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Does the application of lumbar mobilizations prior to the Nordic hamstring exercise influence hamstring measures of knee flexor strength, failure point and muscle activity? A replicated randomized cross-over trial

Objective: The aims of this study were to quantify the effects of unilateral posterior-anterior mobilization on force production, failure point and muscle activity of the hamstrings during the Nordic hamstring exercise (NHE) and explore individual differences in responses**.**

Methods: In a replicated randomized crossover trial, twenty-four participants (age [mean ± SD]: 27 ± 6 y, body mass: 82 ± 17 kg, stature: 181 ± 8 cm) completed two standardized intervention (L4/5 zygapophyseal mobilizations) and two control conditions. The failure point of the Nordic hamstring exercise was determined with 3D motion capture. Peak force, knee flexor torque and electromyography (EMG) of the Biceps Femoris were measured. Data analyses were undertaken to quantify mean intervention response and explore any individual response heterogeneity.

Results: Mean (95% confidence interval) left limb force was higher in intervention vs control by 18.7 (4.6–32) N. Similarly, right limb force was higher by 22.0 (3.4–40.6) N, left peak torque by 0.14 (0.06–0.22) Nm and right peak torque by 0.14 (0.05–0.23) Nm/Kg. Downward Force (DWF) angle was decreased in intervention vs control by 4.1° (0.5–7.6) on the side of application. Both peak EMG

activity (p=.002), and EMG at the DWF (Right) (p= .020) increased in the intervention condition by 16.8 (7.1–26.4) and 8.8 (1.5–16.1) (mV), respectively. Mean downward acceleration angle changed by only 0.3° (-8.9–9.4) in intervention vs control. A clear response heterogeneity was indicated only for force right (participant x intervention interaction: P=.044; Response heterogeneity SD = 34.5 (5.7–48.4) N). Individual response heterogeneity was small for all other outcomes.

Conclusions: Following UPA mobilization, immediate changes in bilateral hamstring force production and peak torque occurred during the NHE. The effect on the NHE failure point was unclear. EMG activity increased on the ipsilateral side. Response heterogeneity was generally similar to the random trial-to-trial variability inherent in the measurement of the outcomes.

Clinical Trials number: NCT03745482 (https://clinicaltrials.gov/ct2/show/NCT03745482)

Introduction

Hamstring strain injuries (HSI) are common across several sports affecting athletes of all ages, genders, and levels of competition.[1-5] Considerable time can be lost from sport related activity, resulting in diminished performance and financial loss.[6] Despite significant emphasis on injury preventive measures, HSI prevalence continues to rise and recurrence rates remain high.[7-8] Over 80% of HSIs involve the Biceps Femoris Long Head (BF_{LH}),[9-11] with the majority occurring in the terminal swing phase of high-speed running,[12] when a forceful eccentric contraction of the hamstrings is required.[13]

Lower eccentric hamstring strength is considered one of the main risk factors for future HSI highlighting the importance of eccentric strength for HSI avoidance.[14-17] The Nordic Hamstring Exercise (NHE) has been shown to be an effective way of increasing eccentric hamstring strength and developing higher maximal knee flexor torques whilst reducing HSI incidence by up to 51%.[18] HSI incidence rates have reduced significantly in athletes who adopted a NHE program within their regular training with a particularly preventive effect in reducing recurrent injuries.[17,19,20] The NHE activates all hamstring muscles, primarily semitendinosis and Biceps Femoris Short Head (BF_{SH}),[21] but also can increase fascicle length in the $BF_{LH.}$[22]. Blazevich et al,[23] suggested that the training range of motion is the dominant stimulus for fascicle length adaptation. Athletes with shorter BF_{LH} fascicles have demonstrated a fourfold greater risk of HSI than those with longer fascicles.[24] HSI risk was reduced by 75% for every 0.5 cm increase in fascicle length,[24] indicating the importance of training eccentric hamstring strength in a lengthened state for HSI avoidance.[25] Numerous authors have concluded that a lengthened based exercise rehabilitation programme, which can mimic important movements including sprinting and kicking, could be a key strategy of HSI management.[25,26] Therefore, extensibility of the hamstring is key to ensure loading can take place at a maximal lengthened state.

Due to its anatomical and functional relationship, the lumbar spine is widely seen as an important area to assess and manage as part of a global hamstring management strategy.[27-29] Recently, an individualised, multifactorial, criteria-based progressive algorithm was proposed for optimum hamstring injury treatment.[30] Within this, lumbar zygapophysial joint (z-joint) mobilizations are suggested in both the regeneration, and functional phase. Increases in hamstring extensibility following unilateral posterior-anterior (UPA) lumbar z-joint mobilizations has been reported in both the general population,[31] and elite athletes.[32] Both increased Biceps Femoris range of motion and reduced electromyography (EMG) activity, at the termination of active knee extension, following lumbar z-joint mobilizations has been demonstrated.[33] This EMG reduction is likely due to increased muscle spindle activity which stimulate golgi-tendon organs to produce a muscle reflex inhibition.[34-36] These changes in hamstring extensibility last between 15 and 20 minutes,[37] suggesting UPA lumbar z-joint mobilizations provide a limited time frame of hamstring adaptations. Nevertheless, due to these reported kinematic and kinetic adaptations, the use of UPA lumbar mobilizations pre NHE could increase the ability for the athlete to extend the hamstring into a desired lengthened state. Therefore, this may be a valuable addition to HSI prevention, and rehabilitation strategies optimizing eccentric strength gains and the muscle's torque-angle curve.

Six- weeks of eccentric strength training using NHE has been shown to optimise the control of the forward fall component of NHE (kinematic) with a concomitant increase in neuromuscular control (increased EMG activity during NHE).[38] This increase in EMG activity is likely due to the high level maximal eccentric activity compared to low level movement/activity and static conditions associated with previous EMG hamstring reductions.[33,37] Therefore, it is unclear if similar changes in extensibility would be apparent with previously reported EMG increases. Additionally, the study did not have access to a dynamometer, therefore it is unclear if force and torque also increased alongside muscle length changes. To date, no studies have addressed whether UPA lumbar mobilizations prior to NHE will improve kinematic and neuromuscular performance during the lowering phase of the NHE.

In the context of precision or personalized medicine, it has been deemed important to quantify any inter-individual variability in response to an intervention alongside the quantification of the mean intervention response.[39-45] Such intervention response heterogeneity cannot be quantified robustly using a typical crossover study design.[43] An approach that has recently been proposed to quantify individual differences in the intervention response involves quantifying the participant-by-response interaction from replicated intervention and control conditions.[39,44,45] Such an approach has rarely been adopted in musculoskeletal research.

Therefore, currently a lack of understanding exists regarding the effect of lumbar mobilizations performed prior to the NHE, specifically regarding the failure point, hamstring EMG activity and force production. The aims of this study were to quantify the effects of UPA mobilizations on force production, failure point and muscle activity of the hamstrings during the NHE and quantify individual differences in responses**.** Knowledge of the intervention's effects, initially in a healthy population, will provide data for evaluation of its value, prior to use with HSI pathology. We hypothesize the application of UPA z-joint mobilizations will result in an increase of peak force and peak torque, EMG activity and failure point of the NHE.

METHODS

Study Design:

Because the proposed intervention was hypothesised to elicit only very short-term changes which would 'wash-out' relatively rapidly, a controlled replicated randomized cross-over design was utilized.[37,42] This reporting will follow recommendations from CONSORT for publishing cross-over trials.[46] Participants were randomized to different trial sequences comprising two intervention (I) trials and two control (C) trials. Each visit was separated by an interval of seven days. Randomization was

conducted by one investigator (GA) using sealed_envelope.com allocating each participant to one of six primary allocation sequences. The six sequences were:

C-I-C-I

C-I-I-C

C-C-I-I

I-C-I-C

I-C-C-I

I-I-C-C

Ethical approval was received from **removed for review** Ethics committee and the research was conducted in accordance with the Declaration of Helsinki. The trial was registered with clinicaltrials.gov prior to study recruitment (NCT03745482). No changes to the methods were implemented following trial commencement.

Participants:

All participants were recruited, via means of a study flyer, from a population of staff and students at Teesside University, United Kingdom, between November 2018 and May 2019. For eligibility all participants were aged 18 and above and were free from musculoskeletal injury of the spine and lower limb. All participants were recreationally active playing a team sport at least once per week (performing moderate intensity activity 3-6 metabolic equivalents, METs).[47] Participants were excluded if they indicated current low back, hamstring or knee pathology; previous spinal or lower limb surgery; or any contraindications to spinal mobilizations.[48] Participants were instructed to refrain from caffeine at least four hours prior to testing and avoid strenuous exercise at least 24 hours prior.[47] A total of 29 participants were recruited to the study but four failed to meet the inclusion criteria and one participant withdrew for personal reasons. Therefore, a total of 24 male participants completed the study (age [mean ± SD]: 27 ± 6 y, body mass: 82 ± 17 kg, stature: 181 ± 8 cm). Outcome measures

were obtained from all participants who completed the intervention and control conditions twice. All participants were asked at each trial to confirm they continued to meet the studies criteria.

Outcome Measures:

The Hamstring Solo (NJ Doherty Solutions, Ireland), and Hamstring Solo Elite app (Version 4.2, ND Sports Performance) is a pressure feedback system which allows the calculation of eccentric force (Newtons) and estimation of peak torque (Newton metres) of the NHE in real time through load cell technology. Participants position themselves on the incline board of the device with ankles fixed below an ankle bar. Participants lowered their torso toward the ground trying to resist the force as slowly as possible by extending at the knee joint until failure. Participants were given visual and coaching cues during familiarization to ensure minimal hip flexion during the trial. Each NHE performance was visually monitored by the trial investigators. Excessive hip movement or the participant not controlling the descent from the start of the movement resulted in the repetition being rejected.[49] We performed pilot testing on 8 participants (age [mean ± SD]: 28 ± 6 y, body mass: 96 ± 22 kg, stature: 183 ± 10 cm) over four testing sessions separated by 72 hours to ascertain the reliability of the Hamstring Solo. Standardized changes in the mean were trivial (trials 2 - 1: -0.06, 95% confidence interval (CI), -0.26 to 0.19; trials 3 - 2: -0.20, -0.52 to 0.32, trials 4 - 3: -0.04, -0.25 to 0.19) between testing sessions and the force typical error was 10% (8.7% to 13%) with a interclass-correlation coefficient ($ICC_{3,1}$) of 0.91 (95% CI: 0.81 to 0.96). The reliability of the solo elite agrees with previous studies of isokinetic dynamometry and the Nordbord.[49,50]

Figure 1 – Representative example of the angular displacement of EMG activity of the downward phase of a NHE

INSERT FIGURE 1 ABOUT HERE

Kinematic data acquisition

The failure point of the NHE, is defined as when the participant can no longer produce sufficient eccentric force to control the descent and finishes the exercise.[51] This is characterized by a loss of tension, and sudden increase in knee angular velocity through loss of trunk control.[52] However, there is no universally accepted measure of finding the failure point. We determined the kinematic changes during NHE via 3D motion capture. Data was collected during the performance of all the NHE trials across both conditions. We used the Vicon plugin gait (PiG) lower body model marker-set to establish the kinematic changes at the knee joint. Retroflected markers (14 mm) with double-side tape were placed bilaterally on the ASIS, PSIS, mid-thigh, lateral knee epicondyle, mid-tibia, lateral malleolus, calcaneus, and 2nd toe (dorsal aspect on the 2nd metatarsal heads proximal to the MP joint). Six wall-mounted Vicon MX13 infrared cameras (Vicon, Vicon Motion Systems Ltd) collected 3D motion capture data at a sampling frequency of 100 Hz. 3D motion capture data was processed via Vicon Nexus (version 1.8.5) using inbuilt pipeline functions to calculate 3D kinematic data.

Kinematic data analysis for NHE

Methods used to establish the failure point range from visual assessment,[53] using an arbitrary cut- off point from an angular acceleration curve of 10 deg·s^{-1} and using algorithms to establish changes in angular displacement.[52] We followed a previously published method to determine the failure point during the downward phase of the NHE.[38] All kinematic data were initially filtered off-line within Vicon Nexus using a low- pass filter (Fourth-order bi-directional Butterworth filter with a cut-off frequency of 6 Hz), and exported as a .CSV file. Subsequently, each .CSV file was imported into a custom-designed programme in MATLAB (MathWorks, Version 2019a). Briefly, the angular displacement of the left and right knee joint was differentiated to angular velocity using the first derivative method.

We calculated the following outcomes, bilaterally, from the angular velocity curve; 1) The angle (°) at downward acceleration (DWA) was obtained by applying a slope function (using the coefficient from

182 the *polyfit* function) to produce an acceleration curve. However, to smooth the data the slope function
183 was applied over a 200 ms window with a 100 ms overlap. The difference in slopes between one-time
184 window and the next was calculated. The angle at the corresponding time point of the highest slope
185 difference was reported as the point of maximum downward acceleration, and thus loss of eccentric
186 control. 2) Additionally, we identified the first point at which an initial downward inflection occurred
187 in the acceleration curve produced from method 1, which we refer to as the angle at downward fall
188 (DWF). 3) The angle at peak velocity was taken as the angle corresponding to the time point at the
189 maximum velocity from the angular velocity curve. A representative displacement-time curve with the
190 three variables can be seen in Figure 1. The PiG lower body model calculates the knee angle via the
191 sagittal shank axis projected into the plane perpendicular to the knee flexion axis. Knee flexion is the
192 angle in that plane between this projection and the sagittal thigh axis. The sign is such that a positive
193 angle corresponds to a flexed knee. Thus, as the athlete lowers themselves to the floor the angle
194 decreases from ~90°. An angle closer to zero at the failure point would represent greater hamstring
195 extension prior to failure.
196
197 *Electromyography (EMG) data acquisition and reduction for NHE*
198 Surface electromyography (EMG) was attached to the Biceps Femoris bilaterally during the NHE Prior
199 to application, the skin was shaved and cleaned with a 70% isopropyl alcohol wipe. Noraxon, self-
200 adhesive Ag/AgCl snap electrodes (Noraxon USA) were applied to the muscle belly on the line halfway
201 between the ischial tuberosity and the lateral epicondyle of the tibia as per SENIAM guidelines.[54] Once
202 placed, electrodes remained in position throughout the testing procedure to eliminate placement
203 error. A wireless EMG system (Cometa Wave, Zerowire wireless EMG, Cometa Srl) synced directly
204 (utilising analog capture functionality of a Vicon connectivity device) with Vicon Nexus was sampled
205 at 1000 Hz. Vicon Nexus acted as the driver for the EMG system to start data capture to synchronise
206 the EMG and Kinematic data. Data imported into MATLAB (MathWorks, Version 2019a) for further
207 data reduction and filtering. Raw EMG data was filtered off-line using a high pass Butterworth filter,

with a cut off frequency of 20 Hz,[54,55] full wave rectified, followed by a low pass bi-directional Butterworth filter with a 20 Hz cut-off frequency to create a linear envelope. EMG data was then time normalized to the kinematic data using spline interpolation (Figure 1). We calculated the following variables for the EMG; 1) peak EMG amplitude (mV), and 2) EMG amplitude at downward fall (mV). The peak EMG amplitude was normalized and expressed as a percentage of the peak amplitude of the EMG value from each of the five repetitions. No changes to outcome measures were implemented following trial commencement.

Intervention:

UPA lumbar mobilizations were applied with the participant in prone position. Mobilizations were applied by a physiotherapist with 15 years clinical experience and postgraduate qualifications in spinal mobilization. Mobilizations were applied to the dominant side decided by kicking foot (right n = 24).[32,33,37,56] Spinal level was determined by passive physiological intervertebral movement and spinal palpation by the same physiotherapist. Grade 3 UPA lumbar mobilizations, defined as large amplitude oscillations into resistance, were applied to the L4/5 unilateral z-joint for 2 min, three times to reflect common clinical application and previous studies.[32,33,48,56] Mobilizations were applied at a frequency of 2 Hz maintained by a metronome to provide sympathetic nervous system excitability.[57] To ensure consistent force application within and between participants, a bipedal force measurement system (F-Scan® 7.0, Tekscan Inc) was specifically cut and placed under the pisiform of the physiotherapist. Standardized changes in mean force application between replicates were trivial (-0.10, -0.99 to 0.78) N, the typical error was 2.5% (2.0% to 3.6%) with $ICC_{3,1}$ of 0.33 (95% CI: -0.08 to 0.64) similar to previous published literature.[33,58]

Procedure:

Participants attended the biomedical sciences laboratory on five separate occasions. One familiarization session, two intervention and two control trials. The familiarisation session of the NHE

234 took place at least one week prior to the first testing session. All testing sessions were performed at
235 the same time of day to reduce the influence of diurnal effects Participant height (cm), mass (kg) and
236 age (y) were recorded.
237
238 All participants watched a video of a subject completing the NHE and received verbal instructions.
239 Participants were instructed to start in a kneeling position, with the upper body vertical and straight.
240 The participant was then instructed to slowly lower the upper body towards the ground ensuring no
241 hip flexion, maximising loading in the eccentric phase, before breaking the fall with their hands.[19] The
242 video was shown at the beginning of both the familiarisation session, and all respective control and
243 intervention sessions.
244
245 Participants then conducted a standardized warm-up on an ergometer (Wattbike, Nottingham UK)
246 undertaken for 5 minutes at 60% max resting heart rate. Following this either the intervention or
247 control was administered. For the control trials, participants lay prone on a plinth for 10 minutes, the
248 approximate time the intervention took to be applied. After the intervention or control, participants
249 then performed five repetitions of the NHE, as per the initial weeks training protocol in both Mjolsnes
250 et al.[59] and Van der Horst et al.[19] studies. Each repetition was separated by a one-minute rest period.
251 A cool down was offered to all participants on the cycle ergometer for 10 minutes at a self-desired
252 pace.
253
254 **Statistical Analysis:**
255 A replicated cross-over (two intervention and two control conditions) increases statistical power for
256 detection of mean treatment effects over a conventional 2-level crossover study and, crucially,
257 enables the exploration of the participant x treatment interaction term required for robust
258 judgements regarding individual differences in treatment response.[42] The analysis approach was

designed to quantify both mean treatment effects and explore the presence of any inter-individual differences in treatment effect and comprised three components as described by Goltz et al.[45]

Our sample size of 24 participants was dictated by the obligations of the rather time-consuming four-trial protocol, rather than statistical power. Nevertheless, based on our sample size, and knowledge about the reliability of our primary outcome, we can estimate statistical power and/or minimal detectable target effect size. In terms of the detection of a mean target treatment effect, and using GPower 3.1, we estimated that a difference between intervention and control conditions (standardised to the between-subjects SD) of 0.27 would be detected as statistically significant ($P<0.05$) with 80% statistical power, assuming a correlation coefficient between trials of 0.9 (obtained from our prior pilot testing/reliability work). We also highlight the fact that the replicated nature of our study design (both conditions undertaken twice) would be likely to further increase statistical power.

It is difficult to estimate statistical power in the context of treatment response heterogeneity because the within-subjects variability that is of interest in this context is unknown before the replicated crossover study is completed.[60] In addition, "post hoc" statistical power estimations (based on the observed effect size rather than a target effect size) are not appropriate.[61] One approach to quantifying the degree of "true" inter-individual variability in response is to calculate the correlation coefficient between the two replicates of intervention/control (see below).[42] It can be estimated that a sample size of 24 would enable a "moderate" target correlation of 0.4 to be detected as statistically significant. The confidence interval of a target correlation coefficient of 0.4 would be 0.00 to 0.69.

The associations between the first and second replicates of the control-adjusted treatment effect were quantified using Pearson's product-moment correlation coefficients.[42] The first intervention session in any participant's sequence was paired to the first control condition in the same individual's

sequence. Differences in response that are stable within participants would manifest themselves as a high correlation between first and second pairs of replicates. An overall "naïve" estimate of the true (control condition–adjusted) between-subject differences in treatment response were calculated as follows ($SSSSSSSS = \sqrt{SSSSSS^2 - SSSSSS^2}$),.[40] The standard deviation of individual responses (SDIR) represents the true inter-individual variation in treatment effect. Standard deviations of the pre-post change were calculated for the intervention conditions (SDi) and control conditions (SDc). Each of these two SDs was calculated using the relevant equation for pooling SDs because there were 2 sets of data to pool in each condition.[62] A positive SDIR indicates greater treatment response heterogeneity relative to the random trial-to-trial variability. Finally, a within-participant linear mixed model quantified any participant-by condition interaction for each outcome measure.[63] Condition and their interaction effects were modelled as fixed effects, and participant and participant-by-condition terms were modelled as random effects. Standard residual diagnostics were undertaken according to methods reported in Goltz et al.[45]

Mean differences between intervention and control were expressed as raw and standardised mean differences with their uncertainty expressed as 95% CIs with exact *P* values. In the absence of a precise clinical anchor for an important difference in our NHE related outcomes (in their units of measurement), we compared the standardized ESs to conventional thresholds.[64] These thresholds are context-dependent and we recognize that there have been recent calls for some standardized differences to be as high as 0.5 to be considered clinically relevant.[64] An ES of 0.2 denoted the minimum important mean difference for all outcomes, with an ES of 0.5 being moderate and an ES of 0.8 being large.[65] To calculate the minimal clinically important difference (MCID) for individual responses, the threshold of 0.2 for interpreting standardized mean changes was used.[65,66] We recognise that such an interpretation is more of a "fall-back" approach when robust thresholds for clinical/practical importance have yet to be formulated using hard outcomes of morbidity and mortality, or via agreement amongst clinicians.[64]

RESULTS

All 24 participants were randomly assigned, received the intended conditions and were analysed for the outcomes. No unintended adverse effects were reported from any participants and there was no loss to follow-up. The mean and standard deviation for each measurement and the raw mean effects of the intervention versus the control condition are presented in Table 1 and the standardised effects are visualised with their confidence intervals in Figure 2. Small increases were observed in the intervention (vs control) in mean peak force for left (18, 95% confidence interval 4.6 to 33 N, p=.011) and right sides (22, 3.4 to 41 N, p=.020) and mean peak torque left (0.14, 0.06 to 0.22 kg, p=.002) and right (0.14, 0.05 to 0.23 kg, p=.005). A small decrease in the angle at DWF on the participants' dominant right side where the mobilisations were performed, was observed (-4.1, -7.6 to -0.5 degrees, p=.027). Further moderate increases in peak EMG activity were also observed on the right limb (17, 7.1 to 26 mV, p=.002) and EMG at the angle of DWF (8.8, 1.5 to 16 degrees, p=.021) with mobilisations. Increases in peak EMG on the left limb were also moderate but the estimate was less precise (0.71, -1.1 to 30 mV, p=.067). Similarly, small decreases were observed in angle at DWA on the left limb (-3.6, -7.3 to 0.1 degrees, p=.055) but the uncertainty in these estimates were large.

Table 1. Means and SDs of the pre-to-post change scores for the mobilization and control (no intervention) conditions

INSERT TABLE 1 ABOUT HERE

Figure 2. Standardised changes in the mean with uncertainty expressed as 95% confidence intervals

INSERT FIGURE 2 ABOUT HERE

The results of the three approaches for quantifying inter-individual differences in intervention response are presented in Table 2. Generally, there was good agreement between the approaches,

whereby a large correlation between crossover replicates was associated with relatively large values for the SDir. Intervention response heterogeneity was most apparent for force right – there was a statistically significant participant by intervention interaction (p=.04) and the SDir was substantially larger than the mean treatment effect for this variable (Table 1). No other statistically significant participant by intervention interaction terms were detected, and SDir were generally smaller than the respective mean intervention effect for each of the other variables. The rather small and not statistically significant correlations between crossover replicates are also presented in the scatterplots of Figure 2. It can be seen that individual differences in response were highly variable between the pairs of intervention and control trials. This indicates an absence of any endogenous intervention heterogeneity over and above the random trial-to-trial within-subjects variability that is present.

Table 2 – True inter-individual differences between the mobilizations and control (no intervention) conditions

INSERT TABLE 2 ABOUT HERE

Figure 3 – Inter-individual differences between mobilizations and control (non-intervention) for all replicated measures

INSERT FIGURE 3 ABOUT HERE

DISCUSSION

The primary findings of this study in healthy recreationally active males were; (1) the application of UPA mobilizations resulted in an increase between conditions for hamstring peak force (bilaterally), peak torque (bilaterally), and a decreased angle at DWF on the right (side of UPA application), (2) an increase of peak EMG activity was observed in the right hamstring as was EMG activity at DWF, (3) no differences were detected between conditions for the angle at DWA (4) inter-individual responses were found for force production of the right hamstring with negligible response heterogeneity for all

other outcomes. No previous researcher has attempted to assess the effect of UPA lumbar z-joint mobilizations on the peak force, peak torque and failure point of the hamstring during an NHE. As such, our study provides novel data to suggest that UPA lumbar z-joint mobilizations increases force production and peak torque bilaterally to the hamstring complex and might improve participant's angle at failure on the applied side during downward phase of NHE.

This is the first study in this field to explore the participant by treatment interaction (for quantification of individual response heterogeneity), alongside mean condition differences. A strength of our study is the replicated cross over design and the statistical approaches employed, which have been advocated to explore inter-individual variability in responses to an intervention.[40,42]

HSIs continue to be problematic, despite significant emphasis on preventive measures. HSI prevalence rates have reduced significantly in athletes who adopted a NHE program within their regular training with a particularly preventive effect in reducing recurrent injuries. [17,19,20,67] The value of treating the hamstring region proximally via the lumbar spine has previously been advocated,[29,30,68] with lumbar spine mobilizations shown to increase hamstring extensibility and potentially reduce Biceps Femoris EMG activity during AKE and lumbar flexion.[32,33] The aim of this study was to investigate how UPA lumbar z-joint mobilizations effect the peak force, EMG activity and failure point of the hamstring during an NHE.

Our study is the first to provide evidence to clinicians that UPA mobilizations can acutely influence force production during a functional eccentric strength exercise. Both increasing hamstring force production, and overall strength over time has been suggested to decrease the incidence of HSI's.[24] The increases in force and torque production bilaterally may be related to increased spinal motor-neuron excitability, increased neural motor-drive and thus increased rate of force development.[69] The application of UPA mobilizations pre NHE may facilitate these central processes to produce the desired

increases in force output throughout the eccentric exercise. This has implications for prevention and management of HSI through increasing eccentric strength which is known to reduce injury risk.[24] The individual variability investigated in all the studied outcomes did not indicate any large response heterogeneity, except for peak force of the right leg (table 2). Nevertheless, we cannot rule out clinically relevant response heterogeneity in all the other study outcomes because our study was not specifically powered to quantify response heterogeneity. Our primary hypotheses were relevant to the mean treatment effect, while we explored the secondary objective of individual heterogeneity in treatment effect.

No accepted measure of finding the failure point of the NHE exists. The angle of DWA was obtained as per Delahunt et al.[38] In addition, we identified the first point of initial downward inflection as the angle of DWF (see methods section). Interestingly, we found that the application of UPA mobilizations did not influence the failure point of the NHE as measured by the angle at DWA, with increased confidence intervals which crossed zero, but this was increased for the angle of DWF of the applied side. Previous research has reported the ability of lumbar mobilizations and specifically UPA's to increase the extensibility in the short- term with effects lasting for approximately 15 to 20 minutes.[34] Potentially, the increase in extensibility, and decreased passive stiffness may have been beneficial to a NHE when the hamstring is stressed through an eccentric lengthened state. However, we cannot conclude with certainty if UPA enable the hamstring's failure point to be increased during the NHE. Further work is required to validate the calculation for measuring the failure point of the exercise. Increased noise within the data was observed for the angle of DWA bilaterally, and therefore the reliability of measures of DWA and DWF are required. The variation in the data might be due to a combination in estimation of angular displacement from 3D motion capture, and noise compounding the data when we differentiated from angular displacement to velocity and acceleration. Therefore, we do not provide definitive evidence for the ability of UPA to increase the failure point.

We observed moderate increases in EMG at DWF on the right side where the mobilisations were administered, but not on the left. Additionally, peak EMG was clearly increased in the right limb but not the left. Side specific changes following L4/5 mobilization have been reported by Perry and Green,[57] with a greater response on the side of application. Whilst Perry and Green,[57] didn't use EMG as an outcome measure, our EMG at DWF data provide some support for their conclusions that neurophysiological and anatomical inter-relationships in the lumbar spine do exist and can be influenced through manual mobilizations. However, we would caution over interpretation of these data particularly considering the width of the confidence intervals for EMG data (Figure 2). Indeed, a similar moderate improvement in peak EMG was observed for the left limb but the wider CI denotes less certainty in the statistical estimation of "true" effect size.

Increased muscle activity can be related to amplified force production.[69-72] Interestingly, Hegyi et al,[73] reported the BF_{LH} produced the lowest level of muscle activity during an NHE and is associated with higher strain close to the proximal muscle-tendon junction.[74] Opar et al.[75] reported that recreational athletes with a previous history of HSI have both decrease biceps femoris muscle activation and eccentric hamstring strength during maximal voluntary contractions. Similar findings have been reported in athletes with a history of HSI during the late swing phase of high velocity running gait.[76] Previously, it has been reported that the activity of the hamstring complex remains elevated during the terminal segment of the NHE.[77] We report that the application of UPA mobilizations increased this peak muscle activity in the immediate term. Longer-term studies including Delahunt and colleagues following a six-week Nordic hamstring program reported a significant increase in EMG activity of both semi-tendinous and biceps femoris during the eccentric exercise.[38] This increase is likely due to the neural adaptations of exercise programs.[78] To achieve such electromyographic changes in Delahunt et al's.[38] study a total of 340 repetitions of the NHE were performed and produced similar results to studies assessing activity changes in the quadricep muscle group.[79,80] These longer-term changes are proposed to result from preferential recruitment of type II muscle fibers.[81] The significantly higher

EMG activity of the hamstrings in the later segments of the NHE may be explained by a greater recruitment of available motor units to generate sufficient torque to control the fall of the torso which is compensating for the reduced mechanical advantage a lengthened position.[77] These adaptations would not have occurred in the small dose each participant in our study was exposed to. Whilst we did not attempt to evaluate how these changes occurred in our study it is likely that increased EMG activity is related to increased central motor output to the hamstrings to maintain the fixed task requirements.[82] From the observed data in our study the application of UPA mobilizations may provide an important strength stimulus by increasing muscle activity.

Limitations and future research

It is important to acknowledge some limitations with our study when interpreting the results. Our study used a healthy population and thus the effect of UPA lumbar z-joint mobilizations on peak force, torque, EMG activity and failure point in athletes with HSI is currently unknown and requires investigation. Currently a minimally important clinical difference for force production or muscle activity during the NHE is unknown. Therefore, we cannot be certain that the increases reported within our study would be clinically meaningful. The Hamstring solo directly calculates force and estimates peak torque. Readers should be aware that this estimation is based on several assumptions including segment mass and caution should be applied when interpreting results. Finally, the effect of skin movement artefact on joint motion when using a marker set up has been established and could have led to measurement error of the failure point of the NHE.[83,84]

Conclusion

Our results help to inform practitioners of the variations observed from the administration of UPA lumbar mobilizations to the hamstring complex during the NHE. Following UPA application to the L4/5 facet joint immediate changes in bilateral hamstring force production and peak torque occurred

during the NHE. The failure point as measured by angle at DWF was decreased for the mobilization side but this was not replicated when measured via DWA. Further work is required to ascertain gold standard calculation of the failure point. Peak EMG muscle activity of the hamstring complex was observed together with increased activity during the DWA on the ipsilateral side of mobilization application. Only force production of the dominant leg resulted in inter-individual differences and larger samples are required to investigate this further.

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Table 1. Means and SDs of the pre-to-post change scores for the mobilization and control (no intervention) conditions

Variable	Mean ± standard deviation measurement				Raw mean difference (95%CI) (Pooled over replicates) Intervention minus control	P-value
	Intervention 1	Intervention 2	Control 1	Control 2		
Force Left (N)	297 ± 102	303 ± 85	271 ± 90	292 ± 94	18.7 (4.6 to 32.8)	.011
Force Right (N)	320 ± 81	336 ± 70	287 ± 66	321 ± 66	22.0 (3.4 to 40.6)	.020
Peak Torque Left (Nm/Kg)	1.52 ± 0.45	1.50 ± 0.35	1.32 ± 0.38	1.42 ± 0.36	0.14 (0.06 to 0.22)	.002
Peak Torque Right (Nm/Kg)	1.64 ± 0.39	1.66 ± 0.28	1.45 ± 0.32	1.58 ± 0.26	0.14 (0.05 to 0.23)	.005
Angle DWF Left (°)	66.4 ± 13.1	62.4 ± 12.9	72.3 ± 12.2	66.6 ± 13.6	-2.5 (-10.7 to 5.7)	.537
Angle DWF Right (°)	67.9 ± 13.9	67.8 ± 15	72 ± 15	71.8 ± 12.1	-4.1 (-7.6 to -0.5)	.027
Angle DWA Left (°)	51.1 ± 9.4	48.2 ± 9.7	55.6 ± 12.4	50.1 ± 11.1	-3.6 (-7.3 to 0.1)	.055
Angle DWA Right (°)	50 ± 9.5	52.3 ± 8	53.3 ± 11.7	53.2 ± 9.5	0.3 (-8.9 to 9.4)	.950
EMG at DWF Left (mV)	45.77 ± 16.75	39.42 ± 17.70	39.21 ± 15.25	39.87 ± 16.18	-0.41 (-8.6 to 7.8)	.916
EMG at DWF right (mV)	51.80 ± 20.26	41.37 ± 16.14	37.11 ± 12.41	36.75 ± 15.04	8.8 (1.5 to 16.1)	.021
Peak EMG Left (mV)	67.97 ± 20.46	63.25 ± 22.05	53.44 ± 16.56	59.58 ± 22.37	13.9 (-1.1 to 28.9)	.067
Peak EMG Right (mV)	66.85 ± 21.84	64.83 ± 23.59	51.64 ± 21.20	51.02 ± 19.55	16.8 (7.1 to 26.4)	.002

700

701 **Table 2 –** True inter-individual differences between the mobilizations and control (no intervention) conditions

Variable	Differences between conditions (replicate 1) Mean +SD	Differences between conditions (replicate 2) Mean +SD	Correlation between replicates (R, CI)	SDiR Estimate 1	SDiR, Estimate 2	P-value
Force Left (N)	25.88 (36.61)	11.54 (40.16)	0.38 (-0.02 to 0.68)	16.33	15.41 (-22.36 to 31.22)	.530
Force Right (N)	33.61 (48.01)	14.96 (37.76)	0.59 (0.24 to 0.80)	33.16	34.48 (5.65 to 48.44)	.044
Peak Torque Left (Nm/Kg)	0.20 (0.31)	0.08 (0.21)	0.08 (-0.33 to 0.47)	0.15	0.10 (-0.12 to 0.19)	.441
Peak Torque Right (Nm/Kg)	0.19 (0.33)	0.08 (0.23)	0.05 (-0.36 to 0.45)	0.16	0.35 (-0.14 to 0.21)	.448
Angle DWF Left (°)	-5.95 (15.40)	-4.20 (14.27)	0.13 (-0.29 to 0.50)	-1.69	6.88 (-8.96 to 13.23)	.730
Angle DWF Right (°)	-4.19 (8.63)	-3.94 (14.75)	0.10 (-0.48 to 0.32)	4.95	0.37 (-5.11 to 5.13)	.992
Angle DWA Left (°)	-4.49 (13.84)	-1.96 (11.31)	0.05 (-0.36 to 0.45)	-7.15	-2.8 (-6.09 to 4.62)	.599
Angle DWA Right (°)	-3.28 (8.34)	-0.93 (9.26)	0.34 (-0.08 to 0.65)	-5.94	10.37 (-6.24 to 15.95)	.150
EMG at DWF left (mV)	6.56 (21.52)	-0.45 (19.46)	0.22 (-0.02 to 0.57)	7.68	-4.89 (-12.13 to 9.95)	.702
EMG at DWF right (mV)	14.69 (19.23)	4.62 (19.47)	0.02 (-0.42 to 0.39)	13.04	-11.19 (-16.47 to 4.56)	.093
Peak EMG Left (mV)	14.53 (26.60)	3.67 (29.74)	0.42 (-0.09 to 0.76)	7.73	11.34 (-16.93 to 23.33)	.544
Peak EMG Right (mV)	15.21 (28.86)	13.82 (27.97)	0.05 (-0.45 to 0.54)	9.98	-16.48 (-24.24 to 6.64)	.092

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