DORSIFLEXION, PLANTAR-FLEXION, AND NEUTRAL ANKLE POSITIONS DURING PASSIVE RESISTANCE ASSESSMENTS OF THE POSTERIOR HIP AND THIGH MUSCLES

Journal of Athletic Training 2015;50(5):467–474
doi: 10.4085/1062-6050-49.6.04

www.natajournals.org

original research

Dorsiflexion, Plantar-Flexion, and Neutral Ankle Positions During Passive Resistance Assessments of the Posterior Hip and Thigh Muscles

Ty B. Palmer, MEd, CSCS*; Kazuma Akehi, PhD, ATC†; Ryan M. Thiele, MA, ATC‡; Douglas B. Smith, PhD‡; Aric J. Warren, EdD, ATC, CSCS‡; Brennan J. Thompson, PhD, CSCS§

*School of Health Sciences, Kent State University, OH; †Department of Kinesiology and Sports Science, University of Nebraska at Kearney; ‡Department of Health and Human Performance, Oklahoma State University, Stillwater; §Department of Health, Exercise, and Sport Sciences, Texas Tech University, Lubbock

Context: Passive straight-legged–raise (SLR) assessments have been performed with the ankle fixed in dorsiflexion (DF), plantar-flexion (PF), or neutral (NTRL) position. However, it is unclear whether ankle position contributes to differences in the passive resistance measured during an SLR assessment.

Objective: To examine the influence of ankle position during an SLR on the passive torque, range of motion (ROM), and hamstrings electromyographic (EMG) responses to passive stretch of the posterior hip and thigh muscles.

Design: Crossover study.

Setting: Research laboratory.

Patients or Other Participants: A total of 13 healthy volunteers (5 men: age = 24 ± 3 years, height = 178 ± 6 cm, mass = 85 ± 10 kg; 8 women: age = 21 ± 1 years, height = 163 ± 8 cm, mass = 60 ± 6 kg).

Intervention(s): Participants performed 6 randomly ordered passive SLR assessments involving 2 assessments at each condition, which included the ankle positioned in DF, PF, and NTRL. All SLRs were performed using an isokinetic dynamometer programmed in passive mode to move the limb toward the head at 5°/s.

Main Outcome Measure(s): During each SLR, maximal ROM was determined as the point of discomfort but not pain, as indicated by the participant. Passive torque and EMG amplitude were determined at 4 common joint angles (θ) separated by 5° during the final common 15° of ROM for each participant.

Results: Passive torque was greater for the DF condition than the NTRL ($P = .008$) and PF ($P = .03$) conditions at θ_3 and greater for the DF than NTRL condition ($P = .02$) at θ_4. Maximal ROM was lower for the DF condition than the NTRL ($P = .003$) and PF ($P < .001$) conditions. However, we found no differences among conditions for EMG amplitude ($P = .86$).

Conclusions: These findings suggest that performing SLRs with the ankle positioned in DF may elicit greater passive torque and lower ROM than SLRs with the ankle positioned in PF or NTRL. The greater passive torque and lower ROM induced by the DF condition possibly were due to increased tension in the neural structures of the proximal thigh.

Key Words: hamstrings muscles, passive torque, range of motion, stiffness, neural tension

Key Points

- Passive torque was greater and maximal range of motion was lower for the straight-legged raises with the ankle positioned in dorsiflexion than in plantar flexion and neutral.
- Hamstrings electromyographic amplitude values were not different among the dorsiflexion, plantar-flexion, and neutral conditions and remained relatively low across the range of motion.
- The greater passive torque and lower ROM induced by the dorsiflexion condition may have been due to increased passive tension in the neural structures of the proximal thigh.

Passive musculotendinous resistive properties are commonly assessed via the application of a dynamic stretch.[1–4] For the hamstrings specifically, the use of a straight-legged–raise (SLR) movement to assess passive resistive properties, such as passive torque and range-of-motion (ROM) measurements, may be important for determining athletic[5,6] and health[7] status and predicting lower body injuries.[8] Tafazzoli and Lamontagne[7] showed that passive torque values of the hamstrings measured during an SLR effectively discriminated between individuals with and without low back pain. Furthermore, Witvrouw et al[8] reported that the maximal ROM of the hamstrings achieved during an SLR could be used to identify athletes who were at risk for developing hamstrings muscle injuries.

Passive musculotendinous properties contributing to the resistance to stretch traditionally have been attributed to several structural factors, including the stretching of the stable cross-links between the actin and myosin filaments, direct resistance from the actin and myosin filaments, stretching of the noncontractile proteins of the endosarcomeric and exosarcomeric cytoskeletons, and deformation of the noncontractile connective tissues located within and surrounding the muscle belly.[2] Eliciting the stretch reflex may also be a factor that contributes to increases in torque measured during passive stretch.[9–11] Specifically, Lamontagne et al[10] indicated that stretch-reflex excitation depends on velocity; therefore, passive SLR assessments performed at higher stretch velocities may elicit greater stretch-reflex excitation, causing increased activation of the stretched muscles and possible contamination of the passive torque measurements with both active force production and passive tension. The need for a device to control for stretch velocity has been addressed by several authors[3,11–14] who have used isokinetic dynamometers when performing SLR testing. Whereas researchers have attempted to control for the potential confounding effects of stretch velocity, few investigators have examined the importance of controlling for ankle position during the SLR. For example, SLR assessments have been performed with the ankle fixed in dorsiflexion (DF),[15] plantar flexion (PF),[16–18] or neutral (NTRL).[3,4,7,11,12] However, it is unclear whether ankle position contributes to differences in the passive resistance to stretch. If ankle position influences the passive resistance to stretch as determined by the SLR, then ankle position may need to be standardized when performing SLR testing to reduce discrepancies (and possibly help explain contrasting findings) across studies and provide more consistency to enhance physiologic comparative analyses.

The incorporation of sensitizing maneuvers, such as ankle DF or cervical flexion, into a passive SLR assessment has been shown to increase tension on the neural structures of the proximal thigh without increasing hamstrings stretch.[15,19,20] Given these findings, authors[21,22] have suggested that greater neural tension created by sensitizing maneuvers may play a role in the passive resistance produced during hamstrings stretch. McHugh et al[23] recently provided evidence to support this hypothesis by demonstrating in vivo that adding cervical flexion to a seated passive knee-extension assessment increased resistance to hamstrings stretch. In addition, Hall et al[24] reported that passive resistance during manually applied passive SLR movements was greater with the ankle positioned in DF than NTRL. However, they did not examine the effect of PF on resistance to stretch, nor did they perform a statistical analysis. Moreover, manually applied passive movements do not yield consistent acceleration and deceleration from trial to trial[10]; therefore, potential differences in stretch velocity may have been a confounding factor in the previous study.[24] The use of an isokinetic dynamometer to perform SLR testing at a constant movement velocity provides reliable measurements of passive torque and ROM that are not confounded by velocity-related differences.[11] To our knowledge, no investigators have made direct statistical comparisons of the differences in ROM and passive torque measurements involving SLRs performed at a slow, constant movement velocity using an isokinetic dynamometer with the ankle positioned in DF, PF, and NTRL. Given the importance and prevalence of the SLR as a tool to assess passive musculotendinous resistive properties along with the relationships among these passive properties, health status, and sport-related injuries, further research is warranted to examine the potential effects of placing the ankle in several positions across ankle-joint ROM on passive torque and ROM measurements during stretching of the posterior muscles of the hip and thigh. Therefore, the purpose of our study was to examine the influence of ankle positions (DF, PF, and NTRL) on the passive torque, ROM, and hamstrings electromyographic (EMG) characteristics measured during passive isokinetic SLR assessments of the posterior hip and thigh muscles.

METHODS

Participants

Five healthy men (age = 24 ± 3 years, height = 178 ± 6 cm, mass = 85 ± 10 kg) and 8 healthy women (age = 21 ± 1 years, height = 163 ± 8 cm, mass = 60 ± 6 kg) volunteered for this investigation. No participant reported any acute or chronic neuromuscular diseases or musculoskeletal injuries to the ankle, knee, or hip joints. Of the 13 participants, 12 reported engaging in 1 to 10 h/wk of aerobic exercise, 10 reported 1 to 7 h/wk of resistance exercise, and 5 reported 1 to 4 h/wk of recreational sports. No participant was a competitive athlete; however, given their reported levels of exercise, these individuals might be best categorized as active, recreationally trained participants.[11] All participants provided written informed consent and completed a health history questionnaire, and the study was approved by the Institutional Review Board for Human Subjects Research at Oklahoma State University.

Experimental Design

We used a randomized, repeated-measures design to investigate the influence of ankle position during an instrumented SLR (iSLR) on the passive torque, ROM, and hamstrings EMG responses to passive stretch of the posterior hip and thigh muscles. Each participant visited the laboratory 2 times, with visits separated by 2 to 3 days. During the first visit, participants familiarized themselves with the testing procedures by performing several iSLR trials. During the second visit, participants completed 6 iSLR assessments involving 2 assessments at each condition (DF, PF, and NTRL ankle positions). They rested for 2 minutes after each iSLR assessment.[25] The order of the DF, PF, and NTRL assessments was randomized, and the mean of the 2 assessments for each ankle condition for passive torque, ROM, and electromyography was calculated at each joint angle and used for all subsequent analyses.

Assessment of Passive Torque and Range of Motion

Passive torque and ROM of the posterior hip and thigh muscles were examined with an iSLR technique, which consisted of using a calibrated Biodex System 3 isokinetic dynamometer (Biodex Medical Systems Inc, Shirley, NY) programmed in passive mode to move the limb toward the head at 5°/s (Figure 1). For each iSLR assessment, participants lay supine with the knee braced in full extension and the ankle immobilized in 10° of DF, 10° of PF, or NTRL (0°) with an adjustable, custom-made cast that was fixed around the foot and held with straps placed above the ankle and over the toes and metatarsals. During the iSLRs, the input axis of the dynamometer was aligned slightly superior and anterior to the greater trochanter of the femur to account for movement of the greater trochanter

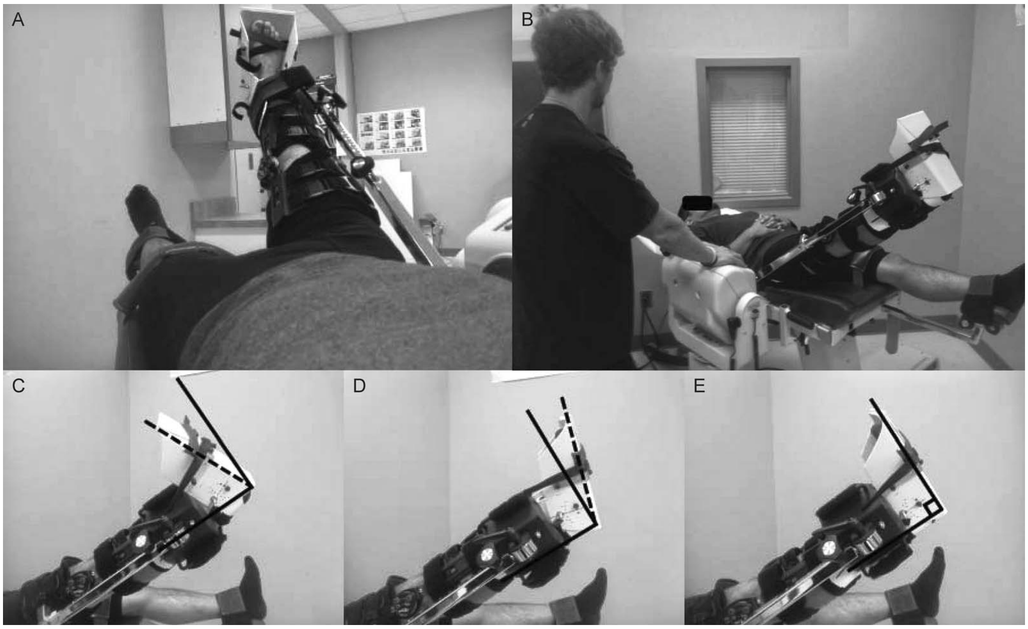

Figure 1. The instrumented straight-legged–raise assessment technique as seen from the, A, left and, B, right sides of the participant. The isokinetic dynamometer was programmed in passive mode to move the limb toward the head at 5°/s. For each instrumented straight-legged raise, the ankle was immobilized in, C, 10° of dorsiflexion, D, 10° of plantar flexion, or E, a neutral 0° position. The lines in C, D, and E represent the angles at which the ankle was positioned for each condition.

during the iSLR, and restraining straps were placed over the left (unstretched) thigh and ankle. All iSLR assessments were performed on the right limb to the point of discomfort but not pain as indicated by the participant; this measurement was regarded as the maximal ROM, and the limb was returned immediately to the baseline position.

Surface Electromyography

Surface electromyography was recorded for the biceps femoris from bipolar preamplified electrodes (model TSD150B; BIOPAC Systems, Inc, Santa Barbara, CA) with a fixed center-to-center interelectrode distance of 20 mm and a gain of 350. To decrease the interelectrode impedance, the skin was cleansed with isopropyl alcohol before electrode placement. The electrodes were taped directly to the skin and were placed at 50% of the distance between the ischial tuberosity and the lateral epicondyle of the tibia. The placements were based on the recommendations of Hermens et al.[26] A single, pregelled, disposable Ag-AgCl electrode (model Quick Prep; Quinton Instruments Co, Bothell, WA) was placed on the palmar side of the right wrist to serve as a reference electrode.

We calculated EMG amplitude with a root mean square function for 200-millisecond epochs corresponding to each whole-number degree during the ROM. According to the procedures of Herda et al,[27] we subtracted EMG amplitude baseline noise values from the EMG amplitude values recorded during the passive iSLR assessments. Furthermore, the corrected EMG amplitude values (μV root mean square) were normalized to the corresponding prestretch maximal voluntary isometric contraction (MVIC) peak electromyography and expressed as a percentage of the MVIC peak EMG amplitude.

For the MVIC assessments, participants performed two 5-second MVICs of the posterior hip and thigh muscles while lying supine with restraining straps placed over the waist, left thigh, and ankle. During the MVICs, the thigh and leg were set in the same position as the starting point of the passive iSLR assessments, which was a hip-joint angle of 20° above the horizontal plane. We instructed participants to extend the thigh as hard as possible for 5 seconds. Isometric peak torque (PT) for each MVIC was determined as the highest mean 500-millisecond epoch during the torque plateau, and the highest PT trial was selected for subsequent EMG normalization. The EMG amplitude was quantified during the same 500-millisecond epoch used to calculate PT and was considered 100% (maximal) voluntary activation. The iSLR assessments could not be considered passive if the corrected and normalized EMG amplitude was greater than 5% of MVIC, in accordance with Gajdosik et al[28] and Herda et al.[27]

Signal Processing

During each iSLR assessment, torque (Nm), joint-angle position (°), and EMG (μV) signals were sampled simultaneously at 1 kHz (model MP100WSW; BIOPAC Systems, Inc), stored on a personal computer (model Inspiron 8200; Dell Inc, Round Rock, TX), and processed offline using custom-written LabVIEW software (version 11.0; National Instruments, Austin, TX). Torque and position signals were low-pass filtered with a zero-phase lag, fourth-order Butterworth filter that had a cutoff of 10 Hz. The EMG signal was scaled and bandpass filtered with a zero-phase lag, fourth-order Butterworth filter from 20 to 400 Hz. All subsequent analyses were conducted on the scaled and filtered signals.

For passive torque, gravity correction was performed during each iSLR using a cosine function in which the limb mass was subtracted from the torque signal across the ROM. The scaled and gravity-corrected torque and joint-angle signals were plotted as passive angle-torque curves and fitted with a fourth-order polynomial regression model

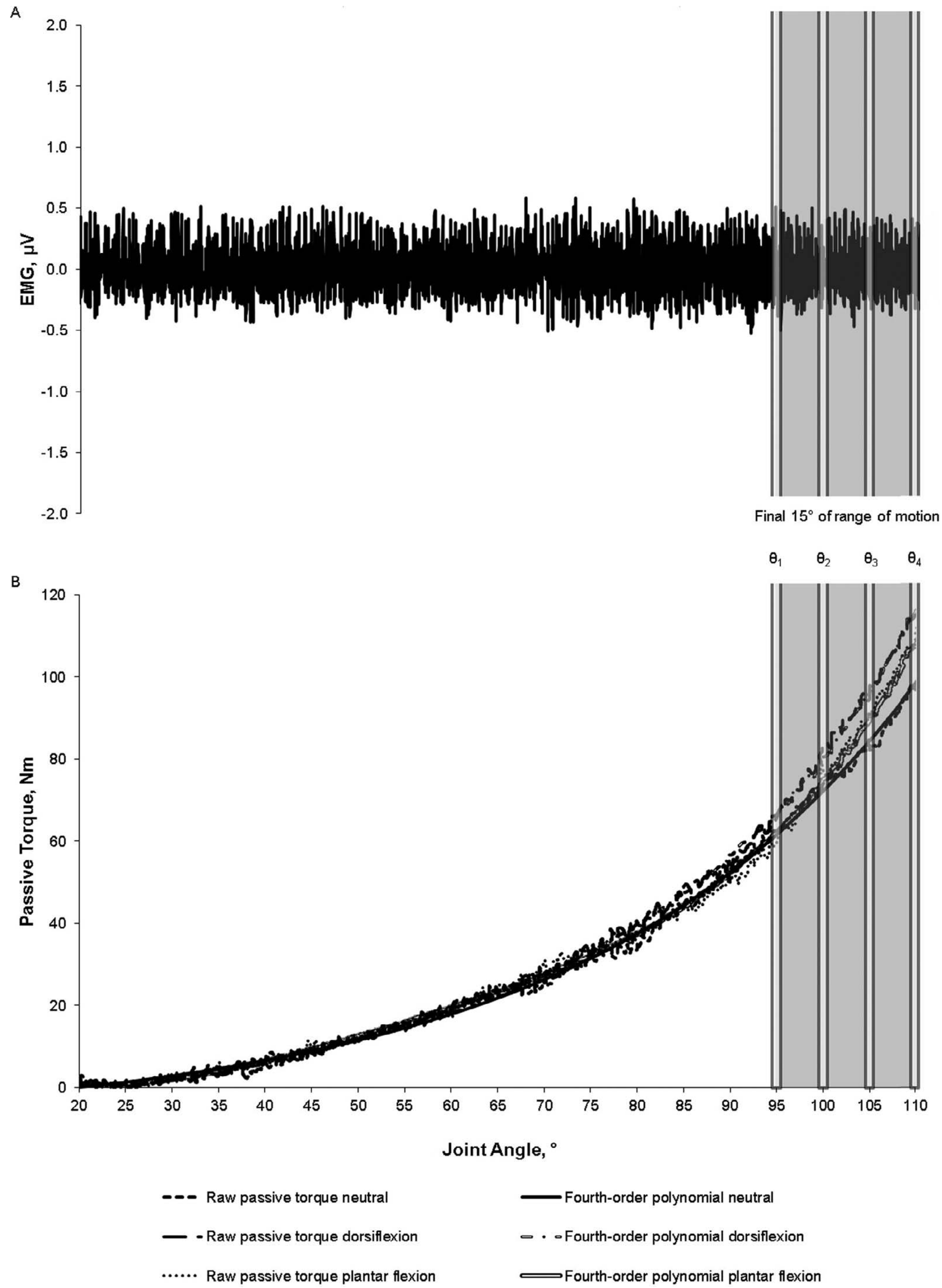

Figure 2. A, The electromyographic (EMG) signal from the biceps femoris and, B, passive angle-torque curve assessed during a single, passive, instrumented straight-legged raise of the posterior muscles of the hip and thigh. For visualization, the fourth-order polynomial regression and the raw angle-torque curves recorded for this participant with the ankle in neutral, dorsiflexion, and plantar-flexion positions also were plotted. The large rectangular shaded area represents the final 15° of range of motion that was common to all ankle conditions (dorsiflexion, plantar flexion, and neutral). The vertical white boxes represent every fifth degree in the final 15° of range of motion (θ_1, θ_2, θ_3, and θ_4), where passive torque and EMG amplitude values were calculated and used for analysis.

based on the equation reported by Nordez et al.[29] Passive torque and electromyography were quantified at 4 common joint angles (θ) separated by 5° during the final common 15° of ROM (at 0°, 5°, 10°, and 15°) for each participant. Consequently, the same absolute joint angles could be used for each participant to calculate passive torque and electromyography for each iSLR assessment (Figure 2).

Statistical Analyses

A separate 1-way repeated-measures analysis of variance (ANOVA) was used to compare means for maximal ROM among the DF, PF, and NTRL conditions. In addition, 2 separate 2-way repeated-measures ANOVAs (condition [DF, PF, NTRL] × joint angle [θ_1, θ_2, θ_3, θ_4]) were used to analyze the passive torque and EMG amplitude data. When

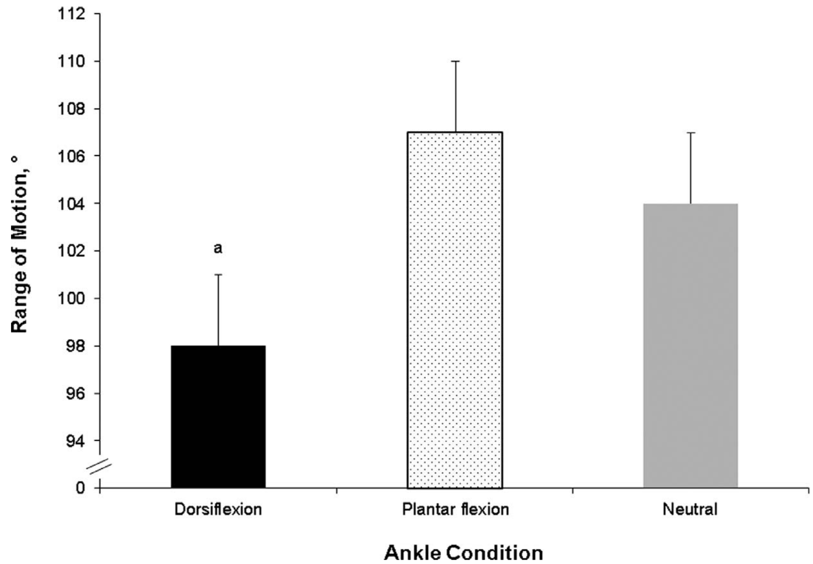

Figure 3. Maximal passive hip-flexion range-of-motion values of the posterior muscles of the hip and thigh for the dorsiflexion, plantar-flexion, and neutral conditions. Values are means ± standard errors. [a] Indicates differences for range of motion among ankle conditions (dorsiflexion < neutral; dorsiflexion < plantar flexion) ($P < .05$).

appropriate, we conducted follow-up analyses that consisted of 1-way repeated-measures ANOVAs and Bonferroni-corrected pairwise comparisons. Statistical analyses were performed using SPSS software (version 20.0; IBM Corp, Armonk, NY). The α level was set at .05.

RESULTS

Maximal ROM was lower for the DF condition than the NTRL ($P = .003$) and PF ($P < .001$) conditions (Figure 3). For passive torque, we found a condition × joint-angle interaction ($F_{6,72} = 4.149$, $P = .04$; Figure 4). Passive torque

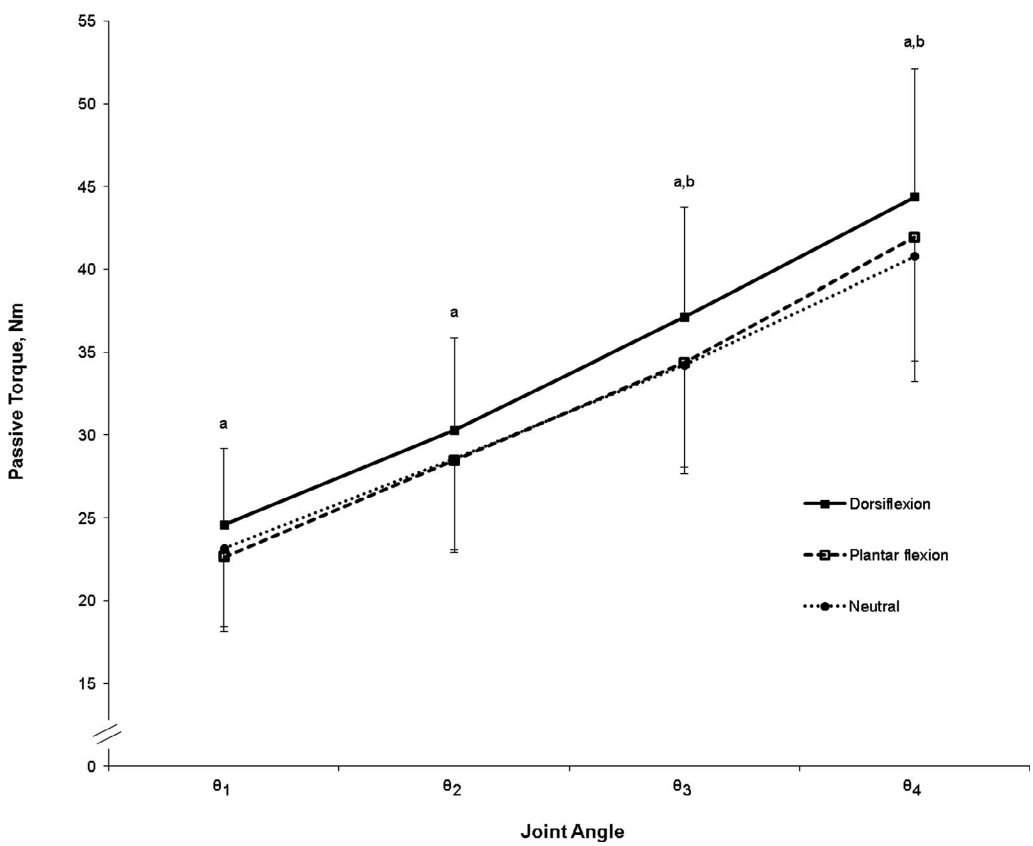

Figure 4. Passive torque values at each joint angle of the posterior muscles of the hip and thigh for the dorsiflexion, plantar-flexion, and neutral conditions. Values are means ± standard errors. [a] Indicates differences among joint angles ($\theta_1 < \theta_2 < \theta_3 < \theta_4$) for the dorsiflexion, neutral, and plantar-flexion conditions ($P < .05$). [b] Indicates differences among ankle conditions at θ_3 (dorsiflexion > neutral; dorsiflexion > plantar flexion) and θ_4 (dorsiflexion > neutral) ($P < .05$).

was greater for the DF condition than the NTRL ($P = .008$) and PF ($P = .03$) conditions at θ_3 and was greater for the DF than the NTRL condition at θ_4 ($P = .02$). However, no differences in passive torque were observed between the DF and PF conditions at θ_4 ($P = .07$) or among the DF, PF, and NTRL conditions at θ_1 (P range, .27–.99) and θ_2 (P range, .12–.99). Passive torque also increased with joint angle ($\theta_1 < \theta_2 < \theta_3 < \theta_4$) for the DF ($F_{3,36} = 37.404$, $P < .001$), PF ($F_{3,36} = 35.665$, $P < .001$), and NTRL ($F_{3,36} = 33.208$, $P < .001$) conditions. For EMG amplitude, we found no condition × joint-angle interaction ($F_{6,72} = 0.903$, $P = .44$), no main effect for condition ($F_{2,24} = 0.157$, $P = .86$), and no main effect for joint angle ($F_{3,36} = 0.096$, $P = .87$). The EMG amplitude values were 0.70%, 0.64%, and 0.74% of MVIC for the DF, PF, and NTRL conditions, respectively.

DISCUSSION

Passive torque was greater at θ_3 and θ_4 and maximal ROM was lower for the DF condition than the NTRL and PF conditions despite a lack of differences among conditions for EMG amplitude (Figures 3 and 4). Researchers have reported similar findings regarding the influence of ankle position on ROM[21,25,30,31] and passive torque[24] during SLR testing. For example, Hall et al[24] noted that passive torque during SLR testing was greater with the ankle positioned in DF than in NTRL. However, in contrast to our findings, they found that the increases in resistance to stretch were accompanied by increases in EMG activity of the hamstrings. The authors hypothesized that because EMG activity was higher for the DF than the NTRL condition, the greater passive torque induced by ankle DF was due to an elicitation of the stretch reflex.[24] Alternatively, the EMG-amplitude values in our study were not different among the DF, PF, and NTRL conditions and remained relatively low (ie, <1% of MVIC) across the ROM, suggesting that an elicitation of the stretch reflex likely did not contribute to the condition-related differences observed in resistance to stretch. The discrepancies in EMG activity between our findings and those reported by Hall et al[24] may be due to differences in the techniques used to assess passive resistance. We assessed passive resistance using an isokinetic iSLR technique, whereas Hall et al[24] used a manual technique, which consisted of the primary investigator applying resistance against a load cell positioned immediately posterior to the heel while the limb was moved toward the head. Consequently, differences in stretch velocities may contribute to these discrepancies,[11] as the SLRs in our study were performed at a slow, constant velocity (ie, 5°/s), which may not have been the case in the previous study[24] as stretch velocities were not reported. Investigators have provided support for velocity discrepancies during manual application, indicating that manually applied passive movements are often performed at high stretch velocities (ie, >5°/s),[9–11] and given that higher stretch velocities may evoke greater stretch-reflex excitation,[11] manual techniques possibly have a greater potential for eliciting the stretch reflex and, consequently, higher EMG activity than iSLR techniques.

The lack of an observable contractile response to stretch for all conditions, which was supported by the EMG inactivity, suggests the absence of a detectible stretch reflex. Therefore, other factors may be responsible for the differences observed in passive torque and ROM among the DF, PF, and NTRL conditions. In support of this hypothesis, Gajdosik et al[21] demonstrated smaller hamstrings ROM values during SLR assessments with the ankle positioned in DF than in PF; however, because EMG amplitude remained low for both the DF and PF conditions, they suggested that the lower ROM in the DF condition was not due to an elicitation of the stretch reflex but rather to an increase in passive tension on the sciatic nerve. Researchers[19,23–25,32,33] have indicated that the sciatic nerve and other peripheral nerves are parts of a large, continuous neural tissue tract in which increases in tension and movement in 1 part created by maneuvers, such as ankle DF or cervical flexion, may create tension and movement of the neural structures in other parts. Using animal models, Boyd et al[19] showed that adding ankle DF to SLR testing increased tension and movement in the sciatic nerve of the proximal thigh. Furthermore, McHugh et al[23] recently demonstrated with in vivo human musculotendinous models that adding cervical flexion to a seated passive knee-extension assessment caused increased passive resistance to hamstrings stretch without any changes in EMG activity. They hypothesized that, in the absence of EMG activity, the greater resistance to stretch for the cervical-flexion condition was due to an increase in passive tension in the neural tissues.[23] Thus, the possibility of increases in neural tension may explain why we observed greater passive torque and lower ROM values for the DF condition. Moreover, our finding of greater passive torque for the DF condition within the final common 5° of ROM (ie, at θ_3 and θ_4) further supports the notion that the increased passive resistance to stretch with the ankle in DF may be a consequence of tension within the neural tissues, which would be subjected to increased tension at greater joint angles as previous authors[34–36] have suggested. Alternatively, however, variations among fascicle connections of the lower body musculature may also help to explain the passive torque and ROM differences that we observed. For example, investigators[21,25,31,37] have suggested that fascicle connections at the popliteal region between the hamstrings and gastrocnemius muscles may allow DF to increase the amount of passive tension on the hamstrings and, thereby, influence the resistance to stretch measured during passive SLR assessments. However, given the scope of our study and the limited data available regarding these findings, it was not feasible to ascertain the underlying mechanisms resulting in the higher passive torque and lower ROM values for the DF condition than the PF and NTRL conditions. Thus, future research involving more invasive measures (ie, ultrasound imaging) is necessary to elucidate more specifically the mechanisms that may be responsible for influencing the passive torque and ROM differences displayed among iSLRs with various ankle positions.

CONCLUSIONS

Our findings align with those of other authors indicating greater passive torque[24] and lower ROM[21,25,30,31] values during SLRs with the ankle positioned in DF than in PF and NTRL. Some authors[24,25] have hypothesized that the increase in passive resistance to stretch for the DF condition may be due to an elicitation of the stretch reflex; however, given that the EMG amplitude values in our study remained low for the DF, PF, and NTRL conditions, we conclude that other factors may be contributing to the condition-related differences observed in passive torque and ROM. One factor may be an increase in the amount of neural tension on the sciatic nerve due to the DF condition[21]; however, we are aware of no studies in which the authors have shown this to definitively contribute to the greater resistance to stretch measured during an SLR assessment. Nevertheless, our findings further support the influence of ankle position on resistance to stretch and highlight the potential importance of ankle position during SLR testing while elucidating the possibility that tension in the neural tissues may contribute to the passive torque measured during posterior hip- and thigh-muscle stretching. Researchers[3,11–14] have indicated that the use of an isokinetic dynamometer during an SLR assessment provides reliable and quantitative measurements of passive torque and ROM of the posterior muscles of the hip and thigh by controlling for the velocity of stretch. However, considering the potential influence of ankle DF on these variables, our findings highlight the importance of standardizing ankle position by fixing the ankle in either PF or NTRL when conducting these types of passive assessments to avoid the confounding effects on resistance to stretch observed in the DF ankle position. Furthermore, athletic trainers, physical therapists, and other practitioners may use these findings and perhaps exercise caution when interpreting passive-stiffness data from SLRs because these types of tests may be influenced by ankle position, which could adversely affect the capacity of the SLR as a diagnostic tool to identify and assess individuals with low back pain and other sport-related injuries.

REFERENCES

1. Ryan ED, Thompson BJ, Herda TJ, et al. The relationship between passive stiffness and evoked twitch properties: the influence of muscle CSA normalization. *Physiol Meas*. 2011;32(6):677–686.
2. Gajdosik RL. Passive extensibility of skeletal muscle: review of the literature with clinical implications. *Clin Biomech (Bristol, Avon)*. 2001;16(2):87–101.
3. Marshall PWM, Mannion J, Murphy BA. Extensibility of the hamstrings is best explained by mechanical components of muscle contraction, not behavioral measures in individuals with chronic low back pain. *PMR*. 2009;1(8):709–718.
4. Palmer TB, Jenkins NDM, Thompson BJ, Smith DB, Cramer JT. The relationship between passive stiffness and muscle power output: influence of muscle CSA normalization. *Muscle Nerve*. 2014;49(1): 69–75.
5. Ekstrand J, Gillquist J. The frequency of muscle tightness and injuries in soccer players. *Am J Sports Med*. 1982;10(2):75–78.
6. Kujala UM, Oksanen A, Taimela S, Salminen JJ. Training does not increase maximal lumbar extension in healthy adolescents. *Clin Biomech (Bristol, Avon)*. 1997;12(3):181–184.
7. Tafazzoli F, Lamontagne M. Mechanical behaviour of hamstring muscles in low-back pain patients and control subjects. *Clin Biomech (Bristol, Avon)*. 1996;11(1):16–24.
8. Witvrouw E, Danneels L, Asselman P, D'Have T, Cambier D. Muscle flexibility as a risk factor for developing muscle injuries in male professional soccer players: a prospective study. *Am J Sports Med*. 2003;31(1):41–46.
9. Boiteau M, Malouin F, Richards CL. Use of a hand-held dynamometer and a Kin-Com dynamometer for evaluating spastic hypertonia in children: a reliability study. *Phys Ther*. 1995;75(9): 796–802.
10. Lamontagne A, Malouin F, Richards CL, Dumas F. Evaluation of reflex- and nonreflex-induced muscle resistance to stretch in adults with spinal cord injury using hand-held and isokinetic dynamometry. *Phys Ther*. 1998;78(9):964–975.
11. Palmer TB, Jenkins ND, Cramer JT. Reliability of manual versus automated techniques for assessing passive stiffness of the posterior muscles of the hip and thigh. *J Sports Sci*. 2013;31(8):867–877.
12. Marshall PW, Cashman A, Cheema BS. A randomized controlled trial for the effect of passive stretching on measures of hamstring extensibility, passive stiffness, strength, and stretch tolerance. *J Sci Med Sport*. 2011;14(6):535–540.
13. Göeken LN, Hof AL. Instrumental straight-leg raising: results in healthy subjects. *Arch Phys Med Rehabil*. 1993;74(2):194–203.
14. Göeken LN, Hof AL. Instrumental straight-leg raising: results in patients. *Arch Phys Med Rehabil*. 1994;75(4):470–477.
15. Breig A, Troup JD. Biomechanical considerations in the straight-leg-raising test: cadaveric and clinical studies of the effects of medial hip rotation. *Spine (Phila Pa 1976)*. 1979;4(3):242–250.
16. Halbertsma JP, Göeken LN, Hof AL, Groothoff JW, Eisma WH. Extensibility and stiffness of the hamstrings in patients with nonspecific low back pain. *Arch Phys Med Rehabil*. 2001;82(2): 232–238.
17. Halbertsma JP, Mulder I, Göeken LN, Eisma WH. Repeated passive stretching: acute effect on the passive muscle moment and extensibility of short hamstrings. *Arch Phys Med Rehabil*. 1999; 80(4):407–414.
18. Halbertsma JP, van Bolhuis AI, Göeken LN. Sport stretching: effect on passive muscle stiffness of short hamstrings. *Arch Phys Med Rehabil*. 1996;77(7):688–692.
19. Boyd BS, Puttlitz C, Gan J, Topp KS. Strain and excursion in the rat sciatic nerve during a modified straight leg raise are altered after traumatic nerve injury. *J Orthop Res*. 2005;23(4):764–770.
20. Babbage CS, Coppieters MW, McGowan CM. Strain and excursion of the sciatic nerve in the dog: biomechanical considerations in the development of a clinical test for increased neural mechanosensitivity. *Vet J*. 2007;174(2):330–336.
21. Gajdosik RL, LeVeau BF, Bohannon RW. Effects of ankle dorsiflexion on active and passive unilateral straight leg raising. *Phys Ther*. 1985;65(10):1478–1482.
22. Læssøe U, Voigt M. Modification of stretch tolerance in a stooping position. *Scand J Med Sci Sports*. 2004;14(4):239–244.
23. McHugh MP, Johnson CD, Morrison RH. The role of neural tension in hamstring flexibility. *Scand J Med Sci Sports*. 2012;22(2):164–169.
24. Hall T, Zusman M, Elvey R. Adverse mechanical tension in the nervous system? Analysis of straight leg raise. *Man Ther*. 1998;3(3): 140–146.
25. Boyd BS, Wanek L, Gray AT, Topp KS. Mechanosensitivity of the lower extremity nervous system during straight-leg raise neurodynamic testing in healthy individuals. *J Orthop Sports Phys Ther*. 2009;39(11):780–790.
26. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol*. 2000;10(5):361–374.
27. Herda TJ, Costa PB, Walter AA, et al. Effects of two modes of static stretching on muscle strength and stiffness. *Med Sci Sports Exerc*. 2011;43(9):1777–1784.
28. Gajdosik RL, Vander Linden DW, McNair PJ, Williams AK, Riggin TJ. Effects of an eight-week stretching program on the passive-elastic properties and function of the calf muscles of older women. *Clin Biomech (Bristol, Avon)*. 2005;20(9):973–983.
29. Nordez A, Cornu C, McNair P. Acute effects of static stretching on passive stiffness of the hamstring muscles calculated using different mathematical models. *Clin Biomech (Bristol, Avon)*. 2006;21(7): 755–760.

30. Boland RA, Adams RD. Effects of ankle dorsiflexion on range and reliability of straight leg raising. *Aust J Physiother*. 2000;46(3):191–200.
31. Herrington L, Bendix K, Cornwell C, Fielden N, Hankey K. What is the normal response to structural differentiation within the slump and straight leg raise tests? *Man Ther*. 2008;13(4):289–294.
32. Coppieters MW, Alshami AM, Babri AS, et al. Strain and excursion of the sciatic, tibial, and plantar nerves during a modified straight leg raising test. *J Orthop Res*. 2006;24(9):1883–1889.
33. Butler D. Adverse mechanical tension in the nervous system: a model for assessment and treatment. *Aust J Physiother*. 1989;35(4):227–238.
34. Fleming P, Lenehan B, O'Rourke S, McHugh P, Kaar K, McCabe JP. Strain on the human sciatic nerve in vivo during movement of the hip and knee. *J Bone Joint Surg Br*. 2003;85(3):363–365.
35. Kwan MK, Wall EJ, Massie J, Garfin SR. Strain, stress and stretch of peripheral nerve: rabbit experiments in vitro and in vivo. *Acta Orthop Scand*. 1992;63(3):267–272.
36. Millesi H, Zöch G, Reihsner R. Mechanical properties of peripheral nerves. *Clin Orthop Relat Res*. 1995;314:76–83.
37. Lew PC, Briggs CA. Relationship between the cervical component of the slump test and change in hamstring muscle tension. *Man Ther*. 1997;2(2):98–105.

Address correspondence to Brennan J. Thompson, PhD, CSCS, Department of Health, Exercise, and Sport Sciences, Texas Tech University, 3204 Main Street, Box 43011, Room 104, Lubbock, TX 79409-3011. Address e-mail to brennan.thompson@ttu.edu.