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Altered lower extremity muscle activity patterns due to Iliopsoas tightness during single-leg landing

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This study aims to investigate the impact of iliopsoas (IL) tightness on lower extremity muscle activity during single-leg landing, focusing on how IL tightness influences joint protection through feed-forward and feed-back pathways that address known impaired neuromuscular mechanisms and provide a set of variables with which to assess and design the ongoing change from both prevention and management. A cross-sectional study of 28 male soccer players (ages 11–14 yrs) divided into IL tightness ($n = 14$) and normal hip flexor length ($n = 14$) groups assessed hip extension range using the modified Thomas test. Electromyography recorded muscle activity (gluteus maximus [GM], adductor magnus [AM], biceps femoris [BF], rectus femoris [RF], soleus [SOL], and multifidus [MF]) during single-leg landing, with RMS values computed over 50 ms epochs, collected 300 ms before and after ground contact, and normalized to maximal voluntary isometric contractions (MVIC). Statistical analysis using Kolmogorov-Smirnov and homoscedasticity tests confirmed normal distribution and homogeneity. Independent-sample t-tests compared muscle activity between groups and Cohen's d effect size was calculated. All analyses were done using SPSS with significance set at $p \le 0.05$. Specifically, participants with IL tightness had reduced activation of the RF ($p = 0.01$) and SOL ($p = 0.003$) during feed-forward action and increased activation of the MF compared to the normal group ($p = 0.008$). During feed-back action, those with IL tightness demonstrated increased activation of the GM ($p = 0.01$), BF ($p = 0.03$), AM ($p = 0.01$) and MF ($p = 0.017$), whereas showing reduced activation of the RF ($p = 0.02$) and SOL ($p = 0.01$). Subtle differences were observed in how adolescent soccer players with iliopsoas tightness utilize their lower extremity muscles through kinetic chains during single-leg landing compared to healthy controls. These findings enhance our understanding of the complex functional consequences of iliopsoas tightness on motor control changes and underscore the importance of monitoring the effectiveness of interventions aimed at joint protection in this specific demographic.

Keywords Iliopsoas tightness, Muscle recruitment, EMG, Single-leg landing, Joint protection

Restricted hip flexor muscle length or "tightness" assessed via hip extension range of motion has been identified as a risk factor for various lower extremity musculoskeletal injuries[1,2]. Lower extremity injuries are a prevalent concern in both athletic and clinical settings, often leading to significant functional impairments and prolonged recovery periods[3] Among the various factors contributing to these injuries, iliopsoas (IL) tightness has garnered attention due to its profound impact on muscle recruitment patterns and joint biomechanics during dynamic activities[4].

The iliopsoas (IL) is the primary hip flexor and may assist in anteriorly tilting the pelvis, making it important in pelvic rehabilitation. It also functions as an external rotator of the hip[5,6]. Because the IL connects the spine and lower extremities, it plays an important role in many activities of daily living, including sports[7]. The IL stabilises the pelvic and hip, especially during single-leg stance activities[6–8].

In the closed kinematic chain, the interaction between the hip, knee, and ankle joints plays a crucial role in distributing forces during landing, which may influence injury risk[9]. Like many athletic manoeuvres, jumping, such as running and cutting, depends on accurate sensory input and appropriate motor responses[10]. They are

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inherently unstable and require neuromuscular control throughout the kinetic chain to maintain stability[10]. The landing task in this study is motivated by its multifaceted nature, encompassing antigravity function, upright posture, balance and sway, with a recognition of the challenges associated with controlling joint loading and individual antigravity muscle recruitment during treatment[11]. The landing task is particularly pertinent due to its association with jump-landing injuries, making it a realistic scenario to assess the impact of iliopsoas muscle shortness on athletes[1,12].

The documented relationship between tight hamstrings and reduced hip flexion range of motion is one of the contributing factors to low back pain[13], alongside other biomechanical and neuromuscular influences. While the literature extensively covers strategies for lower limb muscle activation during landing[14–19], a notable gap exists as the tightness in the hip flexors, especially the IL, and its impact on reducing hip extension have been largely neglected in a significant portion of the research. Additionally, there is a lack of literature addressing the influence of tight hip flexors on muscular activity in the kinetic chain. A limited controversial study has focused on examining the influence of hip flexor tightness, not the IL specifically, solely on the activity of the gluteus maximus (GM) and biceps femoris (BF)[1]. Differently, Christensen et al. indicated that tight hips do not impact the squat technique[20]. Evans et al.[21] identified reduced hip flexor length as a predictor for nonspecific chronic low back pain (nsCLBP) in elite golfers, while Roach et al.[22] observed limited passive hip extension in subjects with nsCLBP. Stiffness or tension in the hip flexors restricts the range of motion available during hip extension movements.

The primary focus of this study is on conducting an electromyographically focused investigation to elucidate and characterise the proximal neuromuscular mechanisms influenced by IL tightness. Our hypothesis posits that tightness in the IL influences the recruitment patterns of the lower extremity muscles within the kinetic chain. This hypothesis is based on the approach presented by Richardson et al. in 2004[23], highlighting the significant role of the load transfer pathway in influencing various muscles. This influence is particularly notable in weight-bearing muscles, such as the GM, adductor magnus (AM) and soleus (SOL), which provide mechanical support and joint protection during functionally loaded postures. These muscles are intricately linked to the function of the local muscle system[23–26]. The local muscle system refers to a group of deep, intrinsic muscles that play a critical role in stabilizing the spine at the segmental level. These muscles are activated in a feedforward manner during movement to ensure dynamic stability, thereby reducing the risk of injury[23–26]. As described by Webster et al. (2016) and Hides et al. (2011), feed-forward neuromuscular control occurs before landing, allowing anticipatory muscle activation to stabilize the lower extremity, while feedback mechanisms are triggered after ground contact to adjust muscle activity in response to external forces[27,28]. Additionally, non-weight-bearing muscles like the BF and rectus femoris (RF), multijoint muscles, play multiple roles in joint movement but do not assume a weight-bearing role in normal erect working posture[23,29] like jump- landing tasks. The integrated function of both weight-bearing and non-weight-bearing muscles contributes to antigravity joint stiffness and joint protection. However, disturbances in the coordinated activation of these muscle groups may alter joint mechanics and compromise its ability to counteract gravitational forces at each limb joint[23]. On a separate note, the multifidus (MF), a core muscle, is emphasised as a critical factor influencing lower limb functioning alongside abdominal muscles[30]. The MF plays a key role in lumbar lordosis; however, its function is part of a broader system involving multiple stabilizing muscles that contribute to overall lumbar stability[31]. As studies on bed rest have shown, the removal of axial gravitational loading may reduce the necessary stimulus for its normal function, highlighting the importance of the MF in maintaining lumbar stability[31]. These muscles contract in anticipation of reactive forces during lower limb movements, providing a stable foundation for safe and controlled movement distal to the core[30]. Additionally, we aim to evaluate whether the feed-forward action of the MF in the presence of IL tightness is affected.

Such associations are relevant to screening physical examinations, aiming to detect deficiencies in motion or muscular activation patterns predisposing an athlete to injuries. Conducted as a pilot study, our research aims to determine whether the antigravity system is disrupted in athletes with IL tightness. In the event of confirming such disruptions, we envision developing and testing tailored exercises designed for each specific muscle segment, taking into account the nature of the muscular dysfunction. This research contributes to a deeper understanding of the implications of IL tightness on athletes, providing valuable insights for injury prevention, rehabilitation and tailored training programs in sports. Such an analysis would be critical to informing evidence-based prevention strategies.

Methods

In this cross-sectional study design, fourteen healthy soccer players (age 11–14 yrs) and fourteen soccer players with hip flexor tightness (age 11–14 yrs) with two years' experience were recruited to participate in this study. A priori power analysis was conducted using G*Power 3 software to determine the necessary sample size for the study[32]. Based on Cohen's conventions, an effect size of 0.80 was selected. With a desired statistical power of 0.80 and an alpha level of 0.05, the analysis indicated that a minimum of 24 participants would be required to achieve sufficient power. Participants were included in the study if they met the following criteria:[1] male soccer players aged 11–14 years[2], at least two years of soccer experience with regular participation in training and competitions[3], no history of lower extremity musculoskeletal injuries in the past six months[4], no prior hip or lower limb surgery[5], no neurological disorders affecting movement, and[6] willingness to comply with the study protocol, including refraining from intense physical activity 24 h before testing[7]. All participants had bilateral IL tightness.

All the participants signed and read a written assent form. Informed consent was granted by the parents or legal guardians after the testing procedures that would be performed during the study were thoroughly explained. The study was approved by the Nahavand University Research Ethics Committee (IR.NAHGU.REC.1399.013, 2021) in accordance with the Declaration of Helsinki. The participants filled out a form with

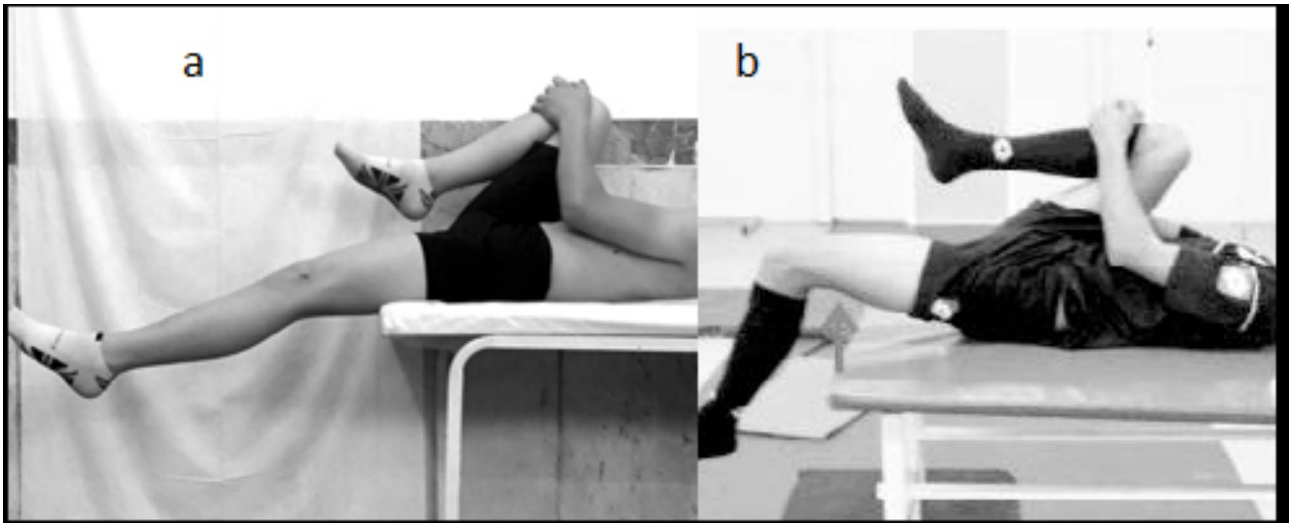

Fig. 1. Modified Thomas Test: Image (**a**) demonstrates a negative test result, indicating normal iliopsoas flexibility with full hip extension as the thigh rests on the table. Image; (**b**) depicts a positive test result, suggesting iliopsoas tightness, characterized by an inability of the thigh to reach the table, indicating restricted hip extension.

Reliability Type	Side	Examiner / Trial	ICC Value	Mean ICC
Intra-rater	Left	Examiner 1	0.939	
		Examiner 2	0.858	0.899
	Right	Examiner 1	0.94	
		Examiner 2	0.907	0.923
Inter-rater	Left	Trial 1	0.844	
		Trial 2	0.898	0.871
	Right	Trial 1	0.785	
		Trial 2	0.877	0.831

Table 1. Intra-rater and Inter-rater reliability ICC values for the modified Thomas test.

demographic information (name, age, height, mass, soccer experience, medical condition). The IL tightness of participants with limited hip extension range of motion was determined using a modified Thomas test and a universal goniometer[6,33] with an accuracy of ± 1 degree, manufactured by LTD (Japan).

To measure the hip extension range of motion, using modified Thomas test (Fig. 1), the subject was sitting on the edge of the examination table and then lying supine so that the tail was at the end of the table. Subject's opposite leg is kept in full flexion (this helps to maintain posterior pelvic tilt and flattened lower back and is essential to avoid stress on the spine[6,34,35]. To minimize the potential measurement error due to posterior pelvic tilt during full flexion in the modified Thomas test, examiners manually stabilized each participant's pelvis throughout the procedure. This stabilization technique, which has been shown to improve the validity of the modified Thomas test when pelvic tilt is controlled[33], was consistently applied by our examiners. All measurements were conducted by two experienced biomechanical assessors, each with over five years of clinical practice. Intra-rater reliability was excellent (ICC > 0.85), as detailed in the Table 1. To assess, the center of the goniometer was fixed on the center of the greater trochanter of the hip, the fixed arm was parallel to the axillary line of the trunk, and the movable arm was placed parallel with the longitudinal line of the femur toward the lateral femoral epicondyle[34,36]. The angle was less than 180 degrees, indicating IL shortness; the values were between 180 and 190 in persons with normal IL length. Before measuring the hip extension angle, differential tests were used to reject the RF and tensor fascia lata shortness, according to Kendal et al.[34]. We adopted an acceptable measurement error of approximately ± 1° for clinical assessments[37]. In our study, reliability analysis revealed an average standard error of measurement (SEM) of 0.51°, aligning with expected variability in the Modified Thomas Test due to minor deviations in anatomical landmark identification and participant factors.

Participants wore standardised footwear (Mizuno K1GA140009, Lurng Furng, Inc., Taipei, Taiwan) to control the effect of differences in footwear properties. Before data collection, participants performed a 15-minute warm-up, which included a 5-minute run on a treadmill. This was followed by a 5-minute dynamic stretching routine targeting the lower extremity muscles, including the quadriceps, hamstrings, calves, and hip flexors, with each stretch held for 30 s[38–40].

A baseline protocol consisting of 3 maximal vertical jumps was performed after an initial warm-up period. Before performing the vertical jump, the subjects were instructed in the proper jumping technique. This involved performing a one-legged vertical countermovement jump[41] with the foot on a stable surface with maximal effort while the hands were on the hips. The dominant leg is identified by having the individual kick a ball[12,39].

For a single-leg jump landing task, the subject was asked to stand on a 40-cm high platform with hands on the hips. Then, bend the non-dominant foot from the knee while standing on the other foot. The subject was then asked to jump up to 5 cm and land with the dominant foot while maintaining balance for 3 s. The 5 cm jump height was measured using a dual-method approach to ensure both precise measurement and clear participant guidance. A physical marker—a bar or horizontal tape—was installed exactly 5 cm above the edge of the platform, requiring participants to execute their jump in such a way that the lateral edge of their foot would reach the marker. This ensured a standardized jump height across trials. Additionally, according to the study by Fitzgerald et al.,[42] a high-speed motion capture system with reflective markers placed on the subject's sacrum was used to track the vertical displacement of the center of mass during the jump. The system was carefully calibrated before data collection, and pilot tests confirmed its accuracy in capturing a 5 cm displacement with a measurement error of less than 1 cm. The repeatability level of jump height in the jump landing task is high. Fitzgerald et al. reported a high Intra-class Correlation Coefficient (0.84) for the repeatability level of jump height[42].

A force-sensitive shoe insole was used to detect the foot-to-ground contact during landing[43]. The shoe insole contains three force sensing resistors (1.5-cm diameter, FSR 402, Interlink Electronics, Irvine, USA), one on the heel area and two on the forefoot area, as denoted in Fig. 2. During the experiment, all participants used the force-sensitive shoe insole and the RFduino unit (Arduino, Somerville, USA) to detect the foot-to-ground contact and send the data via Bluetooth to a laptop to log the results. The force-sensitive insole data were primarily used to determine the foot-to-ground contact during the landing task, ensuring accurate synchronization with the EMG data.

Subjects were permitted a period of practice in the jumping technique before testing. A jump was discarded if the subject required any corrections following landing, such as readjusting their position on the floor or touching the floor with the opposite foot. Each subject performed five single-leg jumps. None of the subjects involved in the study reported any subjective difficulty with the jumping technique or discomfort during testing.

A wireless surface electromyography (EMG) system (8 channels, inter-electrode distance = 20 mm, 20–450 Hz band pass filtering, input impedance > 1015//0.2 w//pf, sampling frequency = 1000 Hz, ME6000, Mega Electronics Ltd, Kuopio, Finland) was used to record muscle activity. EMG signals were collected from the Gluteus Medius (GM), Biceps Femoris (BF), Rectus Femoris (RF), Soleus (SOL), and Adductor Magnus (AM) using the SENIAM protocol[12,39,44]. Electrode placements were as follows: For GM, the electrode was positioned at the midpoint between the iliac crest and the greater trochanter; for BF, at the midpoint of the line connecting the ischial tuberosity and the fibular head; for RF, at the midpoint between the anterior superior iliac spine (ASIS) and the superior patella; and for SOL, over the lateral aspect of the muscle belly[12,39,44]. The AM electrode was placed halfway between the pubic tubercle and the medial femoral epicondyle over the bulk of the adductor muscles[25]. We assumed there was no bilateral difference in this study. Each participant's skin was carefully shaved, abraded and cleaned with alcohol before the attachment of the electrode. Medical adhesive tape was used to fix electrodes and probes on the skin to minimise motion artefacts. Before recording EMG activity during the jump landing task, round force sensing resistors (18.28 mm in diameter, FSR 402, Interlink Electronics, Irvine, USA) were used to determine the foot-to-ground contact during the jump landing phase, ensuring accurate synchronization with the EMG data The sensors were placed under the heel first metatarsal bone and thumb of the dominant leg.

Electromyographic (EMG) data were recorded from 300 milliseconds before foot contact to 300 milliseconds after foot contact during landing[39]. The raw EMG signals were band-pass filtered (20–450 Hz) using a fourth-order Butterworth filter to remove movement artefacts[40,45,46] followed by full-wave rectification. The raw data were processed into root-mean-square (RMS) values smoothed using a 50-millisecond root mean square algorithm[28,45,46]. The linear envelope of the signal was then computed and normalized to the highest maximum voluntary isometric contraction (MVIC) value. Specifically, the EMG data collected before and after foot contact during landing were normalized using MVIC values as the reference.

For MVIC testing, three trials were performed for each of the eight analyzed muscles using standard manual muscle test positions. Each MVIC contraction was sustained for five seconds, with a three-second rest between repetitions, and a 30-second rest was provided between different muscles[47]. The highest MVIC value among the three trials was used for normalization. Following MVIC data collection, participants rested for five minutes before additional experimental procedures. Table 2 provides detailed descriptions of the muscle-specific MVIC test positions. Following MVIC data collection, participants rested five minutes before additional data collection.

The statistical analysis was initially performed using the Kolmogorov-Smirnov normality and homoscedasticity tests. All the variables presented normal distribution and homoscedasticity. The independent t-test was used to compare the two groups' GM, BF, AM, SOL and MF activity during the jump landing task. Cohen's d of effect size for the differences was calculated to indicate the practical relevance of the significance. All analyses were done using SPSS (SPSS, IBM, Armonk, USA) and significance for all analyses, before any corrections, was set at $p \leq 0.05$.

Results

Table 3 concisely presents the demographic characteristics of participants in the IL tightness and normal groups, encompassing age, height, weight and body mass index (BMI). Independent t-tests revealed no statistically significant differences between the groups ($p \geq 0.05$). The study comprised 14 male participants within the age range of 11–14 years. Both groups exhibited comparable mean values for age (12.5 ± 1 yrs), mass (37.8 ± 4.6

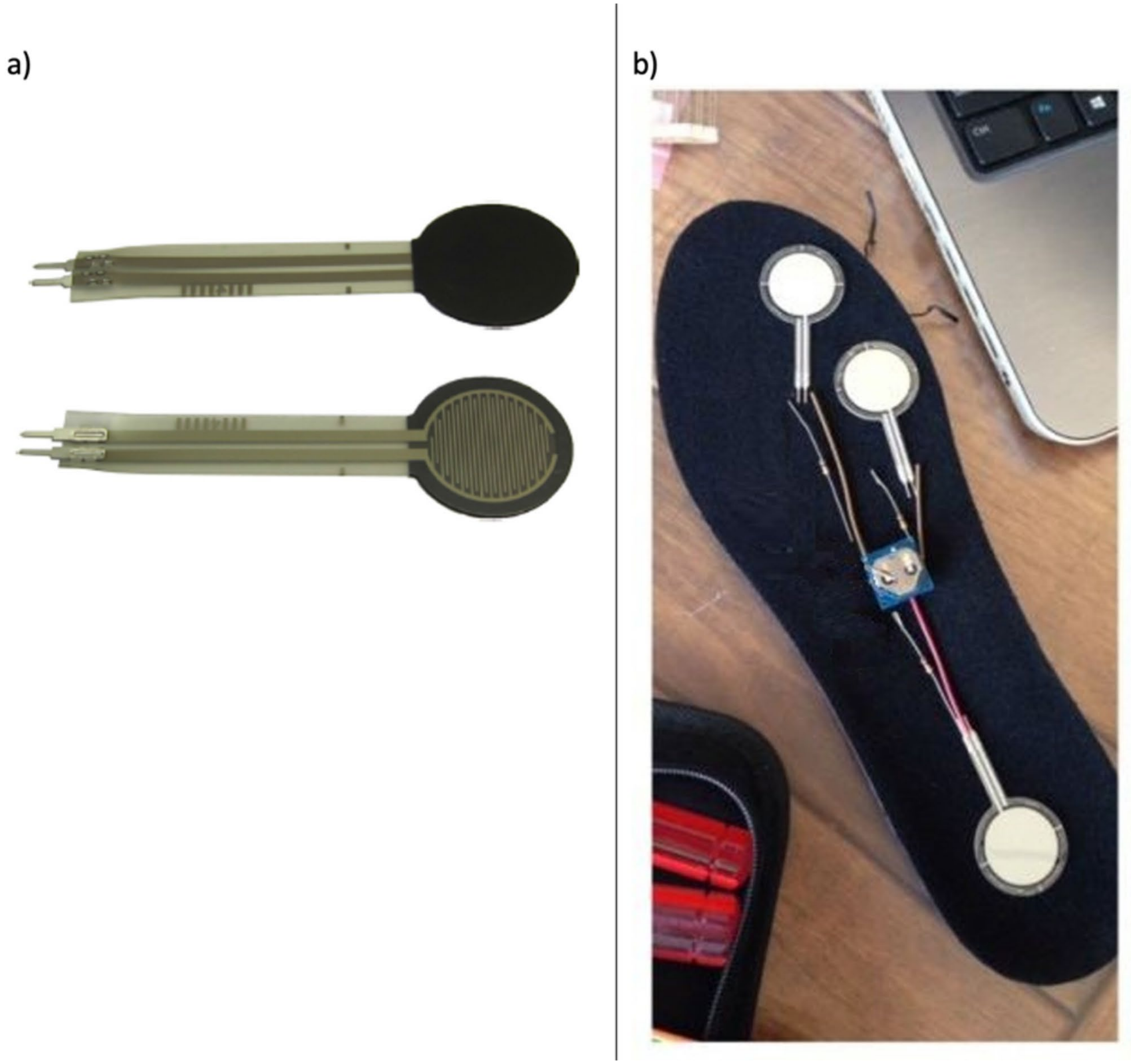

Fig. 2. (**a**) Force sensing resistors (1.5-cm diameter, FSR 402). (**b**) Placement of sensors: the shoe insole contains three force sensing resistors, one on the heel area and two on the forefoot area.

GM	The subject was positioned in prone for the GM, with the knee flexed to 90 degrees and the hip fully extended. Manual resistance was applied on the lower part of the posterior thigh as the hip moved into extension[12,48].
BF RF	In short sitting with the knees flexed to 90 degree using a gait belt around the distal third of the shank during both isometric knee extension and knee flexion. Ninety degrees was used to normalize quadriceps and hamstring activation to maximal activity during peak knee flexion[12,49].
AM	To perform the maximal isometric contractions, participants were positioned in supine with the test limb in a neutral hip position in the frontal plane[25]. A strap was placed around the ankle of the test leg and a stable metal. Contralateral anterior superior iliac spine and ankle were stabilized. Participants were instructed to pull the test leg against the ankle strap in the direction toward the midline as hard as possible without bending or rotating through the hip or bending the knee[25].
MF	For the MF, subjects were positioned in prone, with the arms extended beside the body and the thighs and legs fixed on the ground with the assistance of manual resistance imposed by two assessors. Meanwhile, another assessor applied a resistance in the upper torso, in the opposite direction to the trunk extension movement performed by the subject[50].
SOL	The SOL MVC was tested in a seated position with the knee flexed to 90. A rigid seat belt strap was placed over the knee and secured to the floor providing resistance for the maximal test[12,19,51].

Table 2. Description of the maximum voluntary isometric contraction (MVIC) tests for gluteus Maximus (GM), biceps femoris (BF), rectus femoris (RF), adductor Magnus (AM), soleus (SOL) and multifidus (MF).

	IL tightness	Normal	Sig
N	14	14	NA
Age (year)	12.5 ± 1.0	12.5 ± 1.4	0.635
Mass (kg)	37.8 ± 4.6	37.7 ± 3.3	1.000
Height (cm)	145.9 ± 4.9	148.2 ± 5.7	0.443
BMI	17.0 ± 0.8	17.2 ± 0.3	0.599
Gender	Male	Male	NA
Range of age	11–14	11–14	
Modified Tomas Test	< 180 degrees 170.69 ± 1.73	Between 180 to 190 degrees 185 ± 2.85	0.003*

Table 3. Characteristics of participants. Values are mean ± standard (SD) deviation. n, number of participants; BMI, body mass index; NA, not applicable. *Significance level $p < 0.05$.

	Muscle	Group	Mean (SD)	t	Sig	Cohen's D
Feed-forward	GM	IL tightness	16.12 ± 5.34	-1.37	0.18	0.082
		normal	19.23 ± 6.82			
	BF	IL tightness	28.61 ± 10.75	-0.57	0.573	-0.023
		normal	26.81 ± 6.03			
	RF	IL tightness	9.87 ± 2.89	2.77	0.01*	0.934
		normal	22.61 ± 4.35			
	AM	IL tightness	19.25 ± 6.38	1.28	0.21	0.192
		normal	21.0 ± 11.86			
	MF	IL tightness	39.94 ± 6.92	-3.89	0.008*	1.197
		normal	32.15 ± 6.06			
	SOL	IL tightness	11.75 ± 3.39	3.27	0.003*	0.160
		normal	20.0 ± 9.57			

Table 4. Mean EMG values in the 300 milliseconds window before ground contact (feed-forward action). Data are mean ± sd, expressed as percent maximum voluntary isometric contraction (%MVIC). *Significance level $p < 0.05$.

and 37.7 ± 3.3 kg), height (145.9 ± 4.9 and 148.2 ± 5.7 cm) and BMI (17.0 ± 0.8 and 17.2 ± 0.3 $m^2 \cdot kg^{-1}$) for the IL tightness and normal groups, respectively. These numerical details underscore the homogeneous demographic profiles of the participants, reinforcing the equivalence between the two study groups. In the IL tightness group, 12 participants had a dominant right leg and 2 had a dominant left leg. In the normal group, 13 participants had a dominant right leg and 1 had a dominant left leg.

The data in Table 4 elucidated pre-landing EMG patterns in individuals with IL tightness compared to a normal group, expressed as a percentage of maximum voluntary isometric contraction during a 300-millisecond window before ground contact. GM and AM EMG activities were lower in the IL tightness group. However, that was not statistically significant. BF activity was greater in the IL tightness group but not statically significant (GM: $p = 0.18$, AM: $p = 0.21$, BF: $p = 0.573$). On the other hand, RF exhibited a statistically significant difference ($p = 0.01$*) with lower EMG values in the IL tightness group. MF demonstrated (MF: $p = 0.008$*) a higher EMG value in the IL tightness group, while the SOL activity was significantly lower in the IL tightness group (SOL: $p = 0.003$*).

Table 5 displays the mean EMG values during the 300-millisecond window following ground contact for the IL tightness and normal groups. The results highlight distinct muscle activation patterns in individuals with IL tightness compared to those with normal muscle function following ground contact. All three hip extensors in the present study, including GM, AM and BF, showed greater EMG activity in the presence of IL tightness compared to the normal group, with p-values of 0.01*, 0.01* and 0.03*, respectively. However, the RF and SOL activities continue to decrease, and the MF activity continues to increase in the 300-millisecond window following ground contact in the IL tightness group with p-values of 0.02*,0.01* and 0.017*, respectively.

Discussion

The current study compared the lower extremity muscles' activity in subjects with and without IL tightness during single-leg landing. A crucial aspect of this investigation is examining how alterations in the feed-back and feed-forward pathways through IL tightness impact joint protection.

The muscular activity in the 300 milliseconds before ground contact (i.e., feed-forward response) of the GM and AM was lower in the IL tightness group but not statistically significant. The BF activity was greater in the IL tightness group but not statistically significant. In addition, the RF exhibited a statistically significant difference

	Muscle	Group	Mean (SD)	t	Sig	Cohen's D
Feed- back	GM	IL tightness	41.38 ± 9.55	-2.77	0.01*	-0.324
		normal	23.43 ± 4.41			
	BF	IL tightness	48.15 ± 11.7	-2.77	0.01*	-0.240
		normal	29.5 ± 4.28			
	RF	IL tightness	10.57 ± 1.89	2.47	0.02*	0.675
		normal	21.52 ± 5.37			
	AM	IL tightness	35.25 ± 4.41	-2.29	0.03*	-0.246
		normal	28.5 ± 3.32			
	MF	IL tightness	60.34 ± 7.46	-2.70	0.017*	1.045
		normal	52.32 ± 7.88			
	SOL	IL tightness	30.93 ± 5.17	2.77	0.01*	0.238
		normal	50.0 ± 11.54			

Table 5. Mean EMG values in the 300-millisecond window following ground contact (feed-back action). Data are mean ± sd, expressed as percent maximum voluntary isometric contraction (%MVIC). *Significance level $p < 0.05$.

with lower EMG values in the IL tightness group. MF demonstrated a higher EMG value in the IL tightness group, whereas the SOL activity was significantly lower in the IL tightness group.

In the 300 milliseconds following ground contact (i.e., feed-back response), all three hip extensors, including GM, AM and BF, showed greater EMG activity in the presence of IL tightness than the normal group. However, the RF and SOL activities were lower and the MF activity was greater in the 300 milliseconds following ground contact in the IL tightness group.

Our finding highlighted that IL tightness altered muscle recruitment patterns through the kinetic chain. It is believed that, in weight-bearing activities (e.g. landing), most of the applied forces to the body were absorbed by the lower extremities. Similarly, lower limb muscles are feed-forwardly activated in the pre-landing phase to absorb the impact forces. After ground contact, muscles also eccentrically try to absorb inserted forces[11]. This altered force development may reduce contact forces on the joints, consistent with studies on the erector spine, BF and GM[52,53]. Prior studies showed that in combined hip and knee extension exercises (closed-chain), the vastii are strongly recruited, whereas the RF is not[54]. Similarly, the current study found lower RF contributions during landing, possibly due to its bi-articular role and sacroiliac joint (SIJ) position[12]. Decreased neuromuscular control of the lumbo-pelvic region via RF and increased BF activity impair joint stability, with a greater reliance on uni-articular muscles during simultaneous hip and knee extension[12].

GM is proposed to have three functional subdivisions, with the superficial sacral fibres prone to overactivity and shortness, leading to compensatory movements[55,56]. These fibres extend into the iliotibial band, making GM a multi-joint muscle susceptible to over-activity[57]. Current research did not examine iliotibial band activity, suggesting a direction for future studies. Biomechanical studies consistently associate insufficient extension of the hip joint accompanied by pelvic compensation with alteration in GM contraction and triggering hamstring muscle dominance[4]. It is believed that the short head of BF is involved solely in knee flexion, whereas the other members of the hamstrings group are also involved in hip extension moments[29]. Hamstrings, with anatomical connections to the sacrotuberous ligament and ischial tuberosity, are integral to SIJ stability, producing nutation torque alongside the erector spinae and rectus abdominis[52], which is consistent with our finding regarding increased activation of BF throughout landing. However, it was not statically significant during the 300 milliseconds before ground contact. One explanation for this finding could be the reciprocal inhibition mechanism. In other words, tight hip flexors can cause altered reciprocal inhibition, reducing the neural drive to the functional antagonist (GM) during hip extension[1,58]. This results in synergistic dominance, where the hamstrings overcompensate for the inhibited GM during hip extension, thereby increasing the risk of hamstring injury through arthrokinetic dysfunction during sprinting and jumping movements[59–61].

AM also plays an important role in the extension of the hip and/or eccentric control of hip flexion[29] and is similarly recruited over the adductor longus during combined hip and knee extension (i.e., closed chain exercises[54]. In the weight-bearing task of the current study, the AM functioned as a primary hip extensor alongside the GM and hamstrings. Compared with the gluteus medius and GM, the recruitment of adductor muscles was moderately associated with the ability to absorb landing forces. Peak AM activity occurs during normal gait's initial contact and loading phases[25]. Our findings suggest that individuals with IL tightness may recruit their hip adductors more during landing to decelerate hip flexion. This is consistent with Hides et al.[25], who found higher AM recruitment than adductor longus in weight-bearing tasks. This supports our hypothesis that IL tightness disrupts the feed-forward response of weight-bearing muscles like GM and AM. Whereas this response helps reduce landing forces, increased adductor recruitment may contribute to medial knee translation, a risk factor for ACL injuries[62].

In the present study, the feed-forward and feed-back response of the MF muscle in the IL tightness group was greater than the normal group. There are several possible reasons why the IL tightness group employed a different MF recruitment strategy to maintain the lumbo-pelvic as a stable unit. First, according to recent studies, the psoas major's anterior and posterior fasciculi have distinct functions, with the posterior fasciculi potentially controlling lumbar segmental translation[63]. The activity of the psoas major from the lumbar transverse process is more influenced by lumbar spine position. In contrast, fascicles from the vertebral body are more involved in the hip movement[64]. Additionally, the psoas major aids lumbar spine stability, particularly in full hip extension when muscle tension is highest[65]. Second, limited hip motion may cause compensatory lumbo-pelvic movements, increasing stress on the lumbar spine[66]. The MF and abdominal muscles contract in anticipation of lower limb movements, with trunk preparations occurring before asymmetric limb motion. Third, the role of MF in controlling lumbar lordosis[31] explains this result. The MF may overcompensate to allow controlled movement and maintain joint stability during locomotion[30]. Our findings are broadly consistent with Willson et al., who found that core strength correlates with lower extremity alignment during weight-bearing exercises[67]. Similarly, Yoo showed that the erector spinae muscle aponeurosis links to the sacrum and dorsal SIJ ligaments, producing nutation movement for SIJ stability[53]. Finally, it should not be neglected that although an increased recruitment strategy may increase the stiffness of the spine, it may be associated with a cost. Increased recruitment of the MF may accelerate fatigue and increase vertebral joint compression.

Here, the lower activity of the SOL during landing among individuals with IL tightness compared with the normal group remains a question. Our findings of reduced SOL activity during landing in individuals with IL tightness align with Goldberg and Neptune's research on muscle compensatory strategies[68]. Their dynamic walking simulations showed that reduced SOL contribution leads to increased positive work from the gastrocnemius (GAS) and negative work from the vastii and RF, with a slight increase in GM work. Conversely, decreased GAS strength results in increased positive work from the IL and negative work from the SOL, minor increases in GM work and negative work from the short head of the BF[69]. These findings suggest that IL tightness may impair the neural drive to the SOL. The SOL, a uni-articular muscle, and the bi-articular GAS are crucial for vertical support by modulating the anteroposterior momentum of the centre of mass. The SOL and GAS generate an anteriorly directed impulse, whereas the vastii and RF produce a posteriorly directed impulse[68]. Therefore, neuromuscular control loss in the SOL due to tight IL could disrupt the kinetic chain's integrity, leading to altered movement patterns and compensatory muscle activity during landing. Prospective studies are required to confirm the relationship between the reduced contribution of the SOL and altered movement patterns through the kinetic chain as a basis for designing proper interventions.

One major limitation of the study was the absence of a biomechanical analysis of changes in the presence of IL tightness. Future research should explore the correlation between IL tightness and kinetic and kinematic alterations in the kinetic chain. Subsequent studies may benefit from examining muscle activity during various external loads and athletic movements. Another constraint was the inability to record EMG data of the IL and other deep muscles due to the lack of needle electromyography and the omission of muscle activation timing analysis in the current study. Further investigations could be guided by addressing these issues. Based on Hides et al. (2011), neuromuscular control and lumbopelvic stability play a crucial role in injury prevention, particularly in dynamic tasks like jumping and landing. Given that altered spinal alignment and deficits in core muscle function can impact lower extremity biomechanics, we acknowledge that such factors could have influenced our findings. While our study did not directly assess postural changes or joint play, future research should consider incorporating lumbopelvic assessments to better understand their impact on neuromuscular control during landing. We have now addressed this limitation in the discussion and refined our conclusions accordingly[27].

Kinematic variations in hip, knee, and ankle positions during the analysis window may have influenced muscle length and the EMG pickup area, potentially affecting RMS comparisons. Although normalization to MVIC reduces variability, it does not fully account for these biomechanical differences. Future studies should consider kinematic segmentation to better control for these confounding factors.

Conclusions

This study is the first to report a trend toward altered recruitment patterns of the lower extremity muscles in a weight-bearing task in those with IL tightness. IL tightness significantly impacts muscle recruitment patterns during single-leg landing, affecting feed-forward and feed-back responses. This alteration in muscle activity can lead to compromised joint protection and increased injury risk, highlighting the need for addressing. IL tightness in preventative and rehabilitative strategies for lower extremity injuries. The research enhances the increasing understanding of muscle function during dynamic tasks, especially in the context of muscle impairment and may have important implications for athletic training by indicating which muscles may serve as targets for interventions aiming to improve athletic performance.

Data availability

The data supporting this study's findings are available upon reasonable request with corresponding authors.

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References

1. Mills, M. et al. Effect of restricted hip flexor muscle length on hip extensor muscle activity and lower extremity biomechanics in College-Aged female soccer players. *Int. J. Sports Phys. Ther.* **10** (7), 946–954 (2015).
2. Winters, M. V. et al. Passive versus active stretching of hip flexor muscles in subjects with limited hip extension: a randomized clinical trial. *Phys. Ther.* **84** (9), 800–807 (2004).
3. De Bleecker, C. et al. Relationship between Jump-Landing kinematics and lower extremity overuse injuries in physically active populations: A systematic review and Meta-Analysis. *Sport Med.* **50** (8), 1515–1532. https://doi.org/10.1007/s40279-020-01296-7 (2020).

4. Park, K., Jeon, I., Hwang, U., Jung, S. & hoon, Ha, S. min. Effects of abdominal drawing-in maneuver with or without prior iliopsoas stretching on gluteus maximus activity during prone hip extension un co rre ct pr oo f v er on pr ct rre un co. J Back Musculoskelet Rehabil. 1:1–9. (2021).
5. Blankenbaker Michael, J. D. G. T. Iliopsoas musculotendinous unit. *Semin Musculoskelet. Radiol.* **12** (01), 13–27. https://doi.org/10.1055/s-2008-1067934 (2008).
6. Lifshitz, L., Bar Sela, S., Gal, N., Martin, R. & Fleitman Klar, M. Iliopsoas the hidden muscle: anatomy, diagnosis, and treatment. *Curr. Sports Med. Rep.* **19** (6), 235–243 (2020).
7. Anderson, C. N. Iliopsoas. *Clin. Sports Med.* ;**35**(3):419–433 (2016).
8. Thorborg, K. et al. Large strengthening effect of a hip-flexor training programme: a randomized controlled trial. *Knee Surg. Sport Traumatol. Arthrosc.* **24** (7), 2346–2352. https://doi.org/10.1007/s00167-015-3583-y (2016).
9. Aerts, I., Cumps, E., Verhagen, E., Verschueren, J. & Meeusen, R. A systematic review of different jump-landing variables in relation to injuries. *J. Sports Med. Phys. Fit.* **53** (5), 509–519 (2013).
10. Powers, C. M. The Influence of Abnormal Hip Mechanics on Knee Injury: A Biomechanical Perspective. J Orthop Sport Phys Ther. 40(2):42–51. https://doi.org/10.2519/jospt.2010.3337 (2010).
11. Mahaki, M., Amsterdam, V. U., Mimar, R. & Mahaki, B. On the relationship between lower extremity muscles activation and peak vertical and posterior ground reaction forces during single leg drop landing. (2019). (2015).
12. Maniar, N., Schache, A. G., Pizzolato, C. & Opar, D. A. Muscle function during single leg landing. *Sci. Rep*. 1–14. https://doi.org/10.1038/s41598-022-15024-w (2022).
13. Fasuyi, F. O., Fabunmi, A. A. & Adegoke, B. O. A. Hamstring muscle length and pelvic Tilt range among individuals with and without low back pain. *J. Bodyw. Mov. Ther.* **21** (2), 246–250 (2017).
14. de Britto, M. A., Carpes, F. P., Koutras, G. & Pappas, E. Quadriceps and hamstrings prelanding myoelectric activity during landing from different heights among male and female athletes. *J. Electromyogr. Kinesiol.* **24** (4), 508–512 (2014).
15. Brown, T. N., McLean, S. G. & Palmieri-Smith, R. M. Associations between lower limb muscle activation strategies and resultant multi-planar knee kinetics during single leg landings. *J. Sci. Med. Sport*. **17** (4), 408–413 (2014).
16. Ellenberger, L. et al. Thigh muscle activation patterns and dynamic knee valgus at peak ground reaction force during drop jump landings: reliability, youth competitive alpine skiing-specific reference values and relation to knee overuse complaints. *J. Sci. Med. Sport*. **24** (12), 1230–1234 (2021).
17. Gambelli, C. N. & Schepens, B. Motor control of landing in an unsteady environment. *Gait Posture*. **95**, 235–241 (2022).
18. Morgan, K. D., Donnelly, C. J. & Reinbolt, J. A. Elevated gastrocnemius forces compensate for decreased hamstrings forces during the weight-acceptance phase of single-leg jump landing: implications for anterior cruciate ligament injury risk. *J. Biomech.* **47** (13), 3295–3302 (2014).
19. Peel, S. A., Schroeder, L. E. & Weinhandl, J. T. Lower extremity muscle contributions to ACL loading during a stop-jump task. J Biomech. 121:110426. (2021).
20. Christensen, B. et al. 37th International Society of Biomechanics in Sport Conference, Oxford, OH, United States, July 21–25, 2019. 320–3. (2019).
21. Evans, K., Refshauge, K. M., Adams, R. & Aliprandi, L. Predictors of low back pain in young elite golfers: A preliminary study. *Phys. Ther. Sport*. **6** (3), 122–130 (2005).
22. Roach, S. M. et al. Passive hip range of motion is reduced in active subjects with chronic low back pain compared to controls. *Int. J. Sports Phys. Ther.* **10** (1), 13–20 (2015).
23. Richardson, C., Hodges, P. W. & Hides, J. *Therapeutic Exercise for Lumbopelvic Stabilization* 2nd edn 141–148 (Churchill Livingstone, 2004).
24. Hides, J. A. et al. The effects of rehabilitation on the muscles of the trunk following prolonged bed rest. *Eur. Spine J.* **20** (5), 808–818 (2011).
25. Hides, J. A. et al. Activation of the hip adductor muscles varies during a simulated weight-bearing task. *Phys. Ther. Sport* **17**, 19–23. https://doi.org/10.1016/j.ptsp.2015.06.001 (2016).
26. Hides, J. A., Endicott, T., Mendis, M. D. & Stanton, W. R. The effect of motor control training on abdominal muscle contraction during simulated weight bearing in elite cricketers. *Phys. Ther. Sport* https://doi.org/10.1016/j.ptsp.2016.05.003 (2016).
27. Hides, J., Brown, C. T., Penfold, L. & Stanton, W. R. Screening the lumbopelvic muscles for a relationship to injury of the quadriceps, hamstrings, and adductor muscles among elite Australian football league players. *J. Orthop. Sports Phys. Ther.* **41** (10), 767–775 (2011).
28. Webster, K. A., Pietrosimone, B. G. & Gribble, P. A. Muscle activation during landing before and after fatigue in individuals with or without chronic ankle instability. 629–636. (2016).
29. Belavy, D. L., Armbrecht, G., Richardson, Æ. C. A. & Felsenberg, Æ. D. Differential atrophy of the lower-limb musculature during prolonged bed-rest. 489–499. (2009).
30. De Blaiser, C. et al. Is core stability a risk factor for lower extremity injuries in an athletic population? A systematic review. *Phys. Ther. Sport* **30**, 48–56 (2018).
31. Hides, J. et al. Magnetic resonance imaging assessment of trunk muscles during prolonged bed rest. *Spine (Phila Pa. 1976)* **32** (15), 1687–1692 (2007).
32. Faul, F., Erdfelder, E., Lang, A. G. & Buchner, A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behav. Res. Methods.* **39** (2), 175–191 (2007).
33. Ferber, R., Kendall, K. D. & McElroy, L. Normative and critical criteria for iliotibial band and Iliopsoas muscle flexibility. *J. Athl Train.* **45** (4), 344–348 (2010).
34. Kendall, P. G., FP. P, Romani, W. A., McCreary, E. K., Rodgers, M. M. & Muscles *Testing and Function with Posture and Pain* 5th edn. 5, 3–9 (Lippincott Williams & Wilkins, 2005).
35. Moreside, J. M. & McGill, S. M. Quantifying normal 3D hip ROM in healthy young adult males with clinical and laboratory tools: hip mobility restrictions appear to be plane-specific. *Clin. Biomech.* **26** (8), 824–829. https://doi.org/10.1016/j.clinbiomech.2011.03.015 (2011).
36. Moreside, J. M. & McGill, S. M. Hip jointrange of motion improvementsusing three different interventions. *J. Strength. Cond Res.* **26** (5), 1265–1273 (2012).
37. Vigotsky, A. D. et al. The modified Thomas test is not a valid measure of hip extension unless pelvic Tilt is controlled. *PeerJ* **4**, e2325 (2016).
38. Tai, W. et al. Biomechanical characteristics of single leg jump in collegiate basketball players based on approach technique. *Appl. Sci.* **10** (1), 309 (2019).
39. Letafatkar, A., Mantashloo, Z. & Moradi, M. Comparison the time to stabilization and activity of the lower extremity muscles during jump-landing in subjects with and without Genu varum. *Gait Posture* **65**, 256–261 (2018).
40. Sheikhi, B., Letafatkar, A. & Thomas, A. C. Comparing myofascial meridian activation during single leg vertical drop jump in patients with anterior cruciate ligament reconstruction and healthy participants. *Gait Posture* **88** (February), 66–71. https://doi.org/10.1016/j.gaitpost.2021.05.006 (2021).
41. Anicic, Z. et al. Assessment of Countermovement Jump: What Should We Report? Life (Basel, Switzerland). 13(1). (2023).
42. Fitzgerald, D., Trakarnratanakul, N., Smyth, B. & Caulfield, B. Effects of a wobble Board-Based therapeutic exergaming system for balance training on dynamic postural stability and intrinsic motivation levels. *J. Orthop. Sport Phys. Ther.* **40** (1), 11–19. https://doi.org/10.2519/jospt.2010.3121 (2010).

43. Morris, S. J. & Paradiso, J. A. Shoe-integrated sensor system for wireless gait analysis and real-time feedback. *Proc. Second Jt. 24th Annu. Conf. Annu. Fall Meet Biomed. Eng. Soc. [Engineering Med. Biol.]* **7** (10), 2468–2469 (2012).
44. Jonkers, I., Stewart, C. & Spaepen, A. The complementary role of the plantarflexors, hamstrings and gluteus Maximus in the control of stance limb stability during gait. *Gait Posture* **17** (3), 264–272 (2003).
45. Ko, H., Jeon, S., Kim, S. & hyun, Park, K. Comparison of hip extensor muscle activity including the adductor magnus during three prone hip extension exercises. Physiother. Theory Pract. 35(5):451–7. https://doi.org/10.1080/09593985.2018.1453569 (2019).
46. Moisan, G., Mainville, C., Descarreaux, M. & Cantin, V. Unilateral jump landing neuromechanics of individuals with chronic ankle instability. *J. Sci. Med. Sport* **23** (5), 430–436 (2020).
47. Studnicki, R. et al. Hip Manipulation Increases Electromyography Amplitude and Hip Joint Performance: A Double-Blind Randomized Controlled Study. Life. Vol. 14, (2024).
48. Boling, M. & Padua, D. Relationship between hip strength and trunk, hip, and knee kinematics during a jump-landing task in individuals with patellofemoral pain. *Int. J. Sports Phys. Ther.* **8** (5), 661–669 (2013).
49. Slater, L. V. & Hart, J. M. Muscle activation patterns during different squat techniques. *J. Strength. Cond Res.* **31** (3), 667–676 (2017).
50. Loss J. F. et al. Electrical activity of external oblique and multifidus muscles during the hip flexion-extension exercise performed in the Cadillac with different adjustments of springs and individual positions. *Revista Brasileira De Fisioterapia*. **14**, 510–517 (2010).
51. Reid, D. A. & McNair, P. J. Effects of an acute hamstring stretch in people with and without osteoarthritis of the knee. *Physiotherapy* **96** (1), 14–21 (2010).
52. van Wingerden, J. P., Vleeming, A., Buyruk, H. M. & Raissadat, K. Stabilization of the sacroiliac joint in vivo: verification of muscular contribution to force closure of the pelvis. *Eur. Spine J.* **13** (3), 199–205. https://doi.org/10.1007/s00586-003-0575-2 (2004).
53. Yoo, W. G. Effects of individual strengthening exercises for the stabilization muscles on the nutation torque of the sacroiliac joint in a sedentary worker with nonspecific sacroiliac joint pain. *J. Phys. Ther. Sci.* **27** (1), 313–314 (2015).
54. Enocson, A. G., Berg, H. E., Vargas, R., Jenner, G. & Tesch, P. A. Signal intensity of MR-images of thigh muscles following acute open-and closed chain kinetic knee extensor exercise–index of muscle use. *Eur. J. Appl. Physiol.* **94**, 357–363 (2005).
55. One, S. & Gibbons, S. Clinical_anatomy_and_function_of_psoas_m. 2004;(Lee).
56. Sahrmann, S. *Diagnosis and Treatment of Movement Impairments Syndromes.* 1st edn 193–244. (Mosby, 2002).
57. Comerford, M. J., Mottram, S. L. & Masterclass Functional stability re-training: principles and strategies for managing mechanical dysfunction. *Man. Ther.* **6** (1), 3–14 (2001).
58. Buckthorpe, M., Stride, M. & Della Villa, F. Assessing and treating gluteus Maximus weakness–a clinical commentary. *Int. J. Sports Phys. Ther.* **14** (4), 655 (2019).
59. Aslan, H., Buddhadev, H. H., Suprak, D. N. & San Juan, J. G. Acute effects of two hip flexor stretching techniques on knee joint position sense and balance. *Int. J. Sports Phys. Ther.* **13** (5), 846–859 (2018).
60. Sahrmann, S. *Movement System Impairment Syndromes of the Extremities, Cervical and Thoracic Spines.* (Elsevier/Mosby, 2011).
61. Wagner, T. et al. Strengthening and neuromuscular reeducation of the gluteus Maximus in a triathlete with Exercise-Associated cramping of the hamstrings. *J. Orthop. Sport Phys. Ther.* **40** (2), 112–119 (2010).
62. Padua, D. A., Bell, D. R. & Clark, M. A. Neuromuscular characteristics of individuals displaying excessive medial knee displacement. *J. Athl Train.* **47** (5), 525–536 (2012).
63. Gibbons, S. G. T., Comerford, M. J. & Emerson, P. L. Rehabilitation of the Stability Function of Psoas Major. Orthop Div Rev. (January/February 2002):9–16. Available from: www.orthodiv.org (2002).
64. Park, R. J., Tsao, H., Claus, A., Cresswell, A. G. & Hodges, M. P. W. Changes in regional activity of the Psoas major and quadratus lumborum with voluntary trunk and hip tasks and different spinal curvatures in sitting. ;**43**(2):74–82. (2013).
65. Neumann, D. A. Kinesiology of the hip: A focus on muscular actions. *J. Orthop. Sport Phys. Ther.* **40** (2), 82–94 (2010).
66. Nekoie, F. K., Mohammadi, H. K., Safavi, A. A., Jalali, H. M. & Taheri, N. Assessment of hip range of motion limitations in cases with low back pain based on the classified movement. *Syst. Impairment* ;1–5. (2023).
67. Willson, J. D. & Davis, I. S. Utility of the Frontal Plane Projection Angle in Females With Patellofemoral Pain. J Orthop Sport Phys Ther. 38(10):606–15. https://doi.org/10.2519/jospt.2008.2706 (2008).
68. Goldberg, E. J. & Neptune, R. R. Compensatory strategies during normal walking in response to muscle weakness and increased hip joint stiffness. *Gait Posture*. **25** (3), 360–367 (2007).
69. Goldberg, S. R., Anderson, F. C., Pandy, M. G. & Delp, S. L. Muscles that influence knee flexion velocity in double support: implications for stiff-knee gait. *J. Biomech.* **37** (8), 1189–1196 (2004).

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Author contributions
All authors contributed to the concept and design, data acquisition, analysis and interpretation, drafting and revision, final approval and agreement to be accountable. The corresponding authors attest that all listed authors meet authorship criteria and that no others meeting the criteria have been omitted.

Declarations

Competing interests
The authors declare no competing interests.

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