angles ($p<0.0001$). A large ES was found in higher loads for the ascent phase, with a longer ascent phase for heavier loads ($p<0.05$). No statistical significance was found in the ascent phase length across all conditions ($p>0.05$).

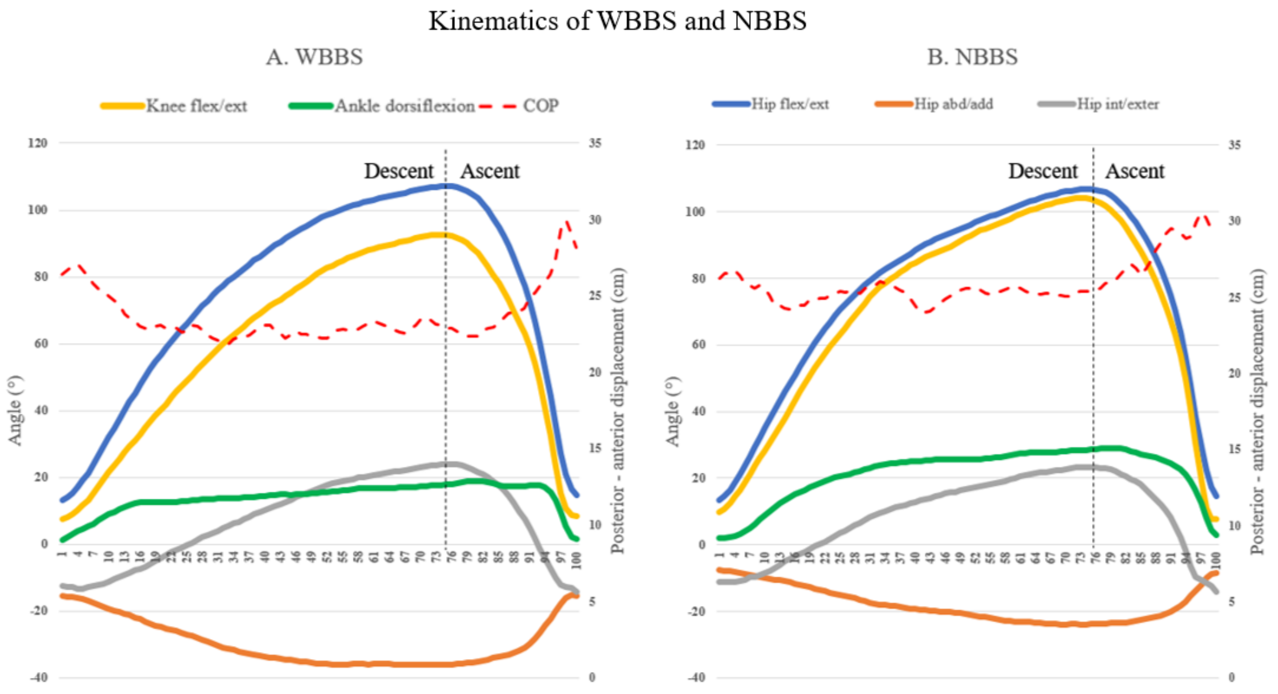


FIGURE 12. Kinematics of the WBBS (A) and NBBS (B) to parallel depth. Interpolated from 70% of 1 RM data. 0% is the start of the decent phase and 100 % is the end of the ascent phase. Dashed line represents the start of the ascent phase. COP posterior-anterior displacement centimeters are arbitrary units. We can observe from the COP that the subjects weight in the beginning on the squat shifts slightly more posteriorly in the WBBS due to larger hip displacement.

6.2 Net joint moments

All NJM results are presented in APPENDIX D table 3 and 7. NJM are present in form of bar charts in tables 13-17 and time interaction in figure 19 and 20. Out of the 12 peak NJM variables (not including the hip-to-knee ratios) 7 reached a moderate to large ES in statistically significant differences between NBBS and WBBS ($p<0.05$). 5 of these variables were higher in both loading conditions for NBBS and WBBS, strengthening the result. These 5 variables included 4 variables that were higher in the WBBS: 3-D hip NJM, hip sagittal NJM, Hip transverse NJM, knee sagittal NJM, and knee frontal NJM ($p<0.05$) and one variable that was higher in the NBBS: knee sagittal NJM ($p<0.05$). At 70% of 1 RM, hip frontal plane NJM was significantly higher in the WBBS ($p<0.05$). At 85% of 1 RM, L5/SI NJM and 3-D knee NJM were significantly higher in NBBS ($p<0.05$).

In terms of effects of load within the same squat width, 8 out of 12 variables reached a moderate to large ES ($p<0.05$). Of these 8 variables, 4 were higher in both WBBS and NBBS load comparisons, further strengthening the result. These 4 variables that were higher in both WBBS and NBBS comparisons included higher 3D L5/SI NJM (p<0.05), 3D hip NJM (p<0.01), hip sagittal NJM (p<0.01), and 3-D knee NJM (p<0.01), all reaching large ES. The other 4 variables that were higher were L5/SI sagittal NJM, hip frontal NJM, knee frontal NJM, and knee transverse NJM in the NBBS load comparison (p<0.01) and higher knee sagittal NJM in the WBBS load comparison (p<0.05), all reaching moderate to large ES.

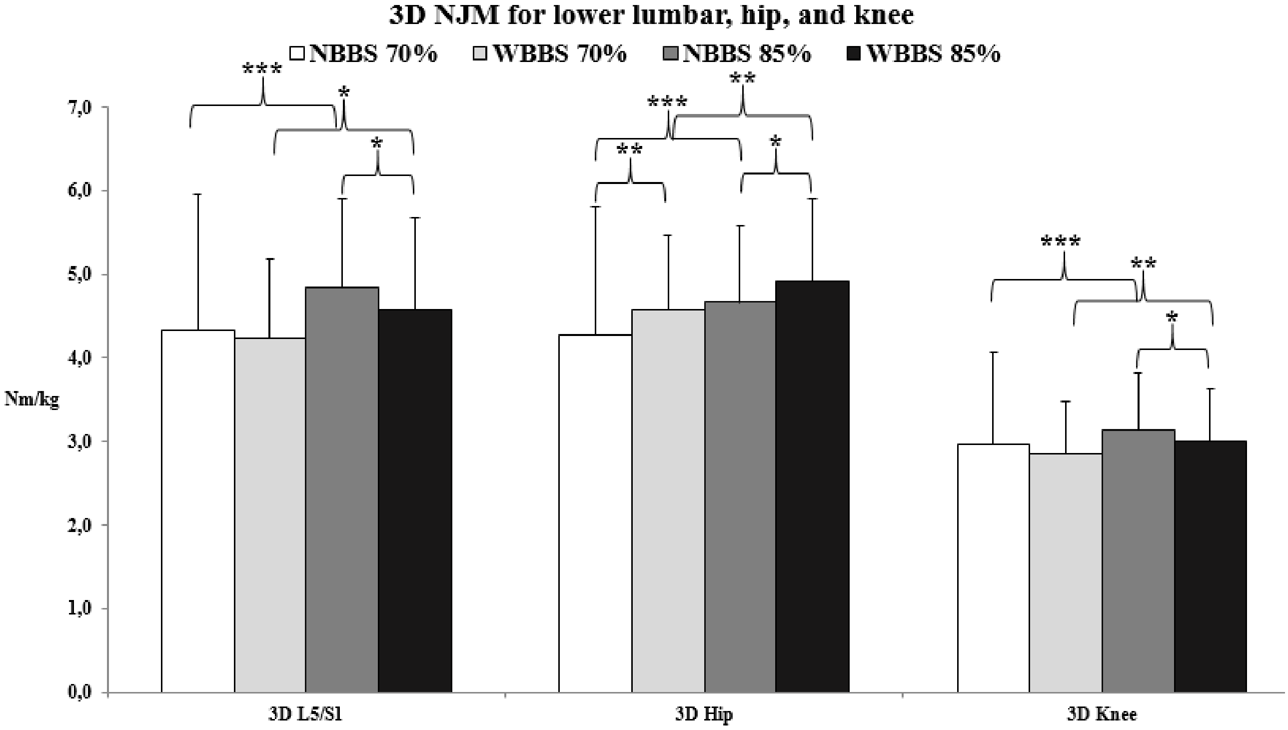


FIGURE 13. 3-D NJM for L5/SI, Hip, and knee. Both width and load interactions included. ***** = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001. A substantial amount of interactions can be found for both squat width and load. The 3-D hip NJM had the most interactions and was the only 3-D NJM that showed significance within widths at both 70 and 85% of 1-RM loads, in this case in favor for the WBBS. The effect of width on 3-D lower lumbar and knee NJM were less clear. Except for the lower lumbar, load interactions ranged from very- to highly significant between both the WBBS and NBBS loads.

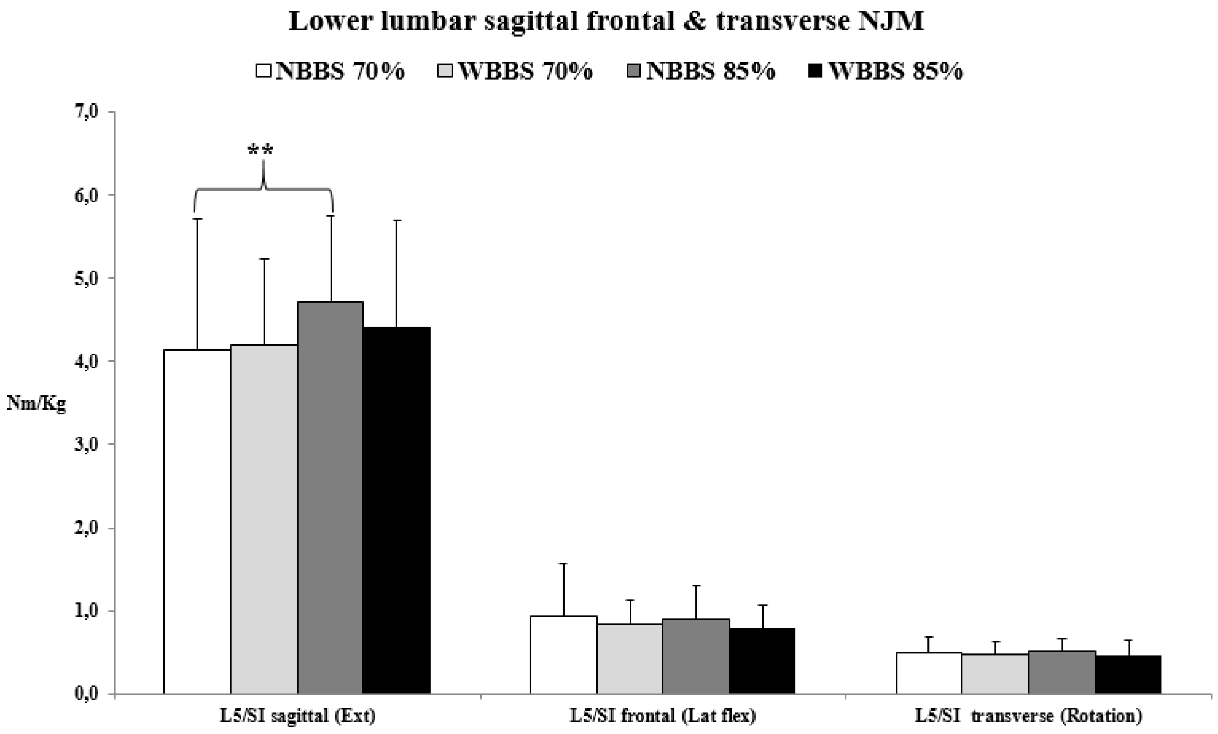


FIGURE 14. L5/SI NJM in all 3 planes. Both width and load interactions included. * = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001. Due to the spine moving mostly in the sagittal in the only interaction found was between loads, in this case the NBBS.

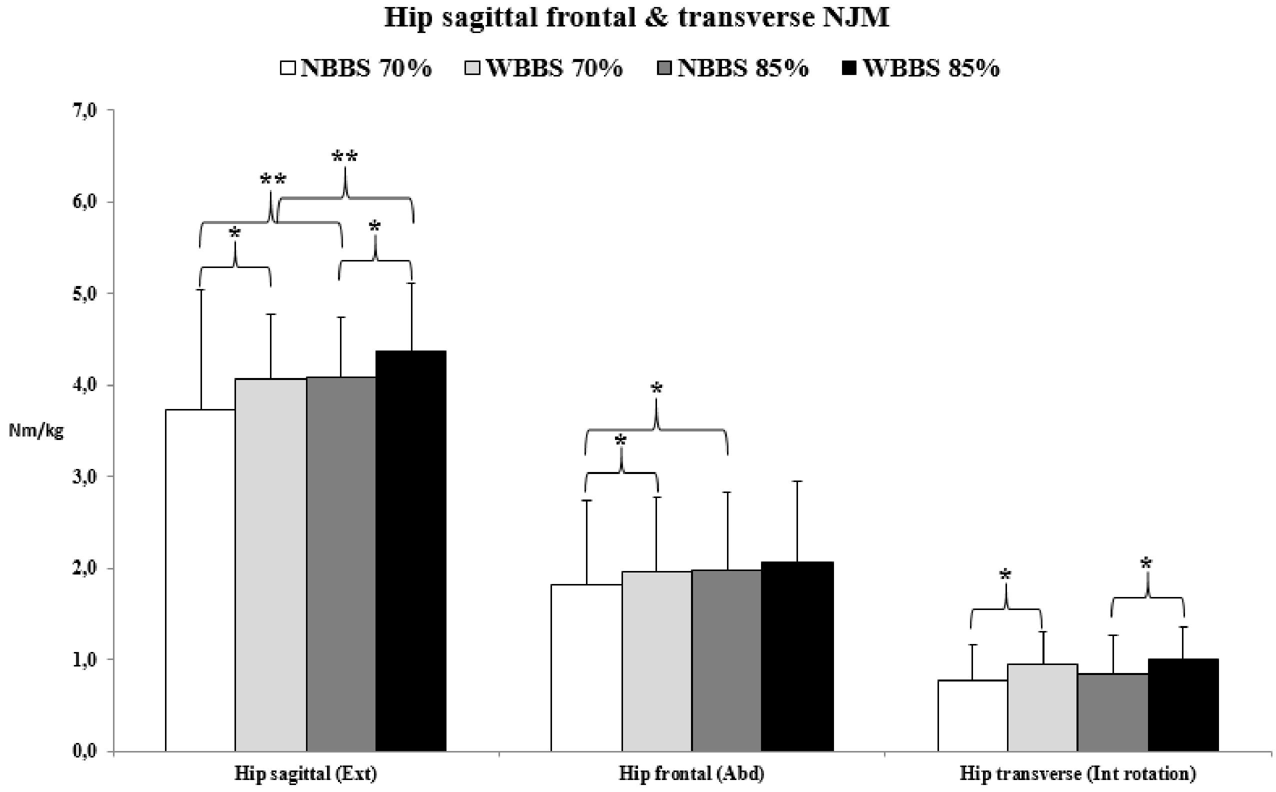


FIGURE 15. Hip NJM in all 3 planes. Both width and load interactions included. * = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates

highly statistically significant p<0.001. Most movement in the BBS is found in the sagittal plane, but significant contributions to total hip torque are also produced from the frontal plane and even to some extent from the transverse plane. These contributions become further evident in the WBBS.

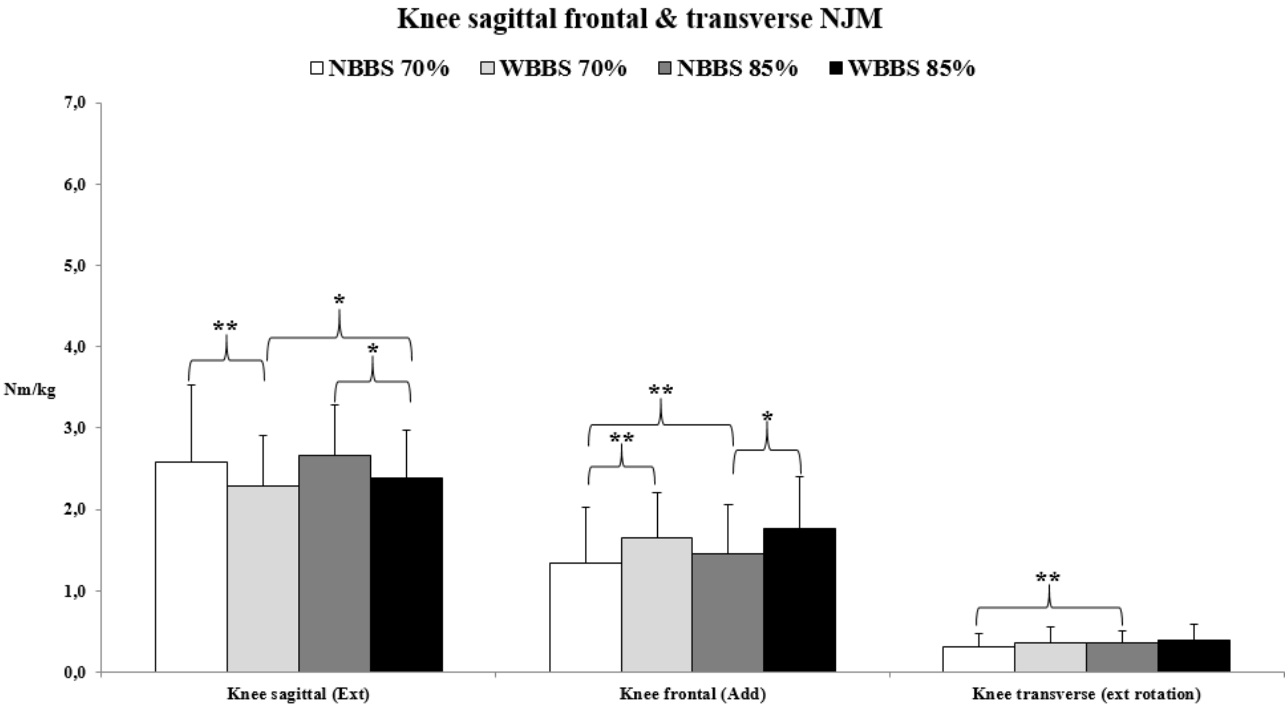


FIGURE 16. Knee NJM in all 3 planes. Both width and load interactions included. ***** = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001. At the knee, sagittal plane NJM are also logically the largest contributor, in this case in favor of the NBBS. Interestingly, frontal plane knee NJM were not far behind and were significantly higher in the WBBS. Load interactions were found in all planes, but more so in the NBBS.

Out of the 2 hip-to-knee NJM ratios measured both reached large ES for higher hip-to-knee ratios in WBBS (p<0.01). Large ES was present in both loading conditions, further strengthening the result. No statistical significance was found for the load condition (p>0.05).

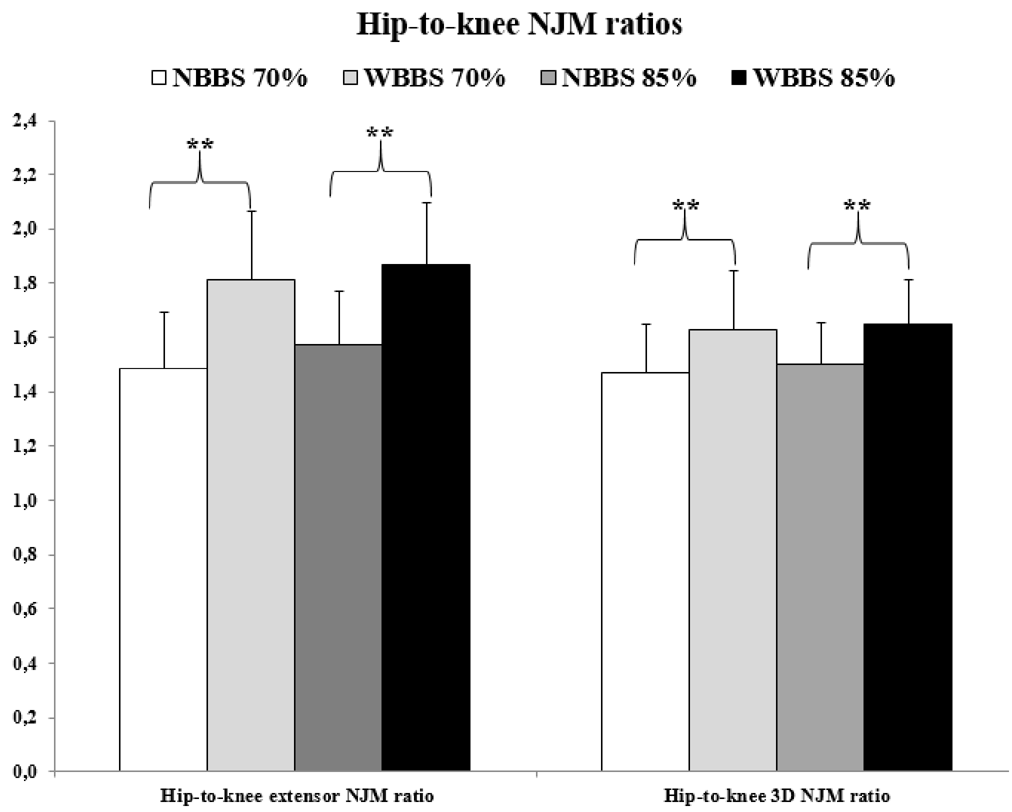


FIGURE 17. Hip-to-knee NJM ratios. Both width and load interactions included. * = indicates statistically significant $p<0.05$, ** = indicates very statistically significant $p<0.01$, *** = indicates highly statistically significant $p<0.001$. Although both extensor and 3-D NJM ratios reached very statistically significant levels in favor the WBBS, the ratio was clearly lowered in the 3-D divide, especially in the WBBS. This was mostly due to that frontal plane knee NJM were higher in the WBBS, leading to a reduction in hip dominance.

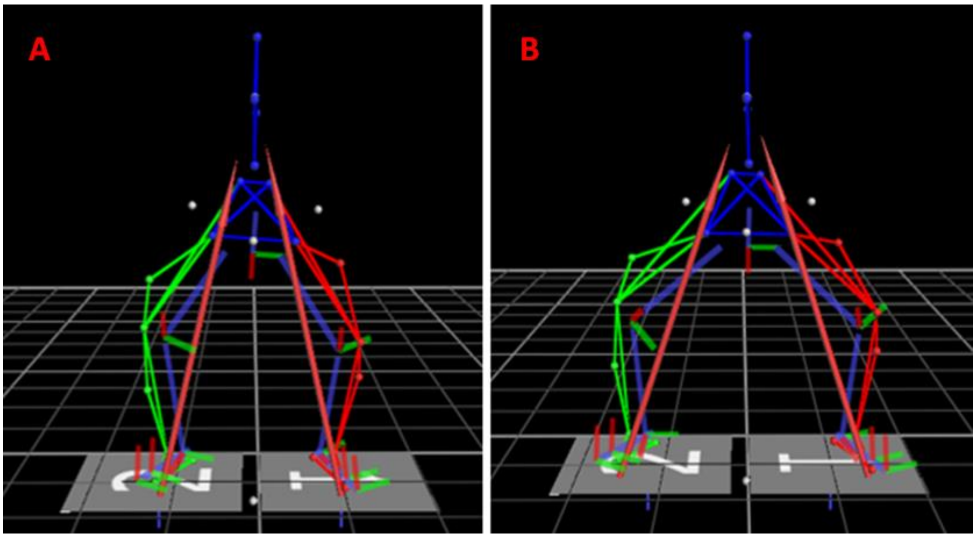


FIGURE 18: Resultant force vector behavior seen from the front in the NBBS (A) and WBBS (B) at 70% of 1 RM. Screenshot taken from the NEXUS program at approximately 70% of the ascent phase. The significant frontal plane knee NJM was found due that the resultant force vector leans medially creating a significant knee adduction demand, further increasing when width was increased.

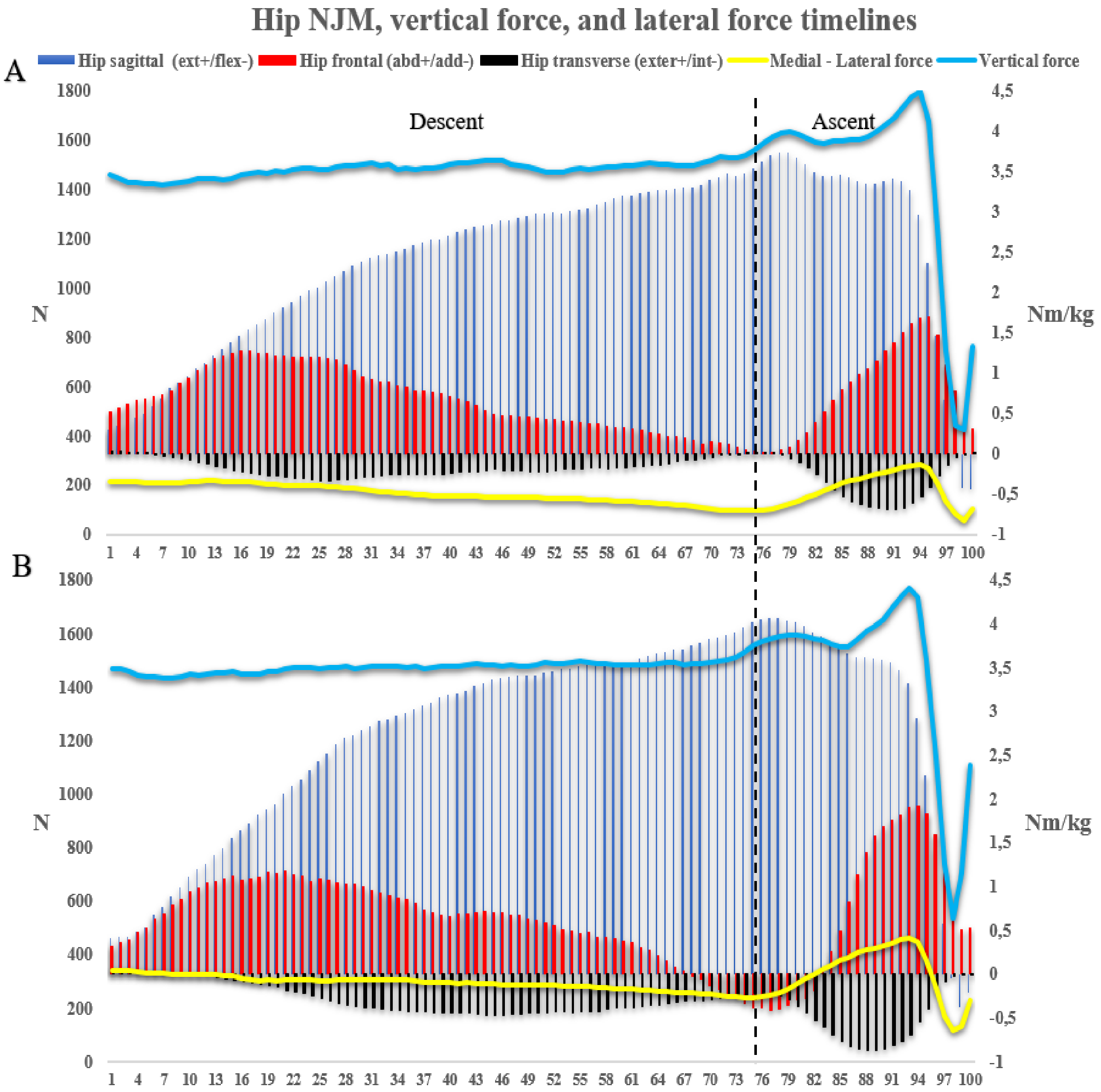


FIGURE 19. Group average of moment-time curves for the hip (sagittal, frontal, transverse) and force-time curves for vertical and lateral (medially directed) directions in the NBBS (A) and WBBS (B). Data taken as an average from the 70% and 85% loads from the first repetition. Here we can see how the external forces interact with the NJM. The medial-lateral force curve is quite clearly coordinated with the frontal and transverse plane NJM (hip abduction & internal rotation) but not the sagittal plane NJM, mostly evident in the ascent phase (around 75% of the total repetition time)

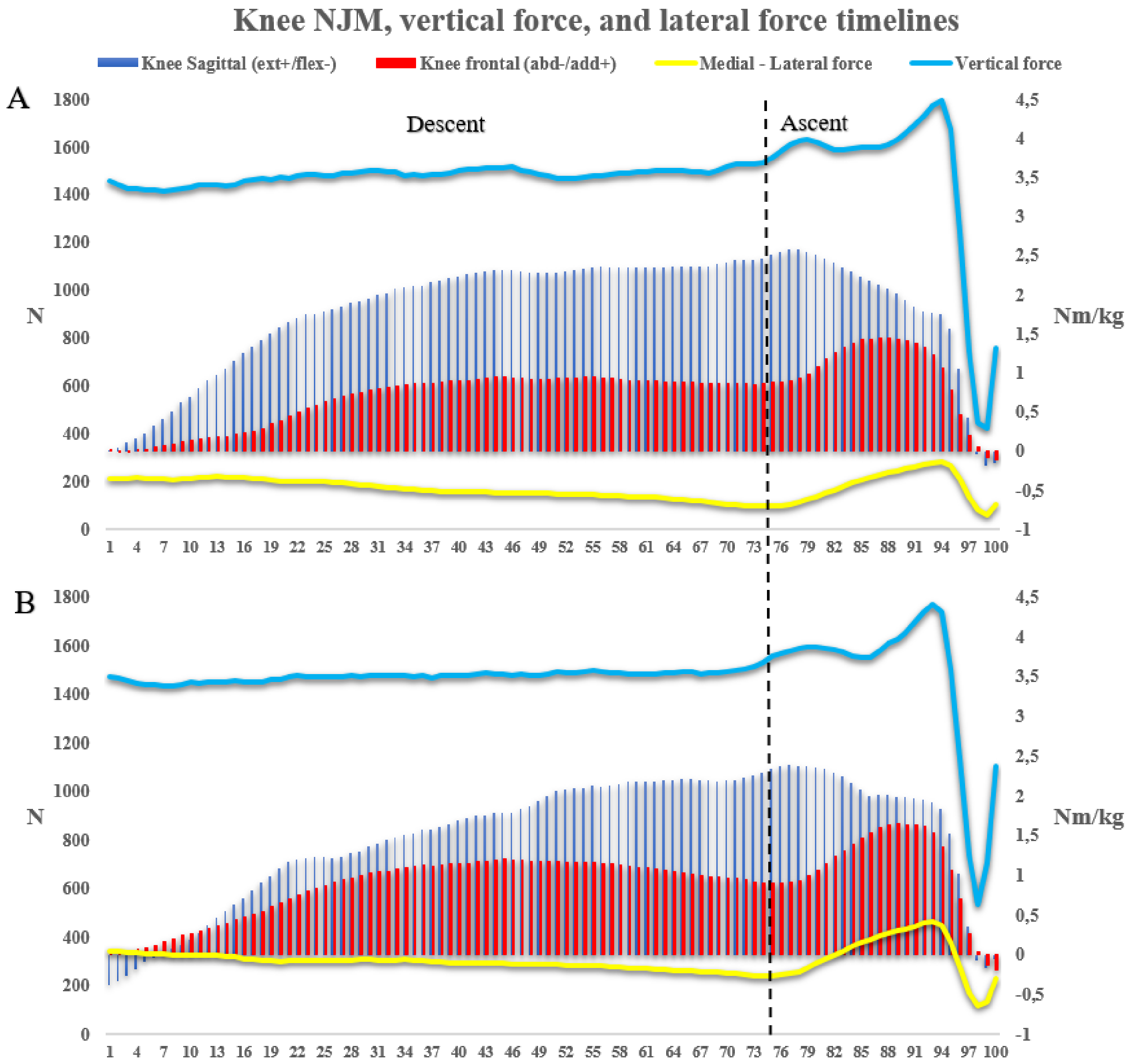


FIGURE 20. Group average of moment-time curves for the knee (sagittal, frontal) and force-time curves for vertical and lateral (medially directed) directions in the NBBS (A) and WBBS (B). Data taken as an average from the 70% and 85% loads from the first repetition. Transverse plane disclosed due to no significant interactions. Here we can also see how the external forces interact with the NJM. Again, a connection between the medial-lateral force and the frontal plane NJM can be observed mostly in the ascent phase, but not the vertical force.

6.3 Peak relative external forces

The peak relative external relative force results are presented APPENDIX D table 6. In both loads, there was a large ES for higher peak relative vertical force in NBBS (p<0.001). In both loads, there was a large ES for higher peak relative medial-lateral force in the WBBS (p<0.001). No statistical significance was found for higher loads (p>0.05).

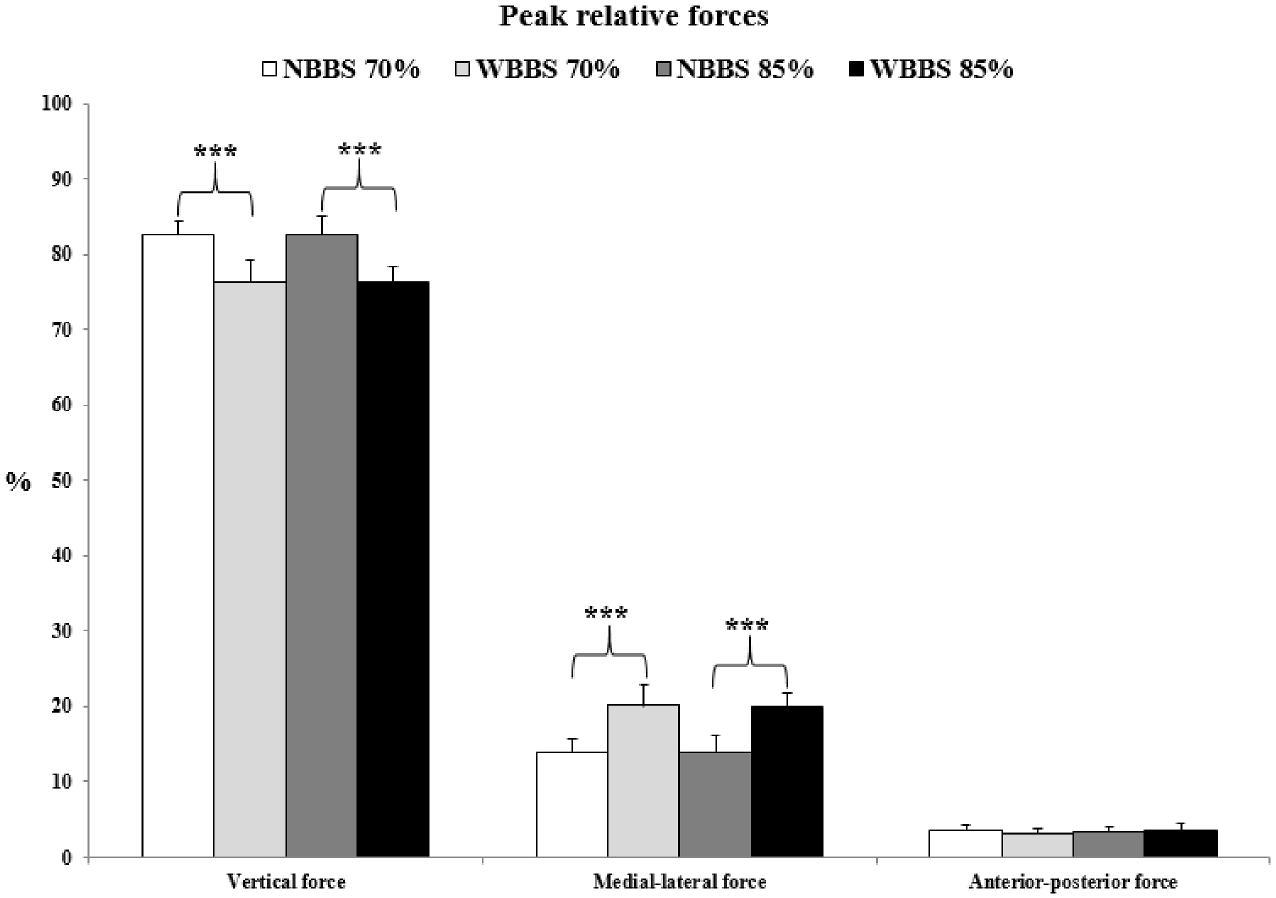


FIGURE 21. Peak relative external force in NBBS and WBBS. Both width and load interactions included. ***** = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001. The contribution of medial-lateral force was evident in both forms of squatting, but significantly more so in the WBBS.

6.4 HD-sEMG

All BFLH HD-sEMG results are presented in APPENDIX D table 4 and 5. For both the BFLH and ST, no statistical significance was found for the regional interactions (p>0.05). The WBBS 85% reached a large ES for higher activity in the BLFH overall measurement (p<0.01), medial region, and at both loads in the proximal region compared to the NBBS 85% (p<0.05). At the ST, WBBS 85% reached a large ES for higher activity at ST overall, medial, and proximal regions (p<0.05). Heavier loads reached a large ES for both NBBS and WBBS. NBBS 85% had higher activity in the proximal region compared to the NBBS 70% (p<0.05). WBBS 85% had higher activity at BFLH overall, distal, ST overall, medial (p<0.01), and proximal region (p<0.05).

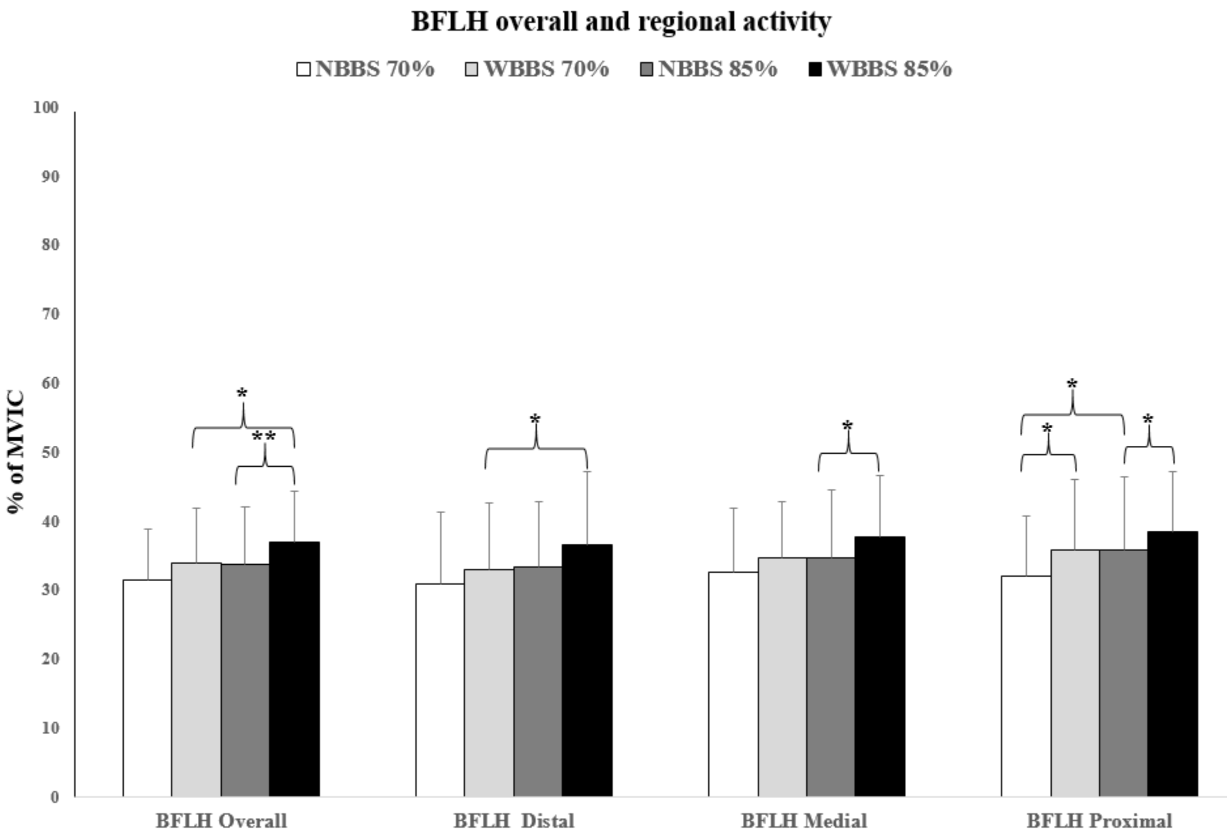


Figure 22. Normalized BFLH overall and regional activity for the ascent phase. Both width and load interactions included. ***** = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001. Mean activity values in both widths ranged at low levels when normalized to MVIC. Significant activity differences were found in favor of the WBBS in all regions except for the medial region. Although only reaching the first level of statistical significance, the strongest p-value was found for width interaction in the BFLH overall comparison.

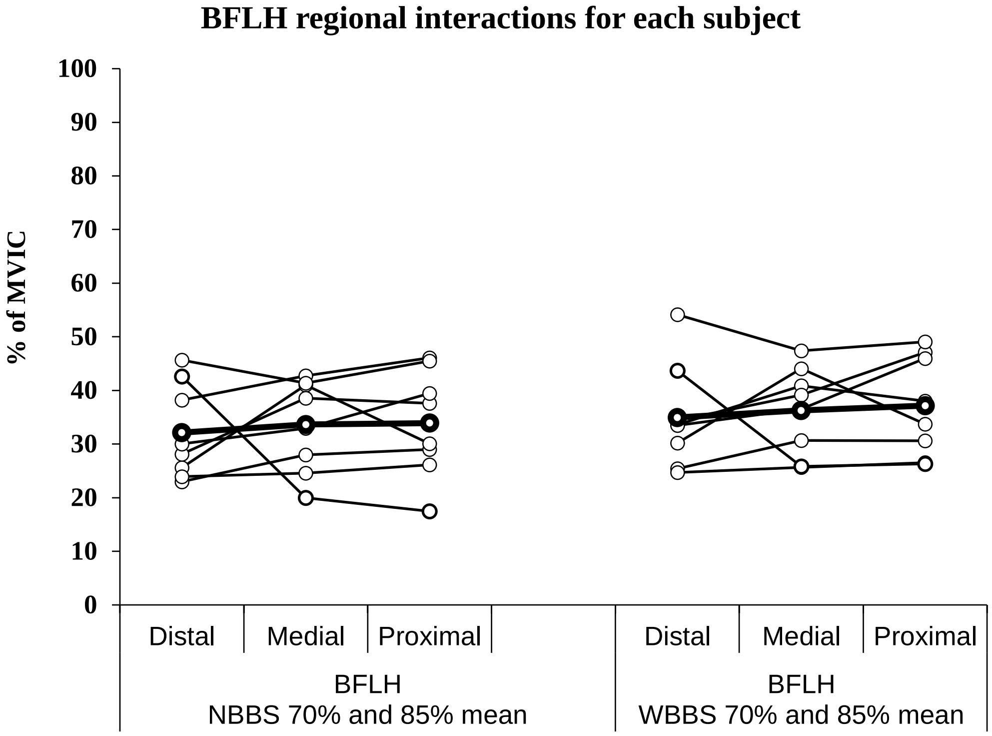


FIGURE 23. Pooled (70+85%) BFLH mean regional activity WBBS and NBBS for all subjects. Thicker black line represents group mean. Within subject activity patterns are visibly evident in both forms of BBS with a couple of outliers. No differences were found in regional interactions between widths.

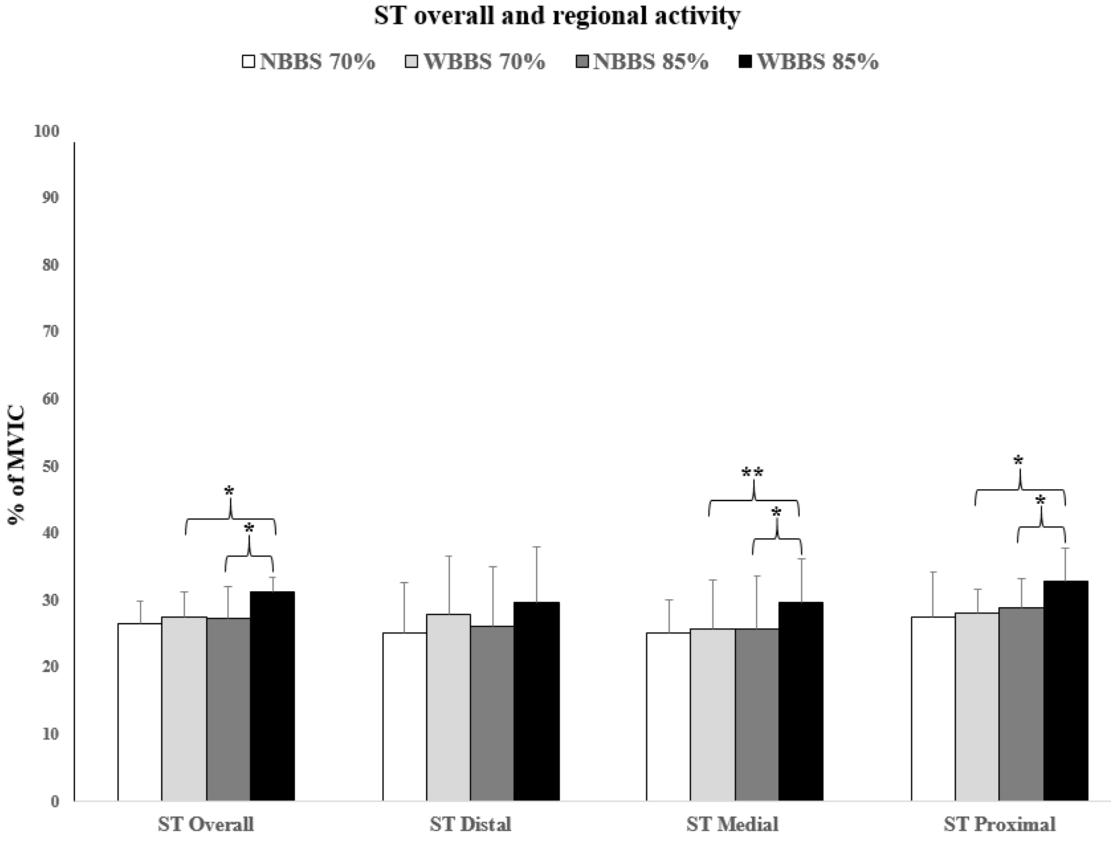


FIGURE 24. Normalized ST overall and regional activity for the ascent phase. Both width and load interactions included. ***** = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001. Levels of activity in the ST were even lower than BFLH. SD were in general low therefore significant activity differences were found in favor for the WBBS, although only reaching the first level of significance.

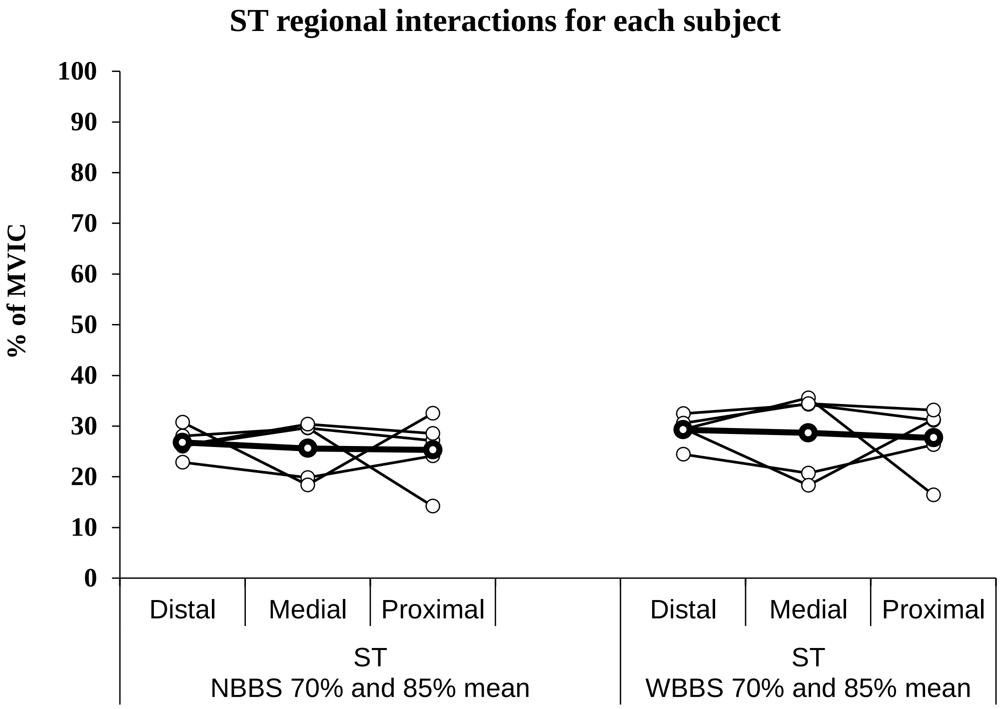


FIGURE 25. ST mean regional activity WBBS and NBBS (70% + 85%) for all subjects. Thicker black line represents group mean. Unfortunately, only 5 subject's data could be used for analysis. Similar to BFLH, differences in within subject regional activity patterns are present with a couple of outliers.

6.5 sEMG

GM and VL sEMG results are presented in APPENDIX D table 8. The WBBS 70% reached a large ES for GM higher activity compared to the NBBS 70% (p<0.05). The NBBS 85% reached a large ES for higher GM activity compared to NBBS 70%. No statistical significance was found for any VL interactions (p>0.05).

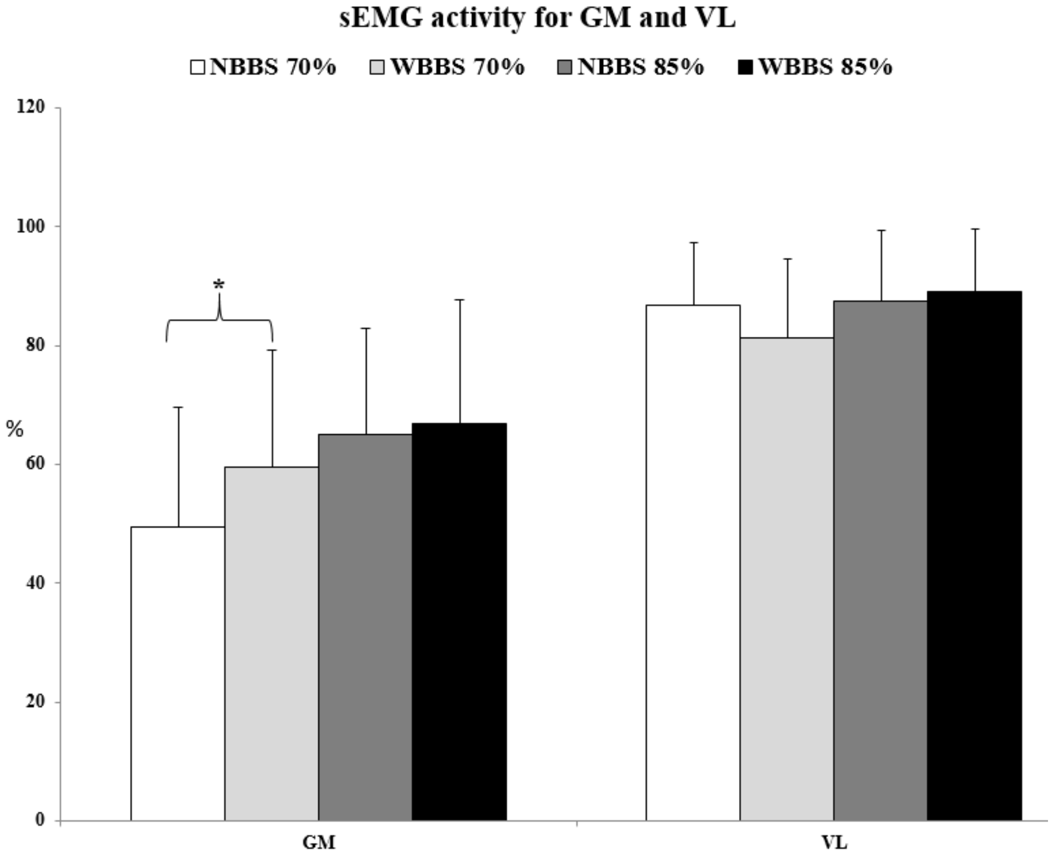


Figure 26. GM and VL activity (% of reference contraction) for the ascent phase. Both width and load interactions included. ***** = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001.

7 DISCUSSION

The main findings were that in the BBS, change in stance width combined with restriction of anterior knee movement had significant interaction effects on kinematics, kinetics, and muscle activity and therefore could lead to different long-term adaptations. Specifically, in athletic populations that utilize the BBS, the increased 3-D kinetic stimulus provided in the WBBS compared to the NBBS might be highly relevant observations for strength transfer. While the hip joint differences were most evident, the finding that the WBBS did not seem to decrease demands for the knee extensors, while not increasing the demands on the lower lumbar are also relevant. Interpretation of EMG results has more value when NJM are provided and vice versa, therefore in depth discussions can be provided with more integrity. Additionally, quantifying the direction of external force production (vertical-, medial-lateral-, anterior-posterior force) was slightly outside of our research questions, but helped us grasp the causality behind certain internal kinetic results.

7.1 Hamstrings HD-sEMG

Our primary goal in this study was to focus on the activity of the hamstrings. Previous research that has used a similar repeated measures study design has failed to show differences between the squat widths (McGaw & Melrose 1999, Escamilla et al. 2001b, Paoli et a. 2009). Although mean activity was in low levels (26-38% of MVIC), to the author's knowledge this is the first study that shows significant hamstring activity differences between the WBBS and NBBS. This was despite the fact that for the BFLH, 5 out 8 subjects had in average 6% higher absolute loads for the NBBS (1 subject had equal loads and 2 higher in the WBBS with an average of 4%). For the ST 3 out of 5 subjects had heavier absolute loads with an average of 4% (1 subject had equal loads and one had higher in the WBBS by 6%). VL activity did not change significantly in the WBBS compared to the NBBS even though the knee extensor NJM was significantly lower in the WBBS, and also despite that the knee flexion angle was significantly higher in the NBBS. Also, there was an increase demand on the hip extensor NJM in the WBBS at significantly higher hip flexion angles, therefore this data suggests that there is probably increased co-contraction at the knee in the WBBS. The positive and negative aspects of this phenomenon can only be discussed on a speculative level and it should be looked at from both an injury prevention and performance perspective. The WBBS increases tibiofemoral compressive forces and therefore seems to minimize tibiofemoral shear forces, thus having the potential to alleviate stress on the ligaments of the knee (Escamilla et al. 2001c). As mentioned before, the increased activity of the hamstrings should further provide a posteriorly directed pull of the tibia to help neutralize anterior shear forces at the knee (Escamilla et al. 2001a). This posterior pull of the hamstrings could even reduce mediolateral shear forces (Palmieri-Smith et al. 2009, Slater & Hart 2017). But in a WBBS there is significantly higher knee adduction demands (figure 19), therefore probably elevating stress differently on the LCL and MCL, which cannot be confirmed until more in depth internal joint force studies are completed. In terms of performance, increased co-contraction might decrease power production in the ascent phase by effecting the speed component, although this would probably lead to more impulse due to a longer TUT. Unfortunately, in this thesis project other external kinetic values than the peak relative external forces (vertical, medial-lateral, anterior-posterior) were not measured. In Swinton et al. (2012) study, although impulse was not measured, they found no significant differences between the NBBS and WBBS in all external kinetic variables (peak force, peak power, RFD), therefore it seems that the increased co-contraction in the WBBS is in such a small extent that it does not affect squatting performance from an external force-velocity-power (FVP) standpoint to a significant extent compared to the NBBS. If there would have been significant differences in performance there might have been differences in the average time of the ascent phase, which we did not find. Also, although using the knee extension exercise, Carolan & Caferalli (1992) demonstrated that in an 8-week strength training intervention first week neural adaptations in training included significant decreases in co-contraction at the knee, whereas there was no significant decrease in the following 7 weeks. Therefore, due to our 3-week familiarization period and 1 week of 1 RM testing, it is unlikely that co-contraction happened due to the movement being unfamiliar.

In terms of sEMG, comparisons to previous literature has multiple limitations. Most importantly, strong limitations are present when taking into consideration the accuracy of electrode placement and/or differences in electrode placement or the lack of electrodes at specific locations. To the authors knowledge, this is also the first squat study that has utilized 2-D ultrasonography to find the midline of the hamstring muscles, in this case the BFLH and ST. This is highly advantageous in terms of accuracy, specifically for taking every subjects anthropometry into consideration. Out of the 3 repeated measures studies that compared WBBS and NBBS, two reported electrode placements. In these studies, the electrodes were placed based on sEMG electrode placement guides approximately on the mid belly of the studied hamstrings (McGaw & Melrose 1999, Escamilla et al. 2001b). Escamilla et al. (2001b) preferably used the terms “lateral” and “medial” hamstrings, due to the cross-talk issues with isolation of specific hamstrings with the help of following generalized sEMG guides. As mentioned in the method section, our HD-sEMG electrode arrays were placed so that the medial region of the HD-sEMG array electrode was in contact with the middle of the muscle belly (Figure 9). This means that if any region can be compared to previous squat sEMG literature, it would be the medial region. This though is a questionable comparison due to this study having 5 electrode pairs per region and using mean activity of working channels, therefore making even the medial region hard to compare to previous literature with high accuracy. Even so, in this study the medial region of the BFLH and ST reached higher activity in favour of the WBBS condition leading to a contradictory result.

Also, although there has been different use of MVIC normalization protocols in similar studies, our results demonstrated mean activity levels in the hamstrings of 26-38 % of MVIC, which is in line with previous WBBS vs. NBBS studies (McGaw & Melrose 1999, Escamilla et al. 2001b). Therefore, it is unlikely that normalization methodology had a significant effect on the different results reported in literature. Therefore, it is highly likely that the combined effect of accuracy of electrode placement using 2-D ultrasonography and increased surface volume of activity readings via HD-sEMG technology were the main culprit to the contradicting results in this study compared to previous studies. Also, other methodological differences in this study compared to previous studies such as completing a 1 RM test for both widths, taking sEMG specifically from the dominant leg, a long familiarization protocol to standardize movement patterns with high accuracy, being strict with forward knee travel in the WBBS, same bar positioning, no biomechanical influence of footwear, and tempo control might have affected the cumulated result.

Our first hypothesis also included that there would be different regional interactions between the squatting widths. As mentioned before the idea of possible differences in regional activity between the squatting conditions was inspired by the idea that multiple studies have shown differences in regional hamstring activity between exercises (Mendiguchia et al. 2013, Schoenfeld et al, 2015, Mendez-Villanueva et al. 2016). Further, it seems that increasing activity in the proximal regions of the hamstrings is possible with hip emphasised exercises (Mendiguchia et al. 2013, Mendez-Villanueva et al. 2016), leading to the theory that higher hip NJM in the WBBS might lead to higher proximal activity. Our results did not confirm this theory. Observing the trends between individuals in figures 23 and 25 one can observe that outliers were present and in such a small group this will inevitably cause issues in data interpretation. The proximal region in the BFLH was the only region that reached statistical significance at both WBBS loads, but t-tests do have a higher risk of type 1 error, therefore the 2x3 anova was conducted to check regional interactions. Clearly, a new study with larger N is needed to confirm or reject this theory. It would be interesting if an fMRI study would be done, where both the NBBS and WBBS (with clear knee movement restriction) would be done to technical failure with specific loads and then tested for regional activity differences.

In terms of the joint kinematics one reason for the higher activity in the WBBS condition can quite possibly be the higher movement from the hip joint relative to the knee joint, therefore not shortening or reducing tension in the hamstrings as much as the higher knee flexion angle would in the NBBS. It is more appropriate to not confirm or deny elongation when it comes to the bi-articular hamstrings, because we do not actually know without appropriate imaging the true nature of the musculotendon unit's contraction. Nonetheless, tension differences can be assumed but to what extent would be highly speculative. As such, it is worth noting at least to some degree the role of the hamstrings in rotation. In addition to supporting hip extension and knee flexion, BFLH can assist in external rotation of the femur and the ST internal rotation (Biel. 2010). The WBBS reached significance for higher internal rotation angles. This means in theory that the increased rotation combined with increased hip flexion compared to knee flexion WBBS could have further affected the length – force relationships in the hamstrings leading to changes in EMG (Mohammend et al. 2003, Jónasson et al. 2015). But the influence of dynamic internal and external rotation is probably small due to the small moment arms the hamstrings have for rotation (Dostal et al. 1986) and their role have been more proposed to be more stabilizing in nature (Hooren & Bosch 2016). Unfortunately, due to the multivariate nature in kinematic differences between the WBBS and NBBS, causality cannot be pinpointed concerning the higher activity with high degrees of accuracy. Specifically, it is not known to what extent higher hip flexion combined with lower knee flexion angles affected the activity demands. It would have been interesting in the NBBS to do a version where the knees are significantly more restricted, similar to Chiu et al. (2016) study set up where a board was used to avoid knees passing midfoot. If in this condition hamstring activity would have increased according to the HD-sEMG electrodes, it would have confirmed that increased hip flexion relative to knee flexion is an influential factor and a comparison could have been made to the influence of internal/external rotation with some accuracy.

7.2 Moments and forces

Our second hypothesis was that the WBBS will have higher 3-D hip-to-knee NJM ratios due to higher 3-D hip NJM and lower 3-D knee NJM was confirmed and with some interesting revelations. Previous literature presenting hip and knee NJM in different forms of BBS have only presented sagittal plane knee NJM (Wretenberg et al. 1996, Escamilla et al. 2001a, Swinton et al. 2012). Also, as mentioned previously, when hip-to-knee NJM ratios have been presented they have been a ratio between sagittal plane NJM, in other words; a divide between peak extensor NJM of the hip and knee (Beardsley & Contreras 2014). Beardsley & Contreras (2014) literature review of different compound strength training exercises hip-to-knee extensor NJM ratios was intriguing and to the authors knowledge is the first that has done so. Because comparing NJM accurately between studies can be questionable (Robertson et al. 2014. p. 121), ratios might increase the reliability. One of the thesis goals was to continue this innovative ratio by observing if similar results could be attained while presenting more data on the topic. There was also a goal to add further credibility to the approach by comparing both hip-to-knee extensor NJM ratios and all three planes in form of a resultant 3-D hip-to-knee NJM ratio at different loads. This way, the ratio takes into consideration the entire net moment requirement of a specific joint in a specific movement pattern. Beardsley & Contreras (2014) calculated that the NBBS had a hip-to-knee extensor NJM ratio of 1.32:1 at 70% of 1 RM and 1.49:1 at a 90% of 1 RM. Our study showed a similar relationship, with a NBBS showing a ratio of 1.48 (± 0.22) at the 70% of 1 RM and 1.57:1 (± 0.20) at 85% of 1 RM. In the WBBS there was approximately a 22% increase, with a hip-to-knee extensor NJM ratio of 1.82:1 (± 0.26) at 70% of 1 RM and 1.87:1 (± 0.25) 85% of 1 RM. Interestingly although still reaching a large ES ($p<0.05$), the hip-to-knee NJM ratio differences between the NBBS and WBBS reduced substantially when it was calculated in a 3-D NJM format at all loads (Figure 18). Although hip-to-knee NJM ratios decreased both in the WBBS and NBBS when transferred from extensor NJM to 3-D NJM, this happened to a larger extent in the WBBS. Specifically, although the knee NJM were lower from a 3-D perspective in the WBBS, the knee frontal plane NJM, or in other words knee adduction NJM, was significantly higher in WBBS at both loads (Figure 17), peaking in the concentric phase (figure 20), leading to the largest contribution in the decrease of the ratio. As one can see from figure 17, the knee extensor NJM still stayed as the dominant contributor to 3-D knee NJM, therefore still leading to the 3-D knee NJM being significantly higher at the knee in the NBBS. The significantly higher knee adduction NJM demands in the WBBS were an interesting unaccepted finding. This made sense directly when observing how the resultant GRF vector behaves both in the NBBS and WBBS (Figure 18).

In figure 18 it is evident that there is a significant medial deviation of the resultant GRF vector in both squat widths, with the deviation further increasing in the WBBS. This is logical, due to as mentioned before, the resultant GRF aims towards COM. As presented in figure 21, the contribution of peak relative external forces in the NBBS and WBBS are significantly different. Although the vertical GRF is by far the dominant force (~76-82%), medial-lateral force contributed an average of 13.8% in the NBBS and 20% in the WBBS for both loads, leading to large ES between both widths ($p<0.001$).

From figures 19 and 20 one can see a timeline from 0-100% how the 3-D NJM at the hip and knee (not including the knee transverse plane due to low values) behave in the ascent and descent phase of the NBBS and WBBS. From the figures 19 and 20 one can see that the medial-lateral force demands (medially directed) are coordinated with not only knee adduction NJM demands, but also hip abduction and internal rotational NJM demands in both the NBBS and WBBS. As one can also see, these NJM demands further increase with width while staying coordinated with the medial-lateral force curve. The hip internal rotation NJM peaking instead of external rotation NJM was slightly surprising at first sight, especially in the ascent phase when the femur is actively moving from internal rotation to external rotation (figure 12). This though is logical once considering the effect of the medially directed resultant GRF vector, which would start placing external rotation demands on the femur, therefore internal rotation torque is needed to counter this. This also applies for the increased hip abduction NJM demands in the ascent phase. The hip is moving actively from hip abduction to adduction when rising, yet abduction demands increase. This again is logical when taking into consideration how the medially directed resultant GRF vector “pulls in” the femur. This means that there is probably quite a bit of eccentric contractions and therefore co-contraction going on around at both the hip and knee in both BBS conditions. At the hip joint, the hip abductors such as the gluteus medius will probably work as its own antagonist, due to the anterior fibres are producing internal rotation torque in form of an eccentric contraction (because the hip is moving into external rotation) and the posterior fibres external rotation torque. Muscles such as adductor magnus are aiding in hip extension (Vigotsky & Bryanton 2016), and therefore also creating some adduction torque, which might cause even higher demands for the hip abductors. Therefore, it would have been very interesting to measure sEMG from many other muscles, including adductor magnus, and sEMG from both the anterior and posterior fibers of the gluteus medius and see how the activity shifts between widths. This leads to a conclusion that the medially directed lateral force demands in the BBS that are further increased in the WBBS (13% vs 20% contribution in the NBBS and WBBS, respectively) should be large enough to be considered to have practical significance. Practitioners might connect this to be a beneficial stimulus for increasing performance in sports with a combination of high lateral and horizontal force demands such as sports involving change of direction (Dos`Santos et al. 2016). In this case, it is important to add that when explosive lateral movement demands are high, such as side stepping, hip extensor NJM seems to increase with intensity while hip abduction NJM demands stay stable across all intensities (Inaba et al. 2013). This might mean that although hip abduction strength is important and should be trained, it seems that it might be more in a role of stabilizing the hip to support the larger role of the hip extensors in increasing performance demands. Therefore, any movement that supports this relationship might have functional benefits. Also from a sports injury prevention perspective, knee abduction coupled with hip adduction and hip internal rotation (knee valgus) in landing mechanics (bilateral movement patterns similar to squatting) is associated with a smaller knee adduction/varus NJM (Kernozek et al. 2005). Therefore, strength training under conditions with higher knee adduction NJM demands combined with increased 3-D hip NJM demands such as in the WBBS might be beneficial from a performance and injury prevention standpoint.

At least two theories might explain why the hip-to-knee NJM ratios have the potential to change with increasing load. These include COM behaviour when the load increases and shifts in movement pattern during the heavier loads that would inevitably lead to changes in NJM behaviour. In terms of how a change in COM could affect hip-to-knee NJM ratios is based on the idea that a heavier load on the back should lead to the COM travelling closer to the bar the higher the load is. This means that if the resultant GRF vector directs itself towards COM when pushing the weight, the elevation of COM should affect the direction of the resultant force vector and therefore should influence NJM at the knee and hip. This movement of COM is highly dependent on the relative strength of the athlete, because COM will probably not move significantly if the athletes’ relative loads are significantly under 1.5 x body mass (subject’s relative strength levels in this study: NBBS: 1.33 ± 0,19, WBBS: 1.33 ± 0,18). Also, the effect of COM’s movement on NJM is probably not sensitive enough to be significantly visible with 15% load increments in the population this study used. The data that Beardsley & Contreras (2014) used to calculate hip-to-knee extensor NJM ratios for the NBBS was taken from Bryanton (2011) master’s thesis. Bryanton (2011) also provided knee and hip sagittal moment data from different loads, which interestingly showed that knee-to-hip extensor NJM ratios increased with load, but only due to the hip NJM increasing while the knee NJM stayed stable. This results from this study partly contradict this by showing increased NJM with load at both the hip and knee. Other squat studies also demonstrate increasing knee extensor NJM with load (Swinton et al. 2012, Cotter et al. 2013). Also, and possibly more importantly, strict technical 1RM testing was used because the goal was to avoid significant movement pattern change between loading conditions. It is possible that Bryanton et al. (2011) study design did not emphasise this criterion as much. If this is the case then shifts in movement patterns at higher loads such as at 90% of 1 RM are quite normal based on anecdotal evidence, which is usually visible with the lifter shifting the hip more posteriorly to gain more hip torque and reduce the demands on the knee extensors. This would quite possibly significantly change the ratio, but to the authors knowledge no squat study exists that has observed the NJM in this fashion. Therefore, it would have been interesting to do a heavier set to technical failure and compare NJM behaviour, even between legs.

In terms of the L5/SI NJM demands, in other words the lower lumbar, the hypothesis was that the demands would be similar between NBBS and WBBS. Previous literature has reported increased lower lumbar extensor NJM in the NBBS compared to the WBBS (Swinton et al. 2012). Because this study used slightly shallower depths (femur parallel vs. thigh parallel) than Swinton et al. (2012), it was predicted that there would have more control over lumbo-pelvic area, which would reduce forces on the lower lumbar. Lumbo-pelvic control was not quantified in anyway, but the results only partly confirmed our hypothesis, with no differences between the NBBS and WBBS lower lumbar NJM at 70% of 1 RM, but significance found in the 85% of 1 RM with higher lower lumbar 3-D NJM in the NBBS 85% compared to WBBS 85%. Also, the NBBS 85% had significantly higher lower lumbar extensor NJM than the NBBS 70%, but the same phenomenon was not found in the WBBS between loads. The load hypothesis was confirmed from the 3D NJM perspective, where higher loads had significantly higher lower lumbar 3-D NJM. Although large ES were found, it is hard to say if there is practical significance in the higher lower lumbar NJM between widths found in the 85 % loads due to the 70% loads showed no clear trend for such results. In fact, although not reaching significance, lower lumbar extensor NJM had a higher mean value in the WBBS 70% compared to NBBS 70%. Also, the load interaction between NBBS 70% and 85% reached highly significant levels ($p<0.001$) while the load comparison between WBBS 70% and 85% only reached the first level of significance ($p<0.05$). Therefore, for now it seems more appropriate to state that lower lumbar loads are similar between widths and they increase with load.

From a 3-D perspective, the NJM at the lower lumbar were similar to the 3-D hip NJM demands. But when observing the extensor NJM demands at the lower lumbar and the hip, the lower lumbar extensor NJM are higher. This is also in line with Swinton et al. (2012) results. Therefore, in terms of relative NJM extensor demands between the lower lumbar, hip and knee, the lower lumbar seems to have the highest demands, which also means that bilateral back squat performance regardless of width is highly dependent by lower lumbar strength. To further add quality to the interpretation of lower lumbar use, sEMG would have been beneficial. But based on previous literature we know that the lower erectors are highly activated in the squat (Clark et al. 2012).

7.3 sEMG

Our third and last hypothesis was that GM activity will be higher in the ascent phase of the WBBS, with VL activity only changing with loading condition. The GM hypothesis was partly confirmed, with a higher GM activity observed in the ascent phase in favour for the WBBS, but only in the 70% condition. The VL activity hypothesis was also only partly confirmed, with no significant activity differences found between the WBBS and NBBS, but our load hypothesis was not confirmed with the VL activity staying stable across loads.

To the authors knowledge, two studies have been published that specifically compare the biomechanics of NBBS and WBBS by measuring among other muscles both GM and VL activity (McGaw & Melrose 1999, Paoli et al. 2009). Both McGaw & Melrose (1999) and Paoli et al. (2009) showed significantly higher GM activity for wider positions while the VL activity did not change. This was despite the fact the studies did not use relative loading for each width, which our study did. Therefore, using relative or absolute loading does not seem to have significant effect on the activity, at least in this population of athletes.

The GM has shown significantly more sEMG activity in a hip extension exercise when the leg is put into abduction and external rotation (Suehiro et al. 2014) and in very low hip flexion angles where its moment arm is the highest (Worrell et al. 2001). Therefore, it is likely that the increased external rotation and abduction in the WBBS are the main culprits for the increased GM activity and not the increased hip flexion (Worrell et al. 2001, Suehiro et al. 2014, Vigotsky & Bryanton 2016). Inconsistency has been shown for GM across loads in other studies, where Paoli et al. (2009) found higher GM activity in the WBBS at 0% and 70% loads but not 30% loads. McGaw & Melrose (1999) found higher activity in the 75% condition but not the 60% condition. For now, it seems that GM shows significance the most consistently around heavier loads, therefore our low N could have contributed to the non-significant differences found at 85% of 1 RM.

The VL muscle was chosen to represent the quadriceps group because previous squat studies have shown that there is no evidence that one can significantly isolate one quadricep muscle more than the other regardless of width or depth (Wilk et al. 1996, McGaw & Melrose 1999, Escamilla et al. 2001c, Paoli et al. 2009). The results of unaffected VL activity between widths is in line with previous research (McGaw & Melrose 1999, Paoli et al. 2009). The squat movement pattern is a good example of when NJM values do not paint a realistic picture of the torque demands of specific muscle groups. As mentioned earlier, the results show that in the WBBS the VL increases in activity to match the activity of NBBS although knee extensor NJM are lower, most likely due to the hamstrings contracting harder when hip extensor demands increase (Figure 5) and also highly likely due to reciprocal inhibition. Also, this study was unable to show a load interaction for the VL that was a part of our third hypothesis. This hypothesis was based this on previous observations in literature where quadriceps activity has been reported to increase with higher loads (Li et al. 2013, Aspe et al. 2014, Gomes et al. 2015). Li et al. (2013), Aspe et al. (2014), and Gomes et al. (2015) had a 30% jump between loads minimum and maximum loads, whereas our study only had 15%, which might have been not sensitive enough to pick up differences. This sensitivity issue is in line with Bryanton et al. (2012) results, where quadriceps utilization was found to be more sensitive to the degree of depth than load.

Movement artefact and fascicle movement is an issue with sEMG, which possibly desensitizes the results and contributes slightly to the cumulative amplitude, especially in dynamic movement such as squatting. This being said, the BBS to parallel depth does not highly stretch the musculature and is quite tamed in limb velocity compared to measuring sEMG from more explosive movements such as jumps and sprints, therefore the effect of movement artefact is possibly reduced.

7.4 Kinematics

None of our hypotheses concerned the kinematics directly, but they were still essential to present when dynamic kinetic comparisons are made. The main objective with measuring the kinematics was to keep specific angles and movement speeds as similar as possible to avoid high variability in the NJM and sEMG. Tempo practice was used in the familiarization because it was also important to have similar time under tension (TUT) between relative loads when comparing kinetic differences. This was achieved, with no significance found between squatting conditions for the descent and ascent phase. Because the subjects were asked to move through the ascent phase as fast as possible while holding form, it was logical that a heavier load would have a longer ascent phase. This was the case, with a large ES found for the higher load in the ascent phase ($p<0.05$). Lastly, it was important to have no load effects for any of the measured joint angles, due to our aim was that the 1 RM was based on a strict technical 1 RM. This study succeeded in this goal by showing no load interactions ($p>0.05$). Previous literature has shown differences in joint kinematics when increasing load (Andrews et al. 1983), but it is highly likely that they did not standardize the 1 RM testing to a specific movement pattern.

Also, as a “kinematic bonus”, the COP was added in the posterior-anterior direction (figure 12) to help show that there is a more posterior shift on the foot in the WBBS compared to the NBBS. To the authors knowledge this has not been previously provided in any squat study. Statistics were not completed for the COP, so caution is advised in interpreting the results.

7.5 limitations

This thesis consists of methodological limitations, which should be considered when interpreting the results. The following limitations are the more substantial ones, but many more minor limitations exist.

Our 1 RM testing approach was also not without its limitations. For example, A subjects 1 RM is something that can fluctuate significantly on a daily basis (Jovanovic & Flanagan 2014). A common method is to have a 5-7-day break between tests and measurements so that the subject can recover adequately, which was a method this study used. Some methods have been proposed to measure daily readiness of an athlete, such as Velocity-Based-Training (VBT) (Jovanovic & Flanagan 2014). This method states that if a person’s mean velocity is known at 1 RM for a specific movement pattern, one can quite accurately predict daily readiness via measuring the mean velocity a couple of warm up sets and creating an extrapolated line between the weight and the speed. To the authors knowledge, this has yet to be used in acute strength training study to test daily 1 RM, but it would be interesting to use to increase the validity of the loads used.

There was different issues with our sEMG, HD-sEMG and NJM calculation systems. Specifically, some HD-sEMG electrodes had a lot of noise issues and led us to not trust the results in many subjects. Some electrodes were acceptable but still sensitive and led us to deleting channels across tasks. If deletion was required on the individual bases, being biased was avoided by deleting the same channel across all tasks, even though it seemed to work in specific cases. Specifically, for ST only 5 subject’s data were considered reliable. Due to human error, there was calibration issues with the AMTI force plates and Vicon global coordinate system and therefore interpretation of kinetic and kinematic data was not reliable for 4 subjects. Also, our bipolar sEMG system had to be changed in the middle of the study due to unknown technical reasons. Theses technical issues presented themselves by inconsistent spikes in activity and high sensitivity to movement artefact. Therefore, the signals recorded from the previous system before braking were not trusted, therefore it was decided to only publish data on the replaced system (7 subjects). Concerning the muscles measured, the primary objective in this study was the hamstrings leading us to use two MVIC tasks to increase the quality of the interpretation of the results. Nonetheless, MVIC methods does not show us if we truly reached the maximal activation potential/neural drive of the muscle (Halaki & Ginn 2012, Earp et al. 2013). Electric stimulation of the peripheral nerve to get an M-mave can be used for normalization, but they are not as practical for all muscle groups in the body, quite time consuming, and for some quite uncomfortable. Our lab has not been able to find an appropriate stimulus area to get a reliable M-wave from the hamstrings.

For normalizing the VL, the squatting task itself was chosen as a reference contraction. Using specific tasks as a reference contraction has also been accepted as an acceptable tool for normalization (Halaki & Ginn 2012) and has been used in previous squat studies (Wretenberg et al 1996, Aspe et al. 2014). A dynamometer would have been more reliable and could have been combined with m-wave methods, but was avoided due to time constraints and because it was not a priority in this study.

There are multiple assumptions made when calculating NJM via inverse dynamics, some that are more far reaching than others, such as segments are rigid, segment boundaries and mass distribution are artificially defined, joints are rotationally frictionless, there is no co-contraction of agonist and antagonist muscles etc. (Hatze. 2012). All this combined with calibration and filtering differences in controlled conditions makes it more difficult to transfer results into the real world. Therefore, it is essential to produce not only acute studies but intervention studies to add validity to the results. Musculoskeletal modelling would help to increase the validity of both NJM and sEMG measurements, but modelling was out of this thesis’s scope.

Other limitations include but are not limited to; having only 2 reps to analyse for each condition, assuming that the legs are being utilized exactly the same, depth was controlled by the practitioner’s oral feedback instead of using a depth marker, using both male and female athletes (although no studies show this is an issue, especially to parallel depth), and not comparing the kinetics and kinematics between the squat and any sport specific movements (sprint, jump, tackle etc.).

7.6 Strengths

This study included multiple strengths in terms of being able to quantify kinetic and kinematic data of a commonly used compound strength training exercise, the BBS. The BBS and its different variations have been vastly studied, but many of these studies have not had the opportunity to research both sEMG and NJM. To the authors knowledge, this is the first repeated measures BBS study that has quantified both sEMG and 3-D NJM. Further, this is the first study that has used HD-sEMG technology to quantify hamstring muscle activity in the BBS, which should add further value to the interpretation of biomechanical differences between the squat types. Although being a multi-joint free weight exercise, the aim was to control the lifting techniques as much as possible. This would help us avoid high variation in such a small population. This meant the subjects had to be proficient in the quantified techniques, which led us to have a proper familiarization protocol. There are BBS studies that have used 1 week for familiarization, but not further than this. Based on anecdotal evidence, even though an athlete might state that they are proficient at a specific version of the BBS, there can still be visibly high variety present within the specific version between populations, even if the population has similar anthropometry. These variations are not necessarily wrong or right, but possibly better stated less optimal or more optimal for a specific purpose. Therefore, we were strict with controlling anterior knee movement, trunk lean, depth, alignment, bar position, lumbo-pelvic control, shoulder and arm positioning, femur rotation, bracing, gaze, tempo, cueing, and even shoe wear. If this study inspires long-term studies that compare the WBBS to NBBS, one can hope the details in this study will help standardize the movement patterns.

7.7 Conclusion

The primary goal within this study was to build on previous research and add more biomechanical detail in interpreting the similarities and differences between standardized versions of the WBBS and NBBS in athletic populations. Specifically, the primary focus was hip and knee joint, with the lower lumbar as secondary interest. The primary findings were that the hamstrings activity was found to be higher in the ascent phase of the WBBS when utilizing HD-sEMG technology. Also, although the hip-to-knee extensor NJM ratio is higher in the WBBS, measuring the ratio in 3-D led to different observations. The secondary findings were that lower lumbar demands were slightly different between the widths while increasing with load and that GM activity increased in the WBBS condition while VL activity stayed the same between conditions. Therefore, this study succeeded in adding further interesting biomechanical detail in comparisons between the WBBS and NBBS, with reporting hamstrings overall- and regional activity via HD-sEMG technology, reporting resultant 3-D NJM for L5/SI, hip and knee, reporting knee NJM in all three planes and comparing hip-to-knee NJM ratios from both an extensor and 3-D perspective. Also, the interesting observations concerning the 3-D hip-to-knee NJM ratio led us to report additional kinetic data, specifically the peak relative contributions of vertical, medial-lateral and anterior-posterior GRF and showing the interactions between the external forces and the hip and knee NJM demands. This was considered important so that the reader can better visualize the lateral force demands in the BBS and how it increases with width.

In terms of comparing to previously published studies on controlled versions of the WBBS and NBBS, we succeeded in confirming previously made observations on GM and VL sEMG activity (McGaw & Melrose 1999, Paoli et al. 2009) kinematic values, and lower lumbar, 3-D hip and knee extensor NJM (Wretenberg et al. 1996, Swinton et al. 2012). The lower lumbar NJM results matched with previous literature on comparisons between WBBS and NBBS. The similarities were that the lower lumbar extensor NJM were the highest NJM compared to hip and knee extensor NJM in both the NBBS and WBBS. Also, lower lumbar NJM were found to be higher in the NBBS and that lower lumbar NJM increased with load in both widths. Outside of the scope of width comparisons, it was important that the movement patterns compared in this study were as similar as possible to each other and therefore standardization was essential. A proper familiarization protocol, the same bar position, depth, tempo, relative load standards and same footwear standards kept the comparisons more appropriate to interpret.

7.8 Practical applications

Although the WBBS reached higher activity in the hamstrings, it’s activity was only in low levels, therefore as an isolated fact, this should not have substantial practical relevance. But considering the combined effect of the biomechanical differences the WBBS should be considered a viable option in many strength & conditioning scenarios. These biomechanical differences found in this study include increased 3-D hip-to-knee NJM ratio, increased 3-D hip NJM demands, increased knee adduction NJM demands (due to the increased lateral force demands), increased GM and hamstring activity while not reducing quadriceps activity, and not increasing the demands on the lower lumbar. Specifically, the WBBS to parallel depth with strict movement pattern standards (anterior knee restriction) should not be considered as a complete replacement for the NBBS. Rather, the WBBS could provide an alternative multidimensional benefit in long-term programming in athletic populations that utilize the bilateral squat for general preparedness training. These added stimuli might have the largest impact for athletes in multidirectional sports, especially when aiming to functionally strengthen the posterior chain through means of a triple extension movement pattern, while not neglecting the knee extensors. Long-term programs comparing adaptations between NBBS and WBBS with strict technical demands are needed to confirm long-term functional benefits.

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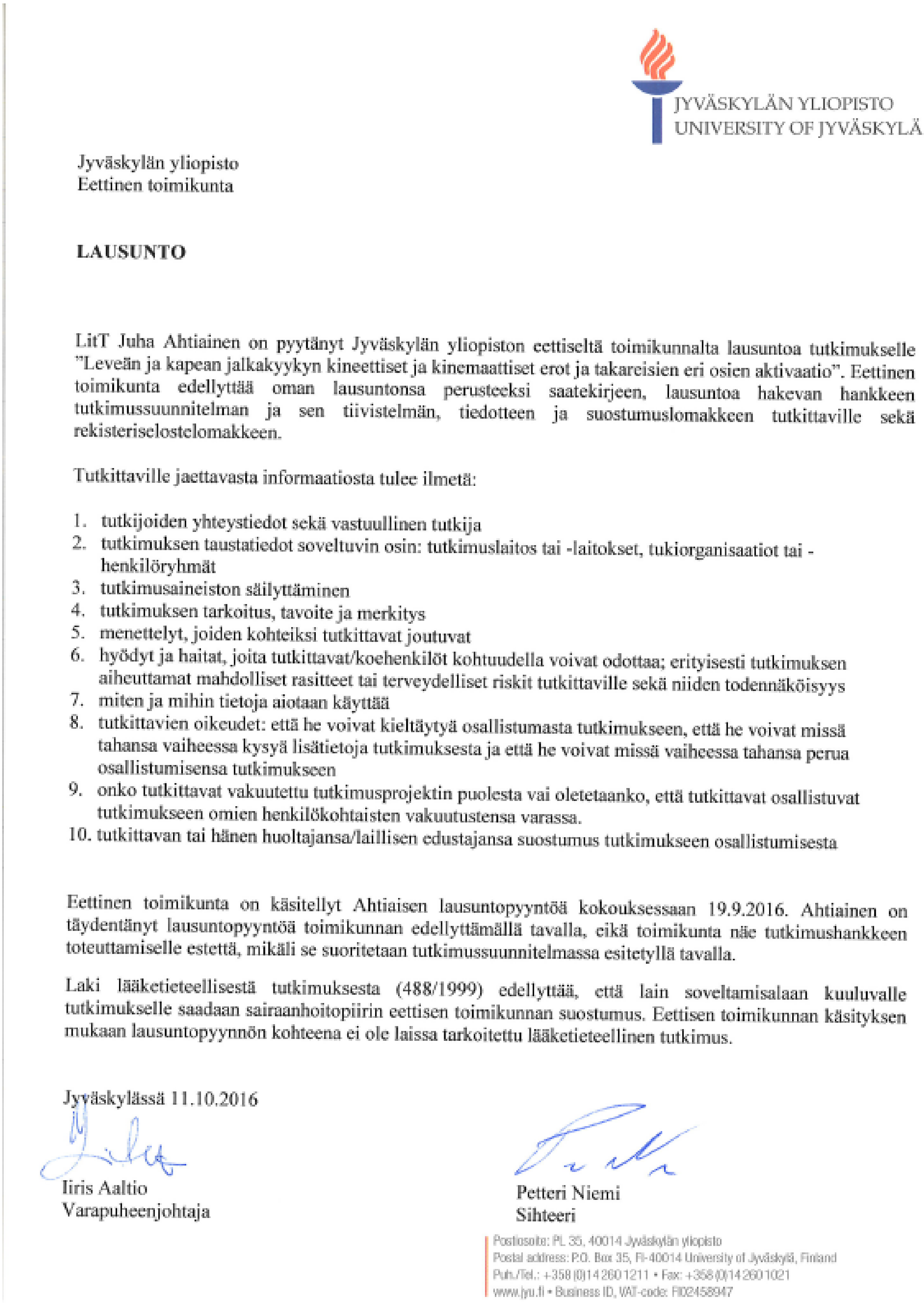
Appendix A: Ethical approval



JYVÄSKYLÄN YLIOPISTO
UNIVERSITY OF JYVÄSKYLÄ

Jyväskylän yliopisto
Eettinen toimikunta

LAUSUNTO

LitT Juha Ahtiainen on pyytänyt Jyväskylän yliopiston eettiseltä toimikunnalta lausuntoa tutkimukselle "Leveän ja kapean jalkakyykyn kineettiset ja kinemaattiset erot ja takareisien eri osien aktivaatio". Eettinen toimikunta edellyttää oman lausuntonsa perusteeksi saatekirjeen, lausuntoa hakevan hankkeen tutkimussuunnitelman ja sen tiivistelmän, tiedotteen ja suostumuslomakkeen tutkittaville sekä rekisteriselostelomakkeen.

Tutkittaville jaettavasta informaatiosta tulee ilmetä:

1. tutkijoiden yhteystiedot sekä vastuullinen tutkija
2. tutkimuksen taustatiedot soveltuvin osin: tutkimuslaitos tai -laitokset, tukiorganisaatiot tai -henkilöryhmät
3. tutkimusaineiston säilyttäminen
4. tutkimuksen tarkoitus, tavoite ja merkitys
5. menettelyt, joiden kohteiksi tutkittavat joutuvat
6. hyödyt ja haitat, joita tutkittavat/koehenkilöt kohtuudella voivat odottaa; erityisesti tutkimuksen aiheuttamat mahdolliset rasitteet tai terveydelliset riskit tutkittaville sekä niiden todennäköisyys
7. miten ja mihin tietoja aiotaan käyttää
8. tutkittavien oikeudet: että he voivat kieltäytyä osallistumasta tutkimukseen, että he voivat missä tahansa vaiheessa kysyä lisätietoja tutkimuksesta ja että he voivat missä vaiheessa tahansa perua osallistumisensa tutkimukseen
9. onko tutkittavat vakuutettu tutkimusprojektin puolesta vai oletetaanko, että tutkittavat osallistuvat tutkimukseen omien henkilökohtaisten vakuutustensa varassa.
10. tutkittavan tai hänen huoltajansa/laillisen edustajansa suostumus tutkimukseen osallistumisesta

Eettinen toimikunta on käsitellyt Ahtiaisen lausuntopyyntöä kokouksessaan 19.9.2016. Ahtiainen on täydentänyt lausuntopyyntöä toimikunnan edellyttämällä tavalla, eikä toimikunta näe tutkimushankkeen toteuttamiselle estettä, mikäli se suoritetaan tutkimussuunnitelmassa esitetyllä tavalla.

Laki lääketieteellisestä tutkimuksesta (488/1999) edellyttää, että lain soveltamisalaan kuuluvalle tutkimukselle saadaan sairaanhoitopiirin eettisen toimikunnan suostumus. Eettisen toimikunnan käsityksen mukaan lausuntopyynnön kohteena ei ole laissa tarkoitettu lääketieteellinen tutkimus.

Jyväskylässä 11.10.2016

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Appendix B: Informed Consent Documents (Finnish)

JY/EETTINEN TOIMIKUNTA

TIEDOTE TUTKITTAVILLE JA SUOSTUMUS TUTKIMUKSEEN OSALLISTUMISESTA

Tutkimuksen nimi: ”Leveän ja kapean jalkakyykyn kineettiset ja kinemaattiset erot ja takareisien eri osien aktivaatio”.

1. Tutkimuksen taustatiedot

Tutkimuksen mittaukset suoritetaan Jyväskylän Yliopiston liikuntabiologian laitoksella. Tutkimus on Johan Lahden Pro Gradu-tutkielma, jonka ohjaajana toimii yliopistotutkija Juha Ahtiainen. Tutkimuksesta pyritään tekemään myös kansainvälinen tieteellinen artikkeli. Tutkimus alkaa lokakuussa 2016 neljän viikon perehdytysjaksolla ja päättyy Marraskuun 2016 lopussa tutkittavien osalta.

2. Tutkimuksen tarkoitus, tavoite ja merkitys

Huippu-urheilussa kyse on aina pienistä marginaaleista. Näin ollen tieto yksityiskohdista kaikista käytetyistä työkaluista mm. voimaliikkeissä on arvokasta tietoa tukemaan urheilijan pitkäaikaista polkua. Tämä tutkimuksen tavoite on tuoda arvokasta tietoa eri urheilulajien valmentajille millaisissa tilanteissa voisi käyttää leveää ja kapeaa takakyykkyä fyysisen valmiuden kehittämiseksi.

3. Tutkimusaineiston käyttötarkoitus, käsittely ja säilyttäminen

Tässä tutkimuksessa saatua aineistoa käytetään vain tutkimuskäyttöön. Tutkimusryhmä vastaa tutkimusaineiston turvallisesta säilyttämisestä ilman tunnistetietoja siten, etteivät tiedot tutkimuksesta ja yksittäisistä tutkittavista päädy ulkopuolisille. Tutkimusta varten kerätty aineisto on muiden tutkijoiden käytössä ainoastaan erikseen sovittavalla tavalla. Eettisen toimikunnan lausuntoa haettaessa on täytetty henkilötietolain mukainen tieteellisen tutkimuksen rekisteriseloste. Tutkimuksessa kerätty manuaalinen aineisto säilytetään liikuntabiologian laitoksen laboratoriossa lukituissa tiloissa ja ATK:lla oleva aineisto säilytetään tutkijan omalla salasanalla suojatulla tietokoneella.

Henkilöitä koskeva tieto muokataan siten, että siitä ei voi tunnistaa osallistujan henkilöllisyyttä. Tutkimuksen johtaja arkistoi tutkimusaineiston pysyvästi ja säilyttää tunnistetiedot aineistosta erillään.

4. Menettelyt, joiden kohteeksi tutkittavat joutuvat

Koehenkilöt suorittavat tutkimuksen aikana takakyykky harjoitukset muun normaalin urheilun ohella. Koehenkilöt saattavat kokea lievää lihaskipua sekä uupumusta suorituksen yhteydessä. Kaikki perehdytykseen liittyvät tapaamiset (vähintään kuusi kolmen viikon aikana), jossa takakyykyn suoritusta harjoitellaan eivät pitäisi vaikuttaa negatiivisesti muihin harjoituksiin ja peleihin, koska harjoitusten volyymi on pieni ja takakyykky on muutenkin osana monen rugbypelaajan koko kauden treeniohjelmaa. Ainoa rasittavampi päivä tulee olevan maksimivoimatestin päivä, mutta koska suoritukset ovat teknisiä maksimeita, niin rasitus on myös hieman tavallista maksimivoimaharjoittelua pienempi. Tekninen maksimi tarkoittaa sitä, että jos tekniikka muuttuu lainkaan sitä, mitä se oli lämmittelysarjoissa (esim. selän neutraali asento ei pysy, lantio nousee selkeästi ensin ym.) toisto hylätään, vaikka paino nousisikin ylös. Teknisen maksimipäivän jälkeen on 5-8 päivän tauko ennen varsinaisia mittauksia. Itse mittauksessa käytetään 70% kuormaa maksimista, joten rasitus ei ole suuri. Mittauksen kokonaiskesto on noin 3 tuntia

5. Tutkimuksen liittyvät testit ja toimenpiteet

Tutkimus sisältää erilaisia ja eri ominaisuuksia mittaavia testejä. Tutkimuksessa suoritetaan neljä sarjaa ja neljä toistoa 70% kuormalla maksimista sekä leveässä, että kapeassa takakyykyssä. Varsinaisia mittauspäiviä on vain yksi per koehenkilö perehdytysjakson jälkeen. Perehdytysjakso kestää noin 3-4 viikkoa. Tämän aikana järjestetään jokaiselle tutkittavalle vähintään 6 valmennusta ja keskimäärin 1-2 tapaamista per viikko. Perehdytysjakso sisältää nousujohteisesta yksilöllistä valmennusta leveää ja kapea takakyykkyä varten enintään 6 urheilijan ryhmissä. Tapaamisten kestoksi arvioidaan noin 45 minuuttia ja ne voi yhdistää urheilijan omaan salitreeniin. Viimeisellä perehdytysjakson kerralla haetaan teknistä maksimia sekä kapeassa, että leveässä takakyykyssä. Tämä tarkoittaa sitä, että jos tekniikka muuttuu lainkaan sitä, mitä se oli lämpösarjoissa (esim. selän neutraali asento ei pysy, lantio nousee selkeästi ensin ym.) toisto on hylätty, vaikka paino tulee ylös. Maksimipäivän jälkeen on noin 5-8 päivää taukoa ennen mittauspäivää, jolloin tehdään kolme maksimaalista isometristä supistusta polven koukistajille takareisien lihasten normalisointia varten, minkä jälkeen siirrytään itse kyykyn mittausprotokollaan. Se sisältää lämmittelysarjojen jälkeen 4x4 toistoa 70% kuormalla, 3-1-XX tahdilla (3 sekuntia alas, yhden sekunnin pito vaakatasossa, räjähtävästi ylös).

6. Tutkimuksen hyödyt ja haitat tutkittaville

Tutkittavat saavat arvokasta tietoa takakyykyn käytöstä omaan fyysisen valmiuden kehittämiseen ja uutta arvokasta tieteellistä tietoa itse tutkimustuloksista.

Tutkimuksessa käytettävät menetelmät ovat pääasiallisesti turvallisia hyvällä suoritustekniikalla toteutettuna. Kuten fyysiseen harjoitteluun ylipäätään, myös tutkimusmittauksiin liittyy loukkaantumisriski, joka on kuitenkin hyvin vähäinen. Maksimaaliset voimatestit aiheuttavat tilapäisesti väsymystä. Maksimitesteissä saattaa tulla vähäisiä lihasvaurioita, jotka voivat aiheuttaa tilapäistä lihasarkuutta kuormitusta seuraavina päivinä. Kovassa kuormituksessa on aina olemassa sydämen, verenkiertoelimistön ja/tai hengityselimistön ylikuormittumisen vaara, mutta terveellä ihmisellä vakavien häiriöiden todennäköisyys on pieni.

7. Miten ja mihin tutkimustuloksia aiotaan käyttää

Tutkimustuloksista valmistuu Johan Lahti Pro Gradu-tutkielma. Tutkimustuloksia tullaan julkaisemaan myös alan kansallisissa ja kansainvälisissä lehdissä ja kongresseissa. Koehenkilöille tiedotetaan tutkimustuloksista kun tutkimustulokset on analysoitu.

8. Tutkittavien oikeudet

Osallistuminen tutkimukseen on täysin vapaaehtoista. Tutkittavilla on tutkimuksen aikana oikeus kieltäytyä tutkimuksesta ja keskeyttää tutkimukseen osallistuminen missä vaiheessa tahansa ilman seuraamuksia. Tutkimuksen järjestelyt ja tulosten raportointi ovat luottamuksellisia. Tutkimuksessa saatavien tutkittavien henkilökohtaiset tiedot tulevat ainoastaan tutkittavan ja tutkijaryhmän käyttöön ja tulokset julkaistaan tutkimusraporteissa siten, ettei yksittäistä tutkittavaa voi tunnistaa.

Tutkittavilla on oikeus saada lisätietoa tutkimuksesta tutkijaryhmän jäseniltä missä vaiheessa tahansa. Tutkimuksesta on täytetty henkilötietolain edellyttämä rekisteriseloste, jonka tutkittava halutessaan saa tutkijoilta nähtäväkseen. Tutkittaville on selvitettävä heille ymmärrettävällä tavalla, selkokielisesti ja tarkasti se, millaisia heidän oikeutensa tutkimuksessa ovat, ja miten sen järjestelyissä, aineiston säilytyksessä ja tulosten raportoinnissa suojataan heidän yksityisyyttään ja henkilöllisyyttään.

9. Vakuutukset

Jyväskylän yliopiston henkilökunta ja toiminta on vakuutettu. Vakuutus sisältää potilasvakuutuksen, toiminnanvastuuvakuutuksen ja vapaaehtoisen tapaturmavakuutuksen

Tutkimuksissa tutkittavat (koehenkilöt) on vakuutettu tutkimuksen ajan ulkoisen syyn aiheuttamien tapaturmien, vahinkojen ja vammojen varalta. Tapaturmavakuutus on voimassa mittauksissa ja niihin välittömästi liittyvillä matkoilla. Tapaturman lisäksi korvataan vakuutetun erityisen ja yksittäisen voimanponnistuksen ja liikkeen välittömästi aiheuttama lihaksen tai jänteen venähdysvamma, johon on annettu lääkärinhoitoa 14 vuorokauden kuluessa vammautumisesta. Korvausta maksetaan enintään kuuden viikon ajan venähdysvamman syntymisestä. Voimanponnistuksen ja liikkeen aiheuttaman venähdysvamman hoitokuluina ei korvata magneettitutkimusta eikä leikkaustoimenpiteitä.

Tapaturmien ja sairastapausten välittömään ensiapuun mittauksissa on varauduttu tutkimusyksikössä. Laboratoriossa on ensiapuvälineet ja varusteet, joiden käyttöön henkilökunta on perehtynyt. Tutkittavalla olisi hyvä olla oma henkilökohtainen tapaturma/sairaus- ja henkivakuutus, koska tutkimusprojekteja varten vakuutusyhtiöt eivät myönnä täysin kattavaa vakuutusturvaa esim. Sairauskohtauksien varalta.

10.Mittauksiin valmistautuminen

Mittauksien edeltävät 5-8 päivää suositellaan alaraajojen voimaharjoitusten välttämistä. Mittausta edeltävät pari päivää suositellaan, että harjoitusten intensiteetti on matala ja alaraajojen kuormitusta vältetään. Vältä kofeiinipitoisia juomia edellisenä päivänä sekä testipäivänä. Muun nesteen nauttimisesta on huolehdittava, jotta testin alkaessa et kärsi nestevajauksesta. Vältä raskasta ruokailua pari tuntia ennen testiä.

11.Suostumus

LEVEÄN JA KAPEAN TAKAKYYKYN KINEETTISET JA KINEMAATTISET EROT JA TAKAREISIEN ERI OSIEN AKTIVAATIO.

TUTKITTAVAN SUOSTUMUS TUTKIMUKSEEN OSALLISTUMISESTA

Olen perehtynyt tämän tutkimuksen tarkoitukseen ja sisältöön, kerättävän tutkimusaineiston käyttöön, tutkittaville aiheutuviin mahdollisiin haittoihin sekä tutkittavien oikeuksiin ja vakuutusturvaan. Suostun osallistumaan tutkimukseen annettujen ohjeiden mukaisesti. En osallistu mittauksia, veri- ym. kokeita tai fyysistä rasitusta sisältäviin tutkimuksiin flunssaisena, kuumeisena, toipilaana tai muuten huonovointisena. Voin halutessani peruuttaa tai keskeyttää osallistumiseni tai kieltäytyä tutkimukseen osallistumisesta missä vaiheessa tahansa ilman, että siitä aiheutuu heille mitään haittaa. Tutkimustuloksiani ja kerättyä aineistoa saa käyttää ja hyödyntää sellaisessa muodossa, jossa yksittäistä tutkittavaa ei voi tunnistaa.

Suostun yllämainitun projektin mittauksiin annettujen ohjeiden mukaisesti Kyllä x Ei x
Annan luvan tulosteni käyttöön tutkimuksen raportoinnissa Kyllä x Ei x
Yhteystietoni saa sisällyttää liikuntabiologian laitoksen henkilörekisteriin ja minuun saa olla myöhemmin yhteydessä haettaessa tutkittavia liikuntabiologian laitoksen tutkimuksiin Kyllä x Ei x
Olen tutustunut suoritettaviin testeihin ja mittauksiin, ja olen ymmärtänyt mittausten tarkoituksen ja niihin liittyvät riski- ja hyötynäkökohdat. Kyllä x Ei x

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Päiväys Tutkittavan allekirjoitus

Päiväys Tutkittavan huoltajan allekirjoitus

Päiväys Tutkijan allekirjoitus

Appendix C: Health History Questionnaire (Finnish)

ESITIETO- JA TERVEYSKYSELY

JYVÄSKYLÄN YLIOPISTO

Nimi:________________________ Synt.aika:___________ Paino:_______kg Pituus:________cm

Takakyykky kokemus (Kuinka kauan olet käyttänyt kyykkyä, maksimit ym.)

Testauksen turvallisuuden kartoittamiseksi pyydämme sinua täyttämään oheisen terveyskyselyn. Tämä kysely on vapaaehtoinen, mutta testaamisen turvallisuuden varmistamiseksi tiedot ovat välttämättömät. (Puutteelliset tiedot johtavat ehdokkaan hylkäämiseen)

Oireet viimeisen 6 kk aikana: Oletko tuntenut...	Kyllä	Ei	En osaa sanoa
1. rintakipuja?			
2. rasitukseen liittyvää hengenahdistusta?			
3. huimausoireita?			
4. rytmihäiriötuntemuksia?			
5. harjoittelua estäviä kipuja liikuntaelimissä?			
Missä?			
6. ylikuormitus- tai stressioireita?			

Todetut sairaudet: Onko sinulla tai onko sinulla ollut jokin/joitakin seuraavista? (ympyröi)

01 sepelvaltimotauti
02 sydäninfarkti
03 kohonnut verenpaine
04 sydänläppävika
05 aivohalvaus
06 aivoverenkierron häiriö
07 sydämen rytmihäiriö
08 sydämentahdistin
09 sydänlihassairaus
10 syvä laskimotukos
11 muu verisuonisairaus
12 krooninen bronkiitti
13 keuhkolaajentuma
14 astma
15 muu keuhkosairaus
16 allergia
17 kilpirauhasen toimintahäiriö
18 diabetes
19 anemia
20 korkea veren kolesteroli
21 nivelreuma
22 nivelrikko, -kuluma
23 krooninen selkäsairaus
24 mahahaava
25 pallea-, nivus- tai napatyrä
26 ruokatorven tulehdus
27 kasvain tai syöpä
28 leikkaus äskettäin
29 mielenterveyden ongelma
30 tapaturma äskettäin
31 matala veren K tai Mg
32 kohonnut silmänpaine

33 näön tai kuulon heikkous	34 urheiluvamma äskettäin

muita sairauksia tai oireita, mitä:____

Lääkitys: Käytätkö jotain lääkitystä tai lääkeainetta säännöllisesti tai usein? 1 En 2 Kyllä, mitä:____
____

Tupakoitko 1 En 2 Kyllä

Onko Sinulla todettu synnynnäinen sydänvika? 1 Ei 2 Kyllä,

mikä:____
____

Onko lähisuvussasi todettu perinnöllisiä sydänsairauksia tai sydänperäisiä äkkikuolemia?
1 Ei 2 Kyllä

Kuumetta, flunssaista oloa tai muuten poikkeavaa väsymystä viimeisen viikon aikana:
1 Ei 2 Kyllä

Olen vastannut kysymyksiin rehellisesti parhaan tietämykseni mukaan,

Päivä________________ Allekirjoitus___________________________________

Appendix D: Result tables

TABLE 1. Comprehensive subject info. GT: Greater trochanter. GT is also divided with stance width. Wide stance strength/BM ratio: Relative strength (divided by athlete’s body mass) of the specific squat. Narrow to wide ratio: a divide between the 1 RM of NBBS and WBBS. Above 1 indicates higher absolute load for NBBS.

Subject code	**Age**	**Height (cm)**	**Weight (kg)**	**Lifting experience (years)**	**Leg length (cm)**	**GT width (cm)**	**Wide stance width (cm)**	**Narrow stance width (cm)**	**Wide stance/GT width**	**Narrow stance/GT width**	**Wide stance technical max (kg)**	**Wide stance strength/BM ratio**	**Narrow stance technical max (kg)**	**Narrow stance strength/BM ratio**	**Narrow to wide ratio**
1	26	159	53	4	81	31	46	30	1,48	0,97	77,5	1,46	82,5	1,56	1,06
2	26	164	65	2	82,5	32	47	31	1,47	0,97	67,5	1,04	65	1,00	0,96
3	25	177,5	68	3	93	33	50	32	1,52	0,97	87,5	1,29	85	1,25	0,97
4	28	164	58	4	83	31,5	46	29	1,46	0,92	75	1,29	80	1,38	1,07
5	27	170	64	5	90	32	48	31	1,50	0,97	102,5	1,60	102,5	1,60	1,00
6	28	167	70	4	85	32	48	30	1,50	0,94	100	1,43	105	1,50	1,05
7	24	164	63	2	81	30	47	31	1,57	1,03	70	1,11	75	1,19	1,07
8	24	166	67	1	87	32	47	32	1,47	1,00	67,5	1,01	67,5	1,01	1,00
9	26	190	108	5	103,5	40	61	40	1,53	1,00	150	1,39	145	1,36	0,98
10	31	178	107	7	91	32	51	33	1,59	1,03	155	1,45	162,5	1,59	1,10
11	35	184	102	4	94	36	54	35	1,50	0,97	122,5	1,20	125	1,23	1,02
12	22	183	93	2	96	39	55	38	1,41	0,97	140	1,51	130	1,40	0,93
13	27	184	115	6	96	34	55	34	1,62	1,00	157,5	1,37	147,5	1,28	0,94
14	34	188	101	6	96	32	54	34	1,69	1,06	145	1,44	135	1,34	0,93
AVR.	27,36	174,18	81,00	3,93	89,93	33,32	50,64	32,86	1,52	0,99	108,39	1,33	107,68	1,33	1,01
SD	3,71	10,43	21,86	1,77	6,89	2,97	4,52	3,13	0,07	0,04	35,39	0,18	32,79	0,19	0,06

TABLE 2. Kinematics results.

SQUAT WIDTH										
Kinematics 70%	**NBBS 70%**	**WBBS 70%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip flexion	107.1 ± 5.6	109.1 ± 4.9	-2,0732	3,6821	-4,7073	0,5608	-1,781	9	0,109	0,56
Knee flexion	105.8 ± 2.8***	96.1 ± 3.9	9,737	3,249	7,4128	12,0612	9,477	9	<0.0001	3,00
Hip abduction	18.5 ± 4.1	24.3 ± 3.4***	-5,774	2,8933	-7,8438	-3,7042	-6,311	9	<0.0001	2,00
Hip internal rotation	15.7 ± 3.1	20.1 ± 3.7***	-4,485	1,8382	-5,8	-3,17	-7,716	9	<0.0001	2,44
Ankle dorsiflexion	29.8 ± 4.0***	18.6 ± 4.1	11,33	3,9017	8,5389	14,1211	9,183	9	<0.0001	2,90
Descent phase (s)	3.22 ± 0.29	3.27 ± 0.44	-0,052	0,37738	-0,32196	0,21796	-0,436	9	0,673	0,14
Ascent phase (s)	1.16 ± 0.20	1.10 ± 0.20	0,064	0,13125	-0,02989	0,15789	1,542	9	0,157	0,49
Kinematics 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip flexion	105.7 ± 6.1	107.79 ± 6.2*	-1,9995	2,778	-3,9868	-0,0123	-2,276	9	0,049	0,72
Knee flexion	104.4 ± 3.1***	95.5 ± 3.8	8,925	4,2963	5,8516	11,9984	6,569	9	<0.0001	2,08
Hip abduction	18.0 ± 4.3	23.1 ± 2.2**	-5,058	3,2175	-7,3596	-2,7564	-4,971	9	0,001	1,57
Hip internal rotation	15.6 ± 2.9	18.6 ± 3.5***	-3,103	1,7814	-4,3773	-1,8287	-5,508	9	<0.0001	1,74
Ankle dorsiflexion	29.4 ± 3.5***	21.2 ± 3.5	8,2275	2,6407	6,3384	10,1165	9,853	9	<0.0001	3,12
Descent phase (s)	3.36 ± 0.50	3.1 ± 0.48	0,051	0,24099	-0,12139	0,22339	0,669	9	0,52	0,21
Ascent phase (s)	1.36 ± 0.28	1.43 ± 0.37	-0,0652	0,1173	-0,14911	0,01871	-1,758	9	0,113	0,56
LOAD										
Kinematics	**NBBS 70%**	**NBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip flexion	107.1 ± 5.6	105.7 ± 6.1	1,285	2,672	-0,6264	3,1964	1,521	9	0,163	0,48
Knee flexion	105.8 ± 2.8	104.4 ± 3.1	1,443	2,9813	-0,6897	3,5757	1,531	9	0,16	0,48
Hip abduction	18.5 ± 4.1	18.0 ± 4.3	0,529	1,7502	-0,723	1,781	0,956	9	0,364	0,30
Hip internal rotation	15.7 ± 3.1	15.6 ± 2.9	0,109	1,925	-1,268	1,486	0,179	9	0,862	0,06
Ankle dorsiflexion	29.8 ± 4.0	29.4 ± 3.5	0,4495	1,9128	-0,9188	1,8179	0,743	9	0,476	0,23
Descent phase (s)	3.22 ± 0.29	3.36 ± 0.50	-0,138	0,41667	-0,43607	0,16007	-1,047	9	0,322	0,33
Ascent phase (s)	1.16 ± 0.20	1.36 ± 0.28**	-0,206	0,15182	-0,3146	-0,0974	-4,291	9	0,002	1,36

Kinematics	WBBS 70%	WBBS 85%	Mean	SD	95% CI of the Difference		t	df	Sig. (2-tailed)	Effect size (Cohens d)
					Lower	Upper				
Hip flexion	109.1 ± 4.9	107.8 ± 6.2	1,3587	3,6051	-1,2202	3,9376	1,192	9	0,264	0,38
Knee flexion	96.1 ± 3.9	95.5 ± 3.8	0,631	2,3995	-1,0855	2,3475	0,832	9	0,427	0,26
Hip abduction	24.3 ± 3.4	23.1 ± 2.2	1,245	2,5322	-0,5664	3,0564	1,555	9	0,154	0,49
Hip internal rotation	20.1 ± 3.7	19.1 ± 3.1	1,091	1,8868	-0,2587	2,4407	1,829	9	0,101	0,58
Ankle dorsiflexion	18.6 ± 4.1	21.2 ± 3.5	-2,653	4,6157	-5,9549	0,6489	-1,818	9	0,102	0,57
Descent phase (s)	3.27 ± 0.4	3.1 ± 0.5	-0,035	0,59151	-0,45814	0,38814	-0,187	9	0,856	0,06
Ascent phase (s)	1.10 ± 0.2	1.43 ± 0.4**	-0,3352	0,2922	-0,54423	-0,12617	-3,628	9	0,006	1,15
* = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001.										

TABLE 3. NJM results

SQUAT WIDTH										
NJM 70%	**NBBS 70%**	**WBBS 70%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
3-D L5/SI	4.32 ± 1.03	4.23 ± 0.94	0,07603	0,4311	-0,21359	0,36565	0,585	9	0,572	0,18
L5/SI sagittal (ext)	4.14 ± 1.00	4.20 ± 1.03	-0,05683	0,59284	-0,48092	0,36727	-0,303	9	0,769	0,10
L5/SI frontal (abd)	0.94 ± 0.61	0.84 ± 0.30	0,09389	0,35353	-0,15901	0,34678	0,84	9	0,423	0,27
L5/SI transverse (rotation)	0.49 ± 0.14	0.47 ± 0.16	0,01752	0,17187	-0,10542	0,14047	0,322	9	0,754	0,10
3-D Hip	4.26 ± 0.90	4.56 ± 0.89**	-0,3044	0,29151	-0,51293	-0,09587	-3,302	9	0,009	1,04
Hip sagittal (ext)	3.73 ± 0.68	4.06 ± 0.71*	-0,32855	0,45277	-0,65244	-0,00466	-2,295	9	0,047	0,73
Hip frontal (abd)	1.82 ± 0.76	1.95 ± 0.81*	-0,13608	0,13817	-0,23492	-0,03724	-3,114	9	0,012	0,98
Hip transverse (int rotation)	0.77 ± 0.32	0.94 ± 0.35*	-0,16679	0,16623	-0,2857	-0,04788	-3,173	9	0,011	1,00
3-D Knee	2.96 ± 0.66	2.84 ± 0.62	0,12275	0,17868	-0,00507	0,25057	2,172	9	0,058	0,69
Knee sagittal (ext)	2.57 ± 0.57**	2.29 ± 0.61	0,28544	0,22796	0,12236	0,44851	3,96	9	0,003	1,25
Knee frontal (add)	1.34 ± 0.57	1.65 ± 0.54**	-0,3049	0,27252	-0,49985	-0,10995	-3,538	9	0,006	1,12
Knee transverse (int rotation)	0.31 ± 0.14	0.35 ± 0.19	-0,04492	0,08905	-0,10863	0,01878	-1,595	9	0,145	0,50
NJM 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**

					Lower	Upper				
3-D L5/SI	4.84 ± 1.05*	4.57 ± 1.10	0,27054	0,26472	0,08118	0,45991	3,232	9	0,01	1,02
L5/SI sagittal (ext)	4.72 ± 1.02	4.41 ± 1.27	0,3061	0,60752	-0,12849	0,7407	1,593	9	0,146	0,50
L5/SI frontal (abd)	0.89 ± 0.30	0.78 ± 0.27	0,10934	0,48652	-0,23869	0,45738	0,711	9	0,495	0,22
L5/SI transverse (rotation)	0.50 ± 0.15	0.46 ± 0.18	0,04449	0,20487	-0,10207	0,19104	0,687	9	0,51	0,22
3-D Hip	4.66 ± 0.91	4.91 ± 0.98*	-0,24907	0,2412	-0,42161	-0,07653	-3,265	9	0,01	1,03
Hip sagittal (ext)	4.08 ± 0.65	4.36 ± 0.74*	-0,28271	0,32927	-0,51825	-0,04716	-2,715	9	0,024	0,86
Hip frontal (abd)	1.97 ± 0.85	2.07 ± 0.87	-0,09976	0,23582	-0,26846	0,06894	-1,338	9	0,214	0,42
Hip transverse (int rotation)	0.85 ± 0.40	1.01 ± 0.34*	-0,16	0,19194	-0,2973	-0,0227	-2,636	9	0,027	0,83
3-D Knee	3.13 ± 0.68*	3.00 ± 0.63	0,13078	0,13036	0,03752	0,22404	3,172	9	0,011	1,00
Knee sagittal (ext)	2.66 ± 0.62*	2.39 ± 0.59	0,26899	0,26326	0,08066	0,45732	3,231	9	0,01	1,02
Knee frontal (add)	1.46 ± 0.60	1.77 ± 0.62*	-0,3128	0,30777	-0,53296	-0,09263	-3,214	9	0,011	1,02
Knee transverse (int rotation)	0.36 ± 0.14	0.39 ± 0.18	-0,03549	0,07809	-0,09135	0,02038	-1,437	9	0,185	0,45

LOAD

NJM	NBBS 70%	NBBS 85%	Mean	SD	95% CI of the Difference		t	df	Sig. (2-tailed)	Effect size (Cohens d)
					Lower	Upper				
3-D L5/SI	4.32 ± 1.03	4.84 ± 1.05***	-0,52513	0,28087	-0,72605	-0,32421	-5,912	9	<0.0001	1,87
L5/SI sagittal (ext)	4.14 ± 1.00	4.72 ± 1.02**	-0,57315	0,34289	-0,81844	-0,32786	-5,286	9	0,001	1,67
L5/SI frontal (abd)	0.94 ± 0.61	0.89 ± 0.30	0,03773	0,28634	-0,1671	0,24257	0,417	9	0,687	0,13
L5/SI transverse (rotation)	0.49 ± 0.14	0.50 ± 0.15	-0,0181	0,13614	-0,11549	0,07929	-0,42	9	0,684	0,13
3-D Hip	4.26 ± 0.90	4.66 ± 0.91***	-0,39756	0,19527	-0,53725	-0,25787	-6,438	9	<0.0001	2,04
Hip sagittal (ext)	3.73 ± 0.68	4.08 ± 0.65**	-0,35242	0,21435	-0,50576	-0,19908	-5,199	9	0,001	1,64
Hip frontal (abd)	1.82 ± 0.76	1.97 ± 0.85**	-0,14652	0,13356	-0,24206	-0,05097	-3,469	9	0,007	1,10
Hip transverse (int rotation)	0.77 ± 0.32	0.85 ± 0.40	-0,07345	0,10424	-0,14802	0,00112	-2,228	9	0,053	0,70
3-D Knee	2.96 ± 0.66	3.13 ± 0.68***	-0,16871	0,08114	-0,22675	-0,11066	-6,575	9	<0.0001	2,08
Knee sagittal (ext)	2.57 ± 0.57	2.66 ± 0.62	-0,08352	0,14592	-0,1879	0,02087	-1,81	9	0,104	0,57
Knee frontal (add)	1.34 ± 0.57	1.46 ± 0.60**	-0,11335	0,08053	-0,17096	-0,05574	-4,451	9	0,002	1,41
Knee transverse (int rotation)	0.31 ± 0.14	0.36 ± 0.14**	-0,04989	0,04705	-0,08355	-0,01623	-3,353	9	0,008	1,06
NJM	**WBBS 70%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
3-D L5/SI	4.23 ± 0.94	4.57 ± 1.10*	-0,34002	0,34323	-0,58556	-0,09449	-3,133	9	0,012	0,99

L5/SI sagittal (ext)	4.20 ± 1.03	4.41 ± 1.27	-0,21022	0,56276	-0,61279	0,19236	-1,181	9	0,268	0,37
L5/SI frontal (abd)	0.84 ± 0.30	0.78 ± 0.27	0,05319	0,3959	-0,23002	0,3364	0,425	9	0,681	0,13
L5/SI transverse (rotation)	0.47 ± 0.16	0.46 ± 0.18	0,00886	0,1384	-0,09014	0,10787	0,203	9	0,844	0,06
3-D Hip	4.56 ± 0.89	4.91 ± 0.98**	-0,34223	0,21387	-0,49522	-0,18924	-5,06	9	0,001	1,60
Hip sagittal (ext)	4.06 ± 0.71	4.36 ± 0.74**	-0,30658	0,18972	-0,4423	-0,17086	-5,11	9	0,001	1,62
Hip frontal (abd)	1.95 ± 0.81	2.07 ± 0.87	-0,1102	0,18386	-0,24172	0,02132	-1,895	9	0,091	0,60
Hip transverse (int rotation)	0.94 ± 0.35	1.01 ± 0.34	-0,06666	0,14671	-0,17161	0,03829	-1,437	9	0,185	0,45
3-D Knee	2.84 ± 0.62	3.00 ± 0.63**	-0,16067	0,12536	-0,25035	-0,071	-4,053	9	0,003	1,28
Knee sagittal (ext)	2.29 ± 0.61	2.39 ± 0.59*	-0,09996	0,13861	-0,19912	-0,00081	-2,281	9	0,049	0,72
Knee frontal (add)	1.65 ± 0.54	1.77 ± 0.62	-0,12124	0,18812	-0,25582	0,01333	-2,038	9	0,072	0,64
Knee transverse (int rotation)	0.35 ± 0.19	0.39 ± 0.18	-0,04045	0,06076	-0,08391	0,00301	-2,105	9	0,065	0,67
* = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001.										

TABLE 4. HD-sEMG results

SQUAT WIDTH										
HD-sEMG 70%	**NBBS 70%**	**WBBS 70%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
BFLH overall	31.7 ± 6.9	33.74 ± 8.1	-2,702	4,768	-6,688	1,284	-1,603	7	0,153	0,57
BFLH distal	30.2 ± 7.9	32.91 ± 9.3	-2,697	4,395	-6,371	0,977	-1,736	7	0,126	0,61
BFLH medial	33.12 ± 8.3	35.1 ± 8.4	-1,991	5,004	-6,174	2,193	-1,125	7	0,298	0,40
BFLH proximal	33.9 ± 10.2	36.9 ± 9.5*	-4,059	4,346	-7,692	-0,426	-2,642	7	**0,033**	**0,93**
ST overall	26.2 ± 3.3	27.6 ± 3.7	-1,4	5,814	-8,619	5,819	-0,538	4	0,619	0,24
ST distal	25.0 ± 3.8	28.0 ± 8.7	-3	5,701	-10,079	4,079	-1,177	4	0,305	0,53
ST medial	25.2 ± 7.5	25.8 ± 7.2	-0,6	5,857	-7,872	6,672	-0,229	4	0,83	0,10
ST proximal	27.4 ± 6.3	27.8 ± 3.6	-0,4	5,413	-7,121	6,321	-0,165	4	0,877	0,07
HD-sEMG 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
BFLH overall	34.4 ± 7.4	37.2 ± 7.1**	-3,423	2,632	-5,623	-1,223	-3,679	7	**0,008**	**1,30**
BFLH distal	32.5 ± 9.4	36.0 ± 10.6	-3,548	4,283	-7,129	0,032	-2,343	7	0,052	0,83

BFLH medial	35.3 ± 9.9	38.3 ± 8.9*	-3,019	2,943	-5,48	-0,559	-2,901	7	**0,023**	**1,03**
BFLH proximal	36.1 ± 10.9	38.4 ± 9.4*	-2,344	2,608	-4,524	-0,163	-2,541	7	**0,039**	**0,90**
ST overall	27.0 ± 4.8	31.0 ± 2.3*	-4	2,646	-7,285	-0,715	-3,381	4	**0,028**	**1,51**
ST distal	26.2 ± 8.9	29.6 ± 8.1	-3,4	3,05	-7,187	0,387	-2,493	4	0,067	1,11
ST medial	25.8 ± 7.9	29.8 ± 6.5*	-4	3	-7,725	-0,275	-2,981	4	**0,041**	**1,33**
ST proximal	28.6 ± 4.5	33.0 ± 5.1*	-4,4	2,881	-7,977	-0,823	-3,415	4	**0,027**	**1,53**
LOAD										
HD-sEMG	**NBBS 70%**	**NBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
BFLH overall	31.7 ± 6.9	34.4 ± 7.4	-2,019	3,871	-5,255	1,218	-1,475	7	0,184	0,52
BFLH distal	30.2 ± 7.9	32.5 ± 9.4	-2,241	4,967	-6,393	1,911	-1,276	7	0,243	0,45
BFLH medial	33.12 ± 8.3	35.3 ± 9.9	-2,232	4,335	-5,856	1,392	-1,456	7	0,189	0,51
BFLH proximal	33.9 ± 10.2	36.1 ± 10.9*	-3,2	3,542	-6,162	-0,239	-2,555	7	**0,038**	**0,90**
ST overall	26.2 ± 3.3	27.0 ± 4.8	-0,8	5,891	-8,114	6,514	-0,304	4	0,777	0,14
ST distal	25.0 ± 3.8	26.2 ± 8.9	-1,2	6,496	-9,266	6,866	-0,413	4	0,701	0,18
ST medial	25.2 ± 7.5	25.8 ± 7.9	-0,6	6,877	-9,14	7,94	-0,195	4	0,855	0,09
ST proximal	27.4 ± 6.3	28.6 ± 4.5	-1,2	4,266	-6,497	4,097	-0,629	4	0,563	0,28
HD-sEMG	**WBBS 70%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
BFLH overall	33.74 ± 8.1	37.2 ± 7.1*	-2,739	3,007	-5,253	-0,226	-2,577	7	**0,037**	**0,91**
BFLH distal	32.91 ± 9.3	36.0 ± 10.6*	-3,093	3,451	-5,978	-0,207	-2,535	7	**0,039**	**0,90**
BFLH medial	35.1 ± 8.4	38.3 ± 8.9	-3,26	4,377	-6,92	0,399	-2,107	7	0,073	0,74
BFLH proximal	36.9 ± 9.5	38.4 ± 9.4	-1,485	2,729	-3,766	0,797	-1,539	7	0,168	0,54
ST overall	27.6 ± 3.7	31.0 ± 2.3*	-3,4	2,191	-6,12	-0,68	-3,47	4	**0,026**	**1,55**
ST distal	28.0 ± 8.7	29.6 ± 8.1	-1,6	3,13	-5,487	2,287	-1,143	4	0,317	0,51
ST medial	25.8 ± 7.2	29.8 ± 6.5**	-4	1,732	-6,151	-1,849	-5,164	4	**0,007**	**2,31**
ST proximal	27.8 ± 3.6	33.0 ± 5.1*	-5,2	3,194	-9,166	-1,234	-3,641	4	**0,022**	**1,63**
* = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001.										

TABLE 5. 2x3 anova for tegional interactions.

Squat type x region	Condition	P value	F value
BLFH	70 %	0,177	(1,7) = 1,966
	85 %	0,383	(1,7) = 0,689
ST	70 %	0,055	(1,4) = 4,232
	85 %	0,45	(1,4) = 0,884

TABLE 6. Relative peak external forces results.

SQUAT WIDTH										
Peak relative external force 70%	**NBBS 70%**	**WBBS 70%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Vertical force (%)	82.8 ± 1.8***	76.7 ± 2.7	6,0295	1,8271	4,7226	7,3365	10,436	9	<0.0001	3,30
Medial - lateral force (%)	13.7 ± 2.0	20.0 ± 2.7***	-6,4267	1,7422	-7,6731	-5,1804	-11,665	9	<0.0001	3,69
Anterior-posterior force (%)	3.6 ± 0.7	3.3 ± 0.7	0,3972	0,4689	0,0618	0,7326	2,679	9	0,051	0,85
Peak relative external force 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Vertical force (%)	82.7 ± 1.9***	76.3 ± 1.9	6,3858	0,869	5,7642	7,0074	23,238	9	<0.0001	7,35
Medial - lateral force (%)	13.9 ± 1.8	20.0 ± 1.6***	-6,0789	0,8707	-6,7017	-5,456	-22,079	9	<0.0001	6,98
Anterior-posterior force (%)	3.3 ± 0.6	3.6 ± 0.7	-0,3069	0,4461	-0,6261	0,0122	-2,176	9	0,058	0,69
LOAD										
Peak relative external force	**NBBS 70%**	**NBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Vertical force (%)	82.8 ± 1.8	82.7 ± 1.9	0,0893	0,6622	-0,3844	0,563	0,427	9	0,68	0,13
Medial - lateral force (%)	13.7 ± 2.0	13.9 ± 1.8	-0,2729	0,7742	-0,8267	0,2809	-1,115	9	0,294	0,35
Anterior-posterior force (%)	3.6 ± 0.7	3.3 ± 0.6	0,1836	0,5532	-0,2121	0,5793	1,049	9	0,321	0,33
Peak relative external force	**WBBS 70%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Vertical force (%)	76.7 ± 2.7	76.3 ± 1.9	0,4456	1,6745	-0,7523	1,6434	0,841	9	0,422	0,27

Medial - lateral force (%)	20.0 ± 2.7	20.0 ± 1.6	0,075	1,791	-1,2062	1,3562	0,132	9	0,898	0,04
Anterior-posterior force (%)	3.2 ± 0.8	3.6 ± 0.7	-0,5205	0,7	-1,0213	-0,0198	-2,351	9	0,053	0,74

= indicates small ES, ## = indicates medium ES, ### = indicates large ES (p<0.05).

TABLE 7. Hip-to-knee NJM ratios results.

SQUAT WIDTH										
Hip-to-knee NJM ratios 70%	**NBBS 70%**	**WBBS 70%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip-to-knee 3-D ratio	1.45 ± 0.19	1.63 ± 0.23**	-0,17462	0,17563	-0,30026	-0,04898	-3,144	9	0,012	0,99
Hip-to-knee extension ratio	1.48 ± 0.22	1.82 ± 0.26**	-0,34383	0,24661	-0,52024	-0,16741	-4,409	9	0,002	1,39
Hip-to-knee NJM ratios 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip-to-knee 3-D ratio	1.50 ± 0.14	1.64 ± 0.16**	-0,14421	0,11427	-0,22595	-0,06247	-3,991	9	0,003	1,26
Hip-to-knee extension ratio	1.57 ± 0.21	1.86 ± 0.23**	-0,28964	0,18896	-0,42482	-0,15447	-4,847	9	0,001	1,53
LOAD										
Hip-to-knee NJM ratios	**NBBS 70%**	**NBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip-to-knee 3-D ratio	1.45 ± 0.19	1.50 ± 0.14	-0,04654	0,06939	-0,09619	0,0031	-2,121	9	0,063	0,67
Hip-to-knee extension ratio	1.48 ± 0.22	1.57 ± 0.21	-0,09054	0,13953	-0,19035	0,00928	-2,052	9	0,07	0,65
Hip-to-knee NJM ratios	**WBBS 70%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
Hip-to-knee 3-D ratio	1.63 ± 0.23	1.64 ± 0.16	-0,01613	0,1193	-0,10148	0,06921	-0,428	9	0,679	0,14
Hip-to-knee extension ratio	1.82 ± 0.26	1.86 ± 0.23	-0,03635	0,189	-0,17155	0,09885	-0,608	9	0,558	0,19

* = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001.

TABLE 8. sEMG results.

SQUAT WIDTH										
sEMG 70%	**NBBS 70%**	**WBBS 70%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
GM	48.8 ± 20.2	59.5 ± 22.4*	-10,7878	7,7636	-17,9679	-3,6076	-3,676	6	0,01	1,39
VL	86.8 ± 11.5	84.4 ± 11.5	2,4794	17,7288	-16,1259	21,0846	0,343	5	0,746	0,14
sEMG 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
GM	65.0 ± 17.7	66.9 ± 20.7	-1,8427	4,7483	-6,2342	2,5487	-1,027	6	0,344	0,39
VL	87.3 ± 13.1	87.6 ± 10.8	-0,2986	12,465	-13,3798	12,7826	-0,059	5	0,955	0,02
LOAD										
sEMG 85%	**NBBS 85%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
GM	48.8 ± 20.2	65.0 ± 17.7*	-16,3426	13,5683	-28,8912	-3,7941	-3,187	6	0,019	1,20
VL	86.8 ± 11.5	87.3 ± 13.1	-0,5198	18,4522	-19,8842	18,8445	-0,069	5	0,948	8,00
sEMG	**WBBS 70%**	**WBBS 85%**	**Mean**	**SD**	**95% CI of the Difference**		**t**	**df**	**Sig. (2-tailed)**	**Effect size (Cohens d)**
					Lower	**Upper**				
GM	59.5 ± 22.4	66.9 ± 20.7	-7,3976	8,7236	-15,4656	0,6704	-2,244	6	0,066	0,85
VL	84.4 ± 11.5	87.6 ± 10.8	-3,2978	9,6384	-13,4126	6,817	-0,838	5	0,44	0,34
* = indicates statistically significant p<0.05, ** = indicates very statistically significant p<0.01, *** = indicates highly statistically significant p<0.001.										

TABLE 9. ICC and CV for all variables in squatting conditions.

Variables	**NBBS 70%**		**WBBS 70%**		**NBBS 85%**		**WBBS 85%**	
	ICC	CV	ICC	CV	ICC	CV	ICC	CV
Hip flexion	0,69	2,63	0,93	1,45	0,97	0,94	0,94	1,19
Hip abduction	0,89	7,81	0,68	6,46	0,88	6,68	0,72	4,32
Hip internal rotation	0,87	8,04	0,98	4,75	0,99	3,82	0,98	3,76
Knee flexion	0,68	2,17	0,81	2,54	0,91	0,97	0,06	4,31
Dorsi flexion	0,84	4,34	0,84	8,24	0,98	3,37	0,90	5,78
Ascent phase	0,69	11,13	0,88	5,65	0,96	4,70	0,70	11,35
Descent phase	0,41	11,56	0,64	8,59	0,91	4,18	0,58	9,54
3-D L5/SI	0,97	4,97	0,80	10,68	0,95	4,46	0,94	4,68
L5/SI sagittal	0,98	5,45	0,79	11,05	0,96	4,29	0,95	4,28
L5/SI frontal	0,84	23,71	0,71	25,40	0,80	15,78	0,59	23,47
L5/SI transverse	0,69	16,28	0,48	24,09	0,92	13,26	0,42	26,66
3-D Hip	0,98	3,55	1,00	1,38	0,98	2,55	0,97	2,44
Hip sagittal	0,98	3,52	0,99	1,70	0,96	2,60	0,98	2,32
Hip frontal	0,98	5,30	0,99	5,42	0,99	5,14	0,97	7,95
Hip transverse	0,93	9,68	0,99	4,89	0,97	7,80	0,90	9,49
3-D Knee	0,98	3,12	0,98	2,86	0,99	2,14	0,97	2,76
Knee sagittal	0,97	4,56	0,97	5,43	0,98	3,64	0,95	4,12
Knee frontal	0,99	2,78	0,99	4,45	0,98	5,31	0,96	5,67
Knee transverse	0,95	13,27	0,96	12,19	0,95	7,27	0,91	12,68
BFLH overall	0,92	7,04	0,85	7,77	0,61	15,18	0,79	9,39
BFLH distal	0,84	10,54	0,90	9,30	0,53	18,21	0,97	3,71
BFLH medial	0,87	9,95	0,85	9,48	0,60	19,18	0,88	7,17
BFLH proximal	0,85	9,76	0,77	9,51	0,52	13,36	0,43	16,00
ST overall	0,79	6,33	0,99	5,25	0,88	10,93	0,64	4,99
ST distal	0,79	9,83	0,98	6,58	0,85	14,94	0,99	4,45
ST medial	0,99	7,52	0,82	8,55	0,88	10,93	0,99	4,80
ST proximal	0,86	9,15	0,40	9,50	0,67	17,42	0,97	4,53
Vertical force	0,94	3,40	0,97	2,98	0,98	3,36	1,00	1,60
Medial-lateral force	0,91	4,28	0,96	4,35	0,97	5,30	0,98	3,47
Anterior-posterior force	0,96	9,68	0,89	10,37	0,71	16,34	0,58	17,26
3-D hip-to-knee	0,87	16,22	0,96	5,64	0,87	6,89	0,90	5,40
Hip-to-knee extensor	0,87	16,85	0,90	7,69	0,87	8,48	0,77	6,76
GM	0,86	22,97	0,95	9,04	0,85	9,83	0,96	7,83
VL	0,41	8,88	0,92	5,35	0,58	5,78	0,44	9,86