Electromyographical differences between the Hyperextension and Reverse-Hyperextension.

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ABSTRACT

The aims of this study were to compare muscle activation of the erector spinae (ES), gluteus maximus (GMax) and biceps femoris (BF) during the hyperextension (HE) and reverse-hyperextension (RHE) exercises. Ten subjects (age = 23±4 years, height = 175.9±6.9 cm; mass = 75.2±9.7 kg) had EMG electrodes placed on the ES, GMax and BF muscles in accordance with SENIAM guidelines. Subjects performed three maximum voluntary isometric contraction trials of lumbar extension and hip extension using a handheld and isokinetic dynamometer, respectively, in order to normalize the EMG during the HE and RHE. Three repetitions of each exercise were executed in a randomized order. High reliability (ICC ≥ 0.925) was observed with low variability (CV < 10%) in all but the GMax during the extension phase of the HE (CV = 10.64%). During the extension and flexion phases, the RHE exhibited significantly greater ($p \leq 0.024$; 34.1-70.7% difference) peak EMG compared to the HE in all muscles tested. Similarly, the RHE resulted in significantly greater mean EMG compared to the HE ($p \leq 0.036$; 28.2-65.0% difference) in all muscles except the BF during the flexion phase ($p = 9.960$). The RHE could therefore be considered as a higher intensity exercise for the posterior chain muscles compared to the HE, potentially eliciting greater increases in strength of the posterior chain muscles.

Key Words: Posterior Chain, Erectors, Hamstrings, Gluteals, EMG

INTRODUCTION

The primary hip extensor muscles (gluteals and hamstrings) form part of the posterior chain and are integral for force production to accelerate an individual's center of mass in a given direction, when performing athletic tasks (2, 4). Hip extension has therefore been highlighted as a key factor for sprinting (particularly from the stance to toe-off phase) (1, 2, 4, 18, 29, 34), jumping (1, 25, 34), and lateral movements such as side shuffles (1, 31, 34). The hip extensors are also essential for rapid force production during many deceleration actions as both a mechanism for injury risk reduction and to increase performance of such tasks. An example of the hamstrings, being a biarticular muscle, is to generate high forces, rapidly, to decelerate the shank during the late swing phase of the gait cycle, particularly in high speed running and sprinting (15, 16, 18), which is the point of the cycle at which hamstring strains are suggested to occur (5, 7, 9, 20, 32). Other examples include rapid deceleration during jump landings and during change of direction tasks, whereby the hip extensors are also understood to attenuate ground reaction forces around the knee, contribute to change of direction performance and improved landing mechanics (12, 26). Appropriate development of the trunk musculature may also contribute to positive enhancement of performance, particularly during change of direction tasks, via the efficient transfer of force generated from the lower body through the whole kinetic chain (11, 14, 27). Moreover, trunk musculature would also aid in lumbo-pelvic control, which has been identified as being particularly important to help avoid hamstring injury occurrence in high-speed running (30). It is therefore important to develop the posterior chain musculature, particularly in sports that involve both rapid accelerations and decelerations, and high-speed running, in order to maximize performance and potentially reduce the risk of injury.

Until recently (21), there has been limited investigation into both the hyperextension (HE), which is sometimes referred to as 90° hip extension, and reverse-hyperextension (RHE), despite these exercises anecdotally being used by competitive athletes. The RHE requires athletes to hang their lower body in a prone position from a padded platform in parallel to a pendulum whereby the feet are attached and the athletes can extend the hip whilst maintaining an extended knee, pulling their lower limbs up from ~90° hip flexion to ~0° hip flexion (see Figure 1). Within the aforementioned study (21), biomechanical differences, including muscle activation and both kinetic and kinematic variables, were calculated across 10 repetitions of both HE and RHE, both of which can typically be executed using the same piece of equipment. No significant differences were present between HE and RHE in peak and mean activation of the erector spinae (ES), gluteus maximus (GMax) and BF. Range of motion (ROM) around the trunk and pelvis was significantly greater during HE with ROM around the trunk and thigh greater in the RHE, which can be intuitively explained due to whether the lower or upper body is held in a fixed position during the HE and RHE exercises, respectively. The significantly greater ROM observed around the trunk and pelvis could be a contraindication, particularly if that ROM is occurring due to spinal flexion, as this may be putting undue pressure on the spine whilst also contradicting the desired bracing action around the trunk usually expected during resistance exercise. Peak and mean moments around the lower back were also significantly greater during the RHE. The difference in lower back moment could have simply been down to the change in lever length, with the majority of weight during the RHE being placed at the end of a pendulum, compared to the HE whereby the additional mass was held to their torso, resulting in a shorter lever. The only differences in muscle activation were the integrated electromyography (EMG), with significantly greater results in the GMax and BF during the HE. An explanation for the significant differences in integrated EMG could be due to the ballistic nature the exercises were performed. As mentioned, the tests were performed whilst participants executed bouts of 10 repetitions, described as using a cadence of 1:1 (1 second up and 1 second down). Keeping a 1:1 cadence within the HE would have meant constant tension and therefore greater time under tension within the muscles assessed, increasing integrated EMG. In contrast, during the RHE there may have been some a short period of reduced activity as the swinging of the pendulum caught up with the action of the lower body, particularly if a large amount of force (as demonstrated by the significantly greater moments within this exercise) is produced rapidly during the initial ROM, this creates momentum that could increase the reduction in activity as certain points of the movement. Whilst a ballistic approach can be viewed as ecologically valid, it is also important to understand what occurs during single repetitions, identifying both the 'concentric' and 'eccentric', or extension and flexion phases in order to know the most appropriate application of the exercise.

The aims of this study were, therefore, to assess differences in surface EMG of the ES, GMax and BF bilaterally during the extension and flexion phases of HE and RHE. It was hypothesized that peak and mean EMG would be greater overall in the RHE compared to the HE due to there being a greater lever length.

METHODS

Experimental Approach to the Problem

To compare differences in EMG, of the ES, GMax and BF muscles, between the HE and RHE exercises, an observational cross-sectional design was implemented whereby subjects performed three repetitions of both exercises in a randomized order. Prior to this, each subject performed three maximum voluntary isometric contractions MVIC during two different exercises (hip extension and back extension) to permit normalization of the EMG during the HE and RHE. Each exercise was divided into an extension and flexion phase, with a brief pause, when fully extended, to permit differentiation of the EMG signal between the two phases. The study was approved by the institutional review board (HST1718-019).

Subjects

Seven male and three female subjects (age = 23±4 years, height = 175.9±6.9 cm; mass = 75.2±9.7 kg) volunteered to participate in this investigation and provided written informed consent. All subjects had been resistance training recreationally for a minimum of 6 months prior to taking part in this investigation and were all familiar with the exercises.

Each subject had EMG electrodes (Noraxon Dual EMG electrode, Noraxon U.S.A. Inc, Scottsdale AZ, USA) placed on their ES, GMax and BF, in accordance with SENIAM (Surface EMG for Non-Invasive Assessment of Muscles) guidelines (13). A standardized protocol for the preparation of skin and application of electrodes was used to ensure stable contact and low skin impedance. This involved shaving of the skin, light abrasion and cleansing using alcohol wipes. Self-adhesive dual snap surface silver/silver chloride (Ag/AgCL) bipolar electrodes (Noraxon Dual EMG electrode, Noraxon USA. Inc, Scottsdale AZ, USA), were placed upon the muscle bellies. The electrode placement was parallel with the orientation of muscle fibres, in accordance with SENIAM guidelines. Wireless EMG sensors (2B EMG Sensor, Noraxon USA. Inc, Scottsdale AZ, USA) attached to the electrodes following correct placement and a quality check was performed. Live EMG data were transmitted via the wireless sensor to a receiver (Desktop DTS Receiver, Noraxon USA. Inc, Scottsdale AZ, USA) connected to a portable laptop running myomuscle software (MR3 Myomuscle, Noraxon USA. Inc, Scottsdale AZ, USA). All EMG data was collected at 1500 Hz.

Maximum Voluntary Isometric Contractions

Initially, subjects performed two MVICs including lumbar extension and prone hip extension to normalize dynamic EMG values. The prone hip extension was assessed using an isokinetic dynamometer (IKD) (Biodex Multi- Joint System 4 Isokinetic Dynamometer, New York, USA), with trials exhibiting a peak force within ±10% of the previous trial accepted as a maximal effort, unless there was a progressive increase in peak force. Joint centers were positioned at the point of rotation of the IKD with the pad placed above the distal portion of Achilles tendon for the hip extension. Hip extension trials were performed with full knee extension. The ES MVIC was assessed during a prone back extension, using a handheld dynamometer placed between the scapulae and with an adjustable strap providing a constant immovable resistance, with the same ±10% threshold applied to ensure maximal effort.

Specific verbal cues were provided for each MVIC, as appropriate cueing has been highlighted to have a positive difference in the timing and magnitude of contraction in gluteal and hamstring activation, during a prone hip extension (22). The instructions of 'raise your heel to the ceiling' and 'raise your chest towards the ceiling' were used for hip extension and back extension, respectively. Subjects were also instructed to contract their muscle as hard and as fast as possible for each trial, to enable achieve peak force (3). Two minutes rest was provided between each trial.

Exercise Performance

Subjects performed three repetitions of the HE and three repetitions of the RHE on a posterior chain developer (PowerLift, Iowa, USA), in a randomized order whereby all repetitions of the HE was followed by the RHE or vice versa, with a one-minute rest between repetitions. Subjects were instructed to remain as relaxed as possible prior to commencing the extension phase (Figure 1a & b), to pause at the end of this phase prior to being given a command to 'flex' to commence the flexion phase (Figure 1a & b), participants were instructed to perform a cadence of 1 second for both the extension and flexion phase through a full range of motion, which was comparable between exercises. At the end of each repetition subjects were asked to relax. Subjects being relaxed at the start of the extension phase and the end of the flexion permitted automated identification of the start and end of these phases, respectively, as described below. RHE trials however included a load attached to a swinging pendulum that was standardized to match the subject's upper body weight, including the torso, head and arms (62.9% of body mass), minus the weight of the pendulum arms and subjects legs (16.1% of bodyweight per leg) in accordance with the segmental model provided by Clauser, McConville and Young (8). This standardization of load allowed direct comparison between exercises.

INSERT FIGURE 1a & b NEAR HERE

Data Analysis

Analysis of EMG was performed using a bespoke Excel spreadsheet, calculating the mean and peak root mean squared (RMS) values during each phase of each exercise with a moving average window of 200 ms. The extension and flexion phases of the movement were identified as follows; onset and termination of movement, was assessed using a threshold of >2 standard deviations plus the mean of the EMG during a one second period of relaxation prior to and following movement, with the end of the extension and start of the flexion phases determined via a manual trigger during the exercise. Data were then expressed as a percentage of the peak EMG during the MVIC for both the peak and mean EMG during both the extension and flexion phases of the exercises.

2.4 Statistical Analyses

Normality of all data was determined via Shapiro-Wilk's test of normality. Within-session reliability was determined using two-way single measures random effects model ICC and 95% confidence intervals (CI) and interpreted based on the lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (19). Percentage coefficient of variation (%CV) and 95% CI was also calculated to determine the within session variability, with <10% classified as acceptable (10).

A series of paired samples t-Tests, or Wilcoxon's test for variables that did not meet parametric assumptions, and Hedges *g* effect sizes were calculated to determine if there were any significant or meaningful differences between exercise. Due to multiple comparisons subsequent Bonferroni corrections was also applied. An *a priori* alpha level was set at *p* ≤0.05 and effect sizes interpreted as trivial (≤0.19), small (0.20 – 0.59), moderate (0.60 – 1.19), large (1.20 – 1.99), and very large (≥2.0) (17). Statistical analyses were performed using SPSS (Version 23. IBM, New York, NY), with individual plots and Cumming estimation plots generated via www.estimationstats.com. Additionally, data are presented in Cumming estimation plots, with individual data and paired mean difference is plotted as a bootstrap sampling distribution and 95% CI.

RESULTS

Peak and mean EMG demonstrated good to excellent reliability (ICC ≥0.918, lower bound 95% CI >0.8) between repetitions (Figure 2) with acceptable variability (<10%) in all conditions excluding peak GMax and mean BF both during the extension phase of the HE (10.86% and 10.35%CV, respectively).

INSERT FIGURE 2 NEAR HERE

For all muscle groups, peak EMG during the extension phase was significantly greater activation during the RHE, with moderate magnitudes (*p* ≤0.024; *g* ≥0.95) (ES = 107.3±37.9%, GMax = 49.3±37.5%, and BF = 58.8±21.7%) compared to HE (ES = 73.1±25.4%, GMax = 18.1±13.1%, and BF = 38.7±18.6%). A similar pattern was also demonstrated during the flexion phase with RHE demonstrating significantly greater (*p* ≤0.024; *g* ≥1.04) peak EMG with moderate to large effect (ES = 101.6±37.1%, GMax = 52.6±33.6%, and BF = 58.9±21.3%) compared to the HE (ES = 59.2±29.1%, GMax = 15.4±9.8%, and BF ≤37.3±18.4%) (Figure 3).

INSERT FIGURE 3 NEAR HERE

Mean EMG followed a similar pattern for ES and GMax in both extension and flexion phases of the RHE (ES = 71.0±20.5% and 51.8±16.1%, GMax = 23.4±15.8% and 18.6±9.2%, respectively) exhibiting significant and moderate to large differences (*p* <0.001; *g* ≥1.03) compared to HE (ES = 43.9±18.3% and 30.5±11.3%, GMax = 8.2±4.5% and 9.1±6.5%) (Figure 4). During the extension phase the BF again elicited a significantly and moderately greater (*p* = 0.036; *g* = 0.75) mean EMG during the RHE (39.7±13.4%)

compared to HE (28.5±15.1%), the flexion phase; however, resulted in non-significant and small differences (*p* = 9.960; *g* = 0.30) between RHE and HE (28.3±2.3% and 22.4±13.3%, respectively) (Figure 4f).

INSERT FIGURE 4 NEAR HERE

DISCUSSION

The aims of this investigation were to assess the differences in surface EMG of the ES, GMax and BF during both extension and flexion phases of the HE and RHE. In agreement with our hypothesis, both the peak and mean EMG were greater during the RHE when compared to the HE. Moderate to large significantly greater differences were observed in peak EMG of all three muscles during both the extension and flexion phases of the RHE with mean EMG demonstrating similar results to peak EMG for the ES and GMax. Mean BF EMG showed small and non-significant differences during the flexion phase of the exercises. These results indicate that the RHE is likely a more effective exercise for training the posterior chain compared to the HE due to the EMG amplitudes elicited in both extension and flexion phases of the exercise.

During both the extension and flexion phases, the RHE elicits moderate to large significantly greater EMG amplitude in the ES and GMax. The biomechanical similarity between the two exercises is not reflected in the magnitude of the activation relative to the maximum capability of the muscle isometrically (the MVIC). The ES evidently produces the greatest percentage of MVIC, when performing the RHE, during both phases. When considering the implications of this in practice, it may suggest that in certain settings, such as rehabilitation from injury, caution should be taken in selecting the RHE due to the very high muscle activation and it therefore may be more appropriate to use the HE as a regression, prior to then progressing an athlete onto the RHE. A reason for the ES EMG values exceeding 100% of its MVIC could have potentially been do with the quality of normalization task chosen, with both GMax and BF normalization tasks being unilateral (hip extension and knee flexion) compared to of course a bilateral task (back extension) used for the ES, as well as a greater ROM during both the HE and RHE compared to the one static position held during the MVIC. Further to the differences in normalization tasks, there could have been preferential recruitment of the ES due to the technique adopted by the subjects. Macadam et al. (23) highlight how verbal and tactile cues increase GMax activation, therefore without any specific cues given during the exercise, the subjects could have been preferentially utilizing their ES and BF to a greater extent rather than achieving hip extension through gluteal activation. In comparison to a previous study examining the RHE, similar peak EMG values were seen in the ES as this was also above 100% of MVIC, in contrast however the HE also elicited ≥100% MVIC in the same study and with no significant difference present between the two (21). Another obvious difference between the study by Lawrence et al. (21) and the current study, is that the EMG of the GMax during both exercises, which exceed 100% MVIC on average, compared to the current study whereby the GMax only elicits ~50% MVIC. The difference between the two studies again could be down to normalization protocols, considering the Lawrence et al. (21) study utilized a MVIC at the top of the extension phase of the HE which could have been suboptimal for generating maximum contractions compared to separate tasks for each muscle group. The load used differed between the two studies also, with the current study using the equivalent of the subjects own upper body weight, while Lawrence et al. (21) used a similar calculation of upper body weight with the addition of a 20.4 kg weight plate during the HE and the equivalent factoring in the pendulum arm and lower mass of the lower body. The difference in load could also account for some of the increase in activation, as the GMax may have been required to a greater extent due to the ES and BF being overloaded.

The BF followed the same pattern as the ES and GMax in terms of peak EMG with the RHE demonstrating moderate, significantly greater activation during both phases. The only variable not to show a significant difference was in the mean EMG of the BF during the flexion phase, one reason for this could have been due to the subjects 'relaxing' the load, be it upper body weight or the equivalent lower limb plus the weight on the pendulum in a similar manner. In comparison to previous literature, BF activation during the HE much like that of the ES and GMax is lower when compared to Lawrence et al. (21). As previously mentioned, Lawrence et al. (21) used an additional 20.4 kg plate for the HE, which could account for a portion of the increase in activation as findings from Zebis et al. (33) showed a ~20% increase in BF activation when load was added to the HE. In comparison to similar hip extension-based exercises (such as, 45° hip extension), activation produced during the RHE was similar to that of a 45° hip extension (6), however caution must be taken when comparing EMG between studies due to variations in equipment, sampling frequencies, noise, amplification used and filtering processes (28).

The individual differences in EMG between subjects is demonstrated by the range in standard deviations and can be seen in Figure 3 and 4. A limitation of this study based upon the individual differences could be the heterogenous sample used, whilst they were all collegiate athletes, there was mixture of males and females of various body compositions. Due to this study utilizing surface EMG, there are some pitfalls of the equipment that include lower EMG recordings due to increased subcutaneous fat which can either reduce the detection or increase the cross-talk (24). Areas for future research include the effect of different cues on activation of the poster chain muscle groups during these exercises. Another potential area of future research is a comparison between unilateral RHE bilateral RHE particularly in ES activation due to the high levels of normalized EMG observed within this study. Following the identification of activation during these exercises, a training intervention should be applied to determine their adaptations to performance.

PRACTICAL APPLICATIONS

Based on the differences between the HE and RHE demonstrated within this study, practitioners should consider the RHE as a higher intensity exercise for the posterior chain muscles. Strengthening the hip extensors is important for improving different athletic tasks such as sprinting and jumping, with the findings of this study suggesting that it is likely the RHE would elicit greater increases in strength compared to the HE. It is also worth noting that due to the low level of activation observed (< 20% MVIC), without necessary coaching intervention and/or the addition of load, the HE is unlikely to stimulate the GMax to a high enough extent. In addition, without appropriate cueing the introduction of load could potentially increase the load on the spine rather than creating a greater stimulus for the GMax which practitioners should be careful of.

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References

1. Beardsley C and Contreras B. The increasing role of the hip extensor musculature with heavier compound lower-body movements and more explosive sport actions. *Strength and Conditioning Journal* 36: 50-55, 2014.
2. Belli A, Kyröläinen H, and Komi PV. Moment and power of lower limb joints in running. *Int J Sports Med* 23: 136-141, 2002.
3. Bemben MG, Clasey JL, and Massey BH. The effect of the rate of muscle contraction on the force-time curve parameters of male and female subjects. *Res Q Exerc Sport* 61: 96-99, 1990.
4. Blazevich AJ. Optimizing hip musculature for greater sprint running speed. *Strength and Conditioning Journal* 22: 22-27, 2000.
5. Bourne MN, Timmins RG, Opar DA, Pizzari T, Ruddy JD, Sims C, Williams MD, and Shield AJ. An Evidence-Based Framework for Strengthening Exercises to Prevent Hamstring Injury. *Sports Med* 48: 251-267, 2018.
6. Bourne MN, Williams MD, Opar DA, Al Najjar A, Kerr GK, and Shield AJ. Impact of exercise selection on hamstring muscle activation. *Br J Sports Med* 51: 1021-1028, 2017.
7. Chumanov ES, Schache AG, Heiderscheit BC, and Thelen DG. Hamstrings are most susceptible to injury during the late swing phase of sprinting. *Br J Sports Med* 46: 90, 2012.
8. Clauser CE, McConville JT, and Young JW. Weight, volume, and center of mass of segments of the human body. Antioch Coll Yellow Springs OH, 1969.
9. Connell DA, Schneider-Kolsky ME, Hoving JL, Malara F, Buchbinder R, Koulouris G, Burke F, and Bass C. Longitudinal study comparing sonographic and MRI assessments of actue and healing hamstring injuries. *AJR* 183: 975-984, 2004.
10. Cormack SJ, Newton RU, McGuigan MR, and Doyle TLA. Reliability of measures obtained during single and repeated countermovement jumps. *International Journal of Sports Physiology and Performance* 3: 131-144, 2008.
11. Edwards S, Austin AP, and Bird SP. The role of the trunk control in athletic performance of a reactive change-of-direction task. *Journal of Strength and Conditioning Research* 31: 126-139, 2016.
12. Havens KL and Sigward SM. Cutting mechanics: relation to performance and anterior cruciate ligament injury risk. *Med Sci Sports Exerc* 47: 818-824, 2015.
13. Hermens HJ, Freriks B, Disselhorst-Klug C, and Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology* 10: 361-374, 2000.
14. Hibbs AE, Thompson KG, French D, Wrigley A, and Spears I. Optimizing performance by improving core stability and core strength. *Sports Med* 38: 995-1008, 2008.
15. Higashihara A, Nagano Y, Ono T, and Fukubayashi T. Differences in activation properties of the hamstring muscles during overground sprinting. *Gait Posture* 42: 360-364, 2015.
16. Higashihara A, Nagano Y, Ono T, and Fukubayashi T. Relationship between the peak time of hamstring stretch and activation during sprinting. *Eur J Sport Sci* 16: 36-41, 2016.
17. Hopkins W. Linear models and effect magnitudes for research, clinical and practical applications. *Sportscience* 14, 2010.
18. Howard RM, Conway R, and Harrison AJ. Muscle activity in sprinting: a review. *Sports Biomech* 17: 1-17, 2018.
19. Koo TK and Li MY. A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research. *J Chiropr Med* 15: 155-163, 2016.
20. Koulouris G, Connell DA, Brukner P, and Schneider-Kolsky M. Magnetic resonance imaging parameters for assessing risk of recurrent hamstring injuries in elite athletes. *Am J Sports Med* 35: 1500-1506, 2007.

21. Lawrence MA, Chin A, and Swanson BT. Biomechanical comparison of the reverse hyperextension machine, and the hyperextension exercise. *Journal of Strength and Conditioning Research* 33: 2053-2056, 2019.
22. Lewis CL and Sahrmann SA. Musicle activation and movement patterns during prone hip extension exercise in women. *Journal of Athletic Training* 44: 238-248, 2009.
23. Macadam P and Feser EH. Examination of Gluteus Maximus Electromyographic Excitation Associated with Dynamic Hip Extension during Body Weight Exercise: A Systematic Review. *International Journal of Sports Physical Therapy* 14: 14-31, 2019.
24. Marshall P and Murphy B. The validity and reliability of surface EMG to assess the neuromuscular response of the abdominal muscles to rapid limb movement. *Journal of Electromyography and Kinesiology* 13: 477-489, 2003.
25. Pandy MG and Zajac FE. Optimal muscular coordination strategies for jumping. *Journal of Biomechanics* 24: 1-10, 1991.
26. Pollard CD, Sigward SM, and Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech (Bristol, Avon)* 25: 142-146, 2010.
27. Prieske O, Muehlbauer T, and Granacher U. The Role of Trunk Muscle Strength for Physical Fitness and Athletic Performance in Trained Individuals: A Systematic Review and Meta-Analysis. *Sports Med* 46: 401-419, 2016.
28. Raez MB, Hussain MS, and Mohd-Yasin F. Techniques of EMG signal analysis: detection, processing, classification and applications. *Biol Proced Online* 8: 11-35, 2006.
29. Schache AG, Blanch PD, Dorn TW, Brown NA, Rosemond D, and Pandy MG. Effect of running speed on lower limb joint kinetics. *Med Sci Sports Exerc* 43: 1260-1271, 2011.
30. Schuermans J, Van Tiggelen D, Palmans T, Danneels L, and Witvrouw E. Deviating running kinematics and hamstring injury susceptibility in male soccer players: Cause or consequence? *Gait Posture* 57: 270-277, 2017.
31. Shimokochi Y, Ide D, Kokubu M, and Nakaoji T. Relationship among performance of lateral cutting maneuver from lateral sliding and hip extension and abduction motions, ground reaction force, and body center of mass height. *Journal of Strength and Conditioning Research* 27: 1851-1860, 2013.
32. Verrall GM, Slavotinek JP, Barnes PG, and Fon GT. Diagnostic and prognostic value of clinical findings in 83 athletes with posterior thigh injury. *The American Journal of Sports Medicine* 31: 969-973, 2003.
33. Zebis MK, Skotte J, Andersen CH, Mortensen P, Petersen HH, Viskaer TC, Jensen TL, Bencke J, and Andersen LL. Kettlebell swing targets semitendinosus and supine leg curl targets biceps femoris: an EMG study with rehabilitation implications. *Br J Sports Med* 47: 1192-1198, 2013.
34. Zweifel M. Importance of horizontally loaded movements to sports performance. *Strength and Conditioning Journal* 39: 21-26, 2017.

Table 1. Comparison of peak and mean normalized electromyography between the hyperextension and reverse-hyperextension.

	Phase	Muscle	Exercise	Mean (± SD) (% MVIC)	ICC (95% CI)	CV%	% Difference	*p*	Hedges *g*	Power
Peak	Extension	Erector Spinae	RHE	107.3 ± 37.9	0.969 (0.928-0.987)	3.14	46.8	0.012	1.01	0.90
			HE	73.1 ± 25.4	0.989 (0.973-0.996)	5.42				
		Gluteus Maximus	RHE	49.3 ± 37.5	0.961 (0.910-0.984)	8.89	63.2	0.012	1.06	0.93
			HE	18.1 ± 13.1	0.958 (0.891-0.983)	10.64				
		Biceps Femoris	RHE	58.8 ± 21.7	0.966 (0.906-0.986)	6.33	34.1	0.024	0.95	0.87
			HE	38.7 ± 18.6	0.969 (0.924-0.987)	7.40				
	Flexion	Erector Spinae	RHE	101.6 ± 37.1	0.925 (0.830-0.968)	4.90	41.7	< 0.001	1.22	0.97
			HE	59.2 ± 29.1	0.989 0.973-0.996)	4.56				
		Gluteus Maximus	RHE	52.6 ± 33.6	0.972 (0.933-0.988)	5.89	70.7	< 0.001	1.44	0.99
			HE	15.4 ± 9.8	0.963 (0.911-0.985)	9.62				
		Biceps Femoris	RHE	58.9 ± 21.3	0.974 (0.937-0.989)	4.63	36.7	0.024	1.04	0.92
			HE	37.3 ± 18.4	0.966 (0.917-0.986)	7.78				
Mean	Extension	Erector Spinae	RHE	71.0 ± 30.5	0.975 (0.937-0.990)	5.15	38.1	< 0.001	1.03	0.91
			HE	43.9 ± 18.3	0.941 (0.840-0.977)	6.06				
		Gluteus Maximus	RHE	23.4 ± 15.8	0.973 (0.933-0.989)	9.06	65.0	< 0.001	1.25	0.98
			HE	8.2 ± 4.5	0.943 (0.863-0.977)	8.71				
		Biceps Femoris	RHE	39.7 ± 13.4	0.935 (0.844-0.974)	6.93	28.2	0.036	0.75	0.71
			HE	28.5 ± 15.1	0.942 (0.861-0.977)	10.35				
	Flexion	Erector Spinae	RHE	51.8 ± 16.1	0.955 (0.890-0.982)	3.87	41.1	< 0.001	1.47	1.00
			HE	30.5 ± 11.3	0.966 (0.916-0.986)	5.05				
		Gluteus Maximus	RHE	18.6 ± 9.2	0.973 (0.933-0.989)	5.68	51.1	< 0.001	1.15	0.96
			HE	9.1 ± 6.5	0.996 (0.989-0.998)	4.42				
		Biceps Femoris	RHE	28.3 ± 23.3	0.975 (0.937-0.990)	6.73	20.8	9.960	0.30	0.22
			HE	22.4 ± 13.3	0.983 (0.957-0.993)	6.67				

RHE = Reverse Hyperextension, HE = Hyperextension, SD = Standard Deviation, MVIC = Maximum Voluntary Isometric Contraction, ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, CV = Coefficient of Variation

Figure 1. An illustration of the performance of the A) hyperextension and B) reverse-hyperextension

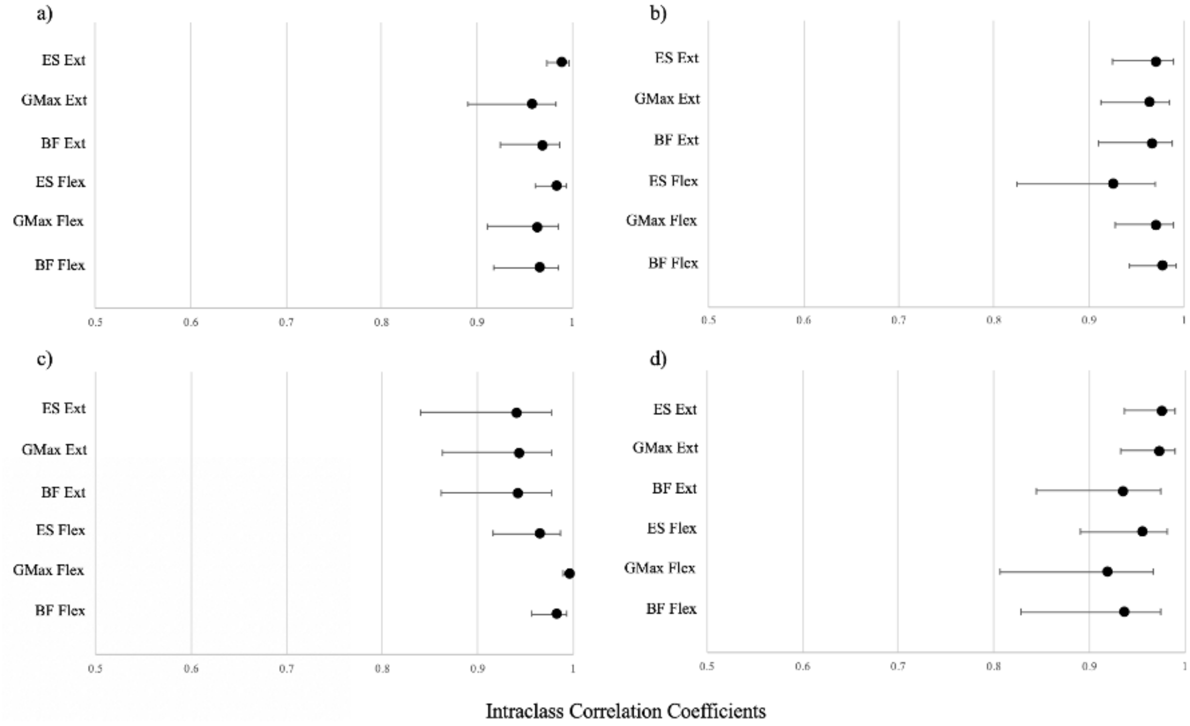

Figure 2. Reliability (intraclass correlation coefficients and 95% confidence intervals) for peak EMG (a) and mean (c) EMG amplitude during the hyperextension and peak (b) and mean (d) EMG amplitude during the reverse-hyperextension (ES = erector spinae); GMax = gluteus maximus; BF = biceps femoris; Ext = extension phase; Flex = flexion phase)

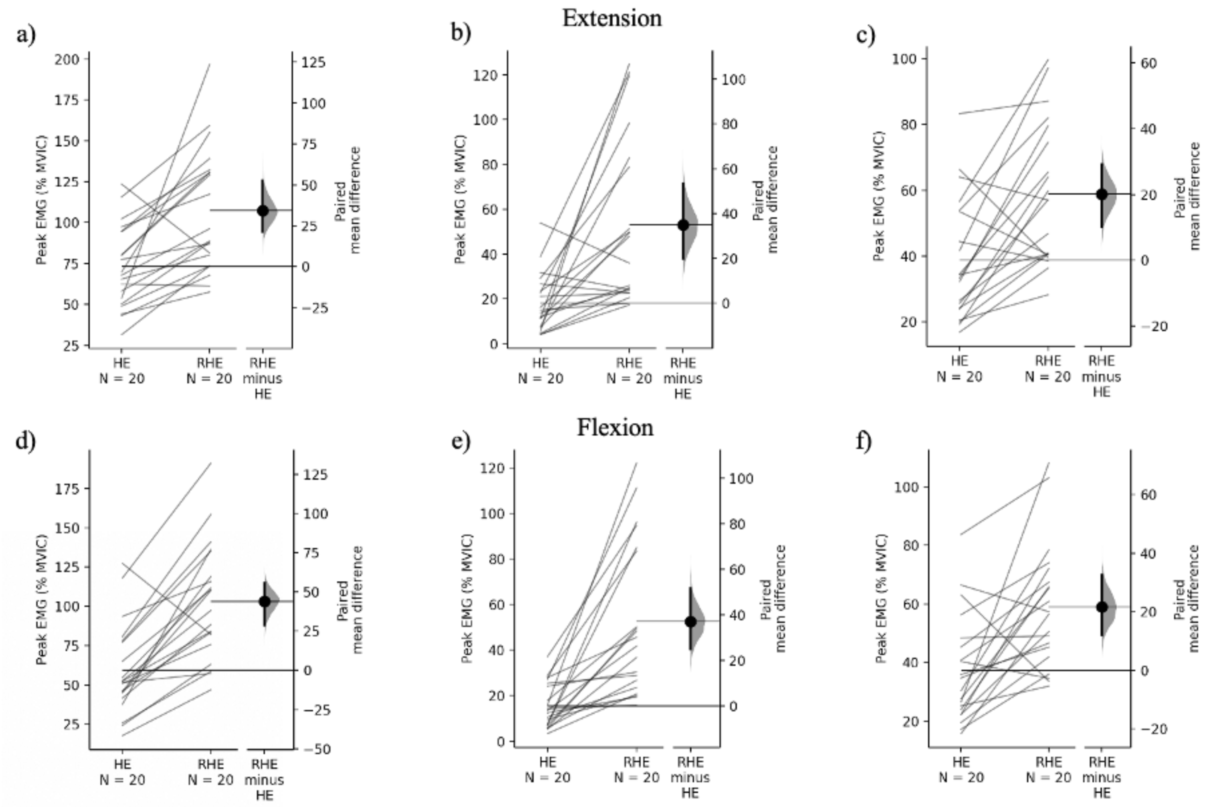

Figure 3. Comparison of normalized peak EMG of the a) erector spinae, b) gluteus maximus c) biceps femoris between exercises during the extension phases and d) erector spinae, e) gluteus maximus f) biceps femoris between exercises during the flexion phases. Individual data is plotted on the primary axis. Paired mean differences are plotted as a bootstrap sampling distribution on the secondary axis. Mean differences are depicted as dots; 95% confidence intervals are indicated by the ends of the vertical error bars.

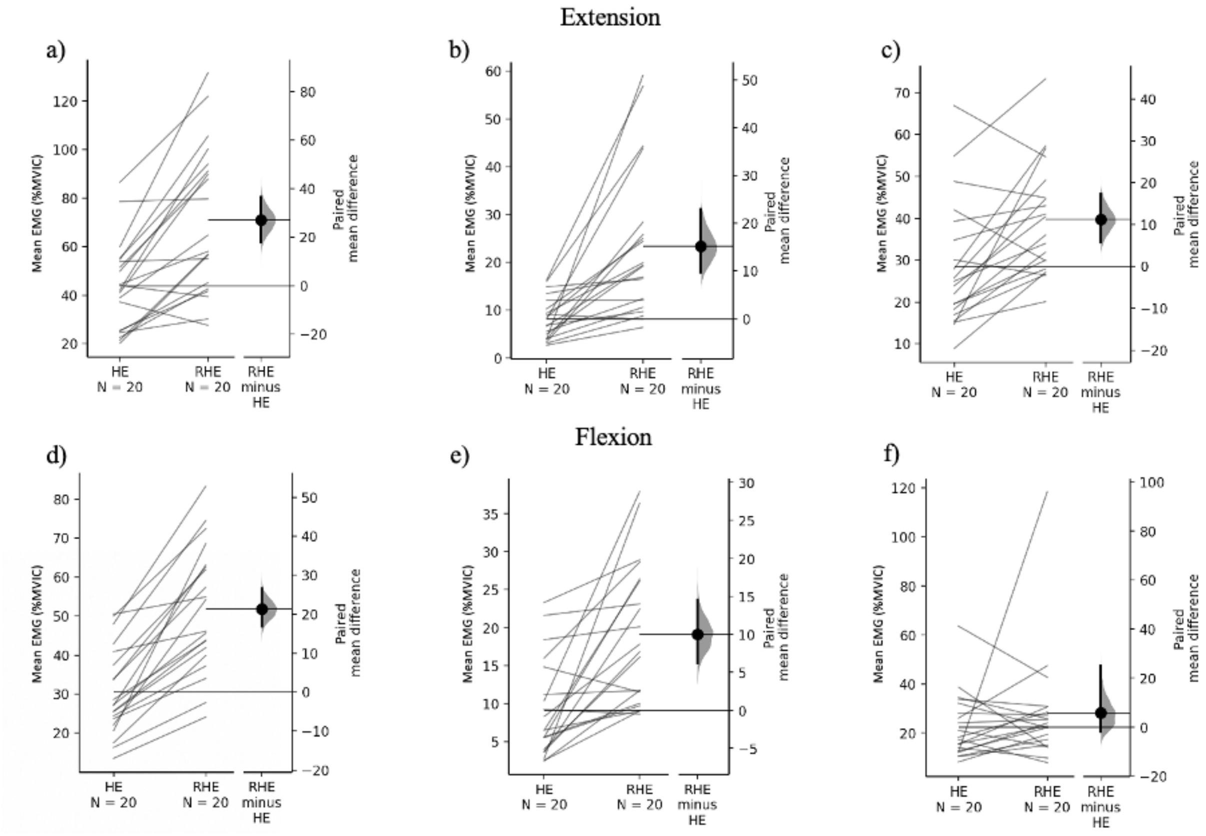

Figure 4. Comparison of normalized mean EMG of the a) erector spinae, b) gluteus maximus c) biceps femoris between exercises during the extension phases and d) erector spinae, e) gluteus maximus f) biceps femoris between exercises during the flexion phases. Individual data is plotted on the primary axis. Paired mean differences are plotted as a bootstrap sampling distribution on the secondary axis. Mean differences are depicted as dots; 95% confidence intervals are indicated by the ends of the vertical error bars.