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Kinematics and muscle forces in women with patellofemoral pain during the propulsion phase of the single leg triple hop test

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ABSTRACT

Background: Approximately 25% of orthopedic knee conditions are related to patellofemoral pain (PFP), with young women being the most affected. It is thought that this condition is associated with modifications in the kinematics and muscle control patterns of the lower limb during weight-bearing support activities, which increases femur movement under the patella.

Objectives: To compare kinematics and muscle induced acceleration patterns between PFP subjects and healthy controls during the preparation phase of the single leg triple hop test.

Study design: Biomechanical analysis was performed using OpenSim. Ten physically active women (23.2 ± 4 years, 59.3 ± 5.8 kg, and 1.63 ± 0.06 m) with no history of lower limb injury (CG) and 11 volunteers (23.5 ± 2 years, 55.4 ± 4.9 kg, and 1.66 ± 0.04 m) with PFP (PFPG) were recruited. The participants performed a series of single leg triple hop tests while the ground reaction forces and kinematic data were recorded.

Results: Vector field statistical analysis indicated increased lumbar extension, anterior pelvic tilt, contralateral pelvic drop, and lower induced accelerations from the core and hip muscles in PFPG.

Conclusion: PFP volunteers presented with alterations in lumbar muscle control associated with a possible compensatory pelvic strategy to minimize knee extensor moment.

1. Introduction

Patellofemoral pain (PFP) is characterized by pain around the peripatellar region of the knee joint. Usually, PEP gradually progresses as a diffuse pain at the anterior knee [1], which increases during squatting, running, stair climbing, and long periods of sitting. Approximately 25% of orthopedic knee conditions are related to PFP, and women aged between 16 and 25 years are the most affected [2].

The first attempts to explain the pathomechanics of patellofemoral pain were based upon quadriceps dysfunction. In this condition, unbalanced forces can shift the patella laterally, increasing the stress between the patellar articular cartilage and the lateral femoral condyle. However, not a single cause, but a combination of kinematical, muscular, and dynamical (joint torques) alterations of lumbar, pelvis, hip [3–5], knee [6], or ankle joint changes could be combined, as a cause or compensation to reduce pain during a stabilization task, with patellofemoral pain. For example, Liao et al. [7], in the investigation of

patellofemoral joint stress in females, showed the presence of higher compression forces when the femur was internally rotated by 5° during bilateral squats.

The literature suggests a relationship between muscle weakness and changes in muscle recruitment patterns and postural sway in PFP subjects [8–10]. However, Thijs et al. [11] found that the lack of isometric hip strength was not related to the development of PFP during running. Dos Reis et al. [12] advise that the single leg triple hop test (SLTHT) could be well suited to evaluate the mechanics of the knee joint because of the higher loads imposed on the joint during the task, thus evidencing changes in body stabilization task control associated with the joint condition.

The association of musculoskeletal models, inverse dynamics, and optimization techniques allow muscle force and activation estimations during a functional task, providing insight into muscle synergies [13]. Joint motion is not only controlled by the muscle crossing the joint, but muscles that span different joints can accelerate other joints [14].

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However, we found no previous research based on musculoskeletal models to estimate the muscle forces during a jumping task in subjects with patellofemoral pain.

This study aimed to compare the kinematics and muscle induced acceleration patterns between PFP subjects and healthy controls. Induced acceleration analysis (IAA) quantifies how much the force acting on the model (inertial, reaction, or active, e.g., muscles) contributes to the acceleration of a generalized coordinate. The entire analysis was carried out on the preparation phase of the SLTHT, using OpenSim tools [13]. Both the preparation and landing phases of SLTHT showed kinematic changes in patients with patellofemoral pain syndrome. However, the preparation phase generates sufficiently high joint powers, such that kinematic changes are likely to emerge in the subjects with patellofemoral pain. In fact, during the propulsion phase of the test, the PFP group presented with higher ipsilateral trunk lean, trunk flexion, contralateral pelvic drop, and higher internal rotation, as observed in a previous study by our group [15]. We hypothesized that patients with patellofemoral pain, compared to asymptomatic subjects, present differences in trunk, pelvis, and lower limb joint kinematics during the propulsion phase of the SLTHT. Such differences can be associated with the interaction of the muscle control synergies with rigid body dynamics, which is revealed by the IAA technique.

2. Methods

2.1. Sample and experimental protocol

Eleven women (23.5 ± 2 years, 55.4 ± 4.9 kg, and 1.66 ± 0.04 m) with PFP (PFPG) and ten physically active women (23.2 ± 4 years, 59.3 ± 5.8 kg, and 1.63 ± 0.06 m) with no history of lower limb injury (CG), matched by age, weight, and height, were recruited to participate in the study. Women with PFP were assessed by a physical therapist with 10 years' experience and were included in the PFPG if they had anterior knee pain (VAS: 4.9 ± 1.6 at the time of assessment) for at least 3 months. They should have also reported an increase in the anterior knee pain for at least two activities, such as ascending or descending stairs, squatting, kneeling, jumping, long periods of sitting, and resisted isometric knee extension at 60° of knee flexion. If volunteers presented with bilateral PFP, self-reported dominant lower limb was assessed. All volunteers reviewed the study protocol and provided written informed consent. The study was approved by Universidade Nove de Julho Ethics Committee under the protocol number 15426/2012.

The methodology used in this study has been partially addressed previously [15,16]. Briefly, the volunteers performed a series of SLTHTs using the affected leg in a force plate (model 9286A, Kistler group, Winterthur, Switzerland) at a 400-Hz sampling frequency. Because the landing phase was not being measured in this study, such a frequency was sufficiently high [17,18]. The coordinates of the reflective markers, positioned according to the Vicon Plug-in Gait model [19], were measured using eight infrared cameras (SMART-D BTS, Milan, Italy), and sampled at 100 Hz. Data were filtered using a fourth-order, zero-lag, Butterworth, 12 Hz low-pass filter. The subjects initially stood still on a force platform, without moving the arms and with the hands crossed on the chest. They performed three consecutive single-leg hops, aiming to reach the farthest possible distance each time. A dataset from one trial for each subject was selected for data processing based on a visual inspection of the best data integrity.

This paper addresses the propulsion phase of SLTHT; the initial propulsion phase begins with the volunteer dorsiflexing the supporting leg ankle and flexing the knee and hip, descending the center of mass. By the middle of the propulsion phase, the knee reaches its maximum flexion angle and the movement reverses. Before the end of the propulsion phase, the muscles deliver the highest forces, expressed by the ground reaction forces (Fig. 1), projecting the body upwards and forwards, until the foot loses contact with the platform. During the initial

propulsion phase, the vertical ground reaction force (GRF) initially decreases until the maximum knee flexion, when the vertical reaction force reaches approximately the same value as the body weight; this moment is defined as the peak knee flexion angle. The leg then extends vigorously until the foot leaves the ground [15]. The end of the initial propulsion was assumed when the GRF reached 90% of the body weight after beginning the task, and the end propulsion was assigned when the GRF reached 10% of the body weight [15].

The experimental data were imported into OpenSim using the authors' C3D2OpenSim MatLab® routines, available at <https://simtk.org/home/c3d2opensim> [16]. A 21 rigid-bodies, 92 muscles, and 39 generalized coordinates model of the whole human body [16] had the knee joint file edited according to OpenSim online documentation (<https://simtk-confluence.stanford.edu:8443/display/OpenSim/Model+Editing>) in order to allow adduction/abduction movement since previous studies suggest that dynamic valgus may occur during patellofemoral pain patients [20]. A degree of freedom was added, by editing *model properties* attributing rotation3 (about x-axis) the value of 1 instead of 0 and naming it "knee_adduction_r(1)" in both sides of the model. The segment lengths and the center of mass position were led for each subject. Inverse kinematics was used to determine the generalized coordinates [16] and run the residual reduction algorithm (RRA) for dynamic consistency [21]. Residual actuators were bound to 20 N for force and 100 Nm for torque, by applying RRA iteratively. The iterative process was interrupted when the residual were smaller than the bounds, the average RRA decrease between two iterations was less than 5%, or when 15 iterations was reached [16]. The scaled and RRA adjusted model was utilized for estimating joint torques and muscle forces using OpenSim inverse dynamics and static optimization tools, and the sum of the squared activations was minimized. In the control group, we demonstrated that the association of the RRA and static optimization outperforms the computed muscle control (CMC) compared to experimental EMG profiles [16].

To address whether the muscle forces in the PFPG interact with rigid body motion differently from controls, induced acceleration analysis (IAA) was performed [22], as implemented in OpenSim 3.3 (<https://simtk-confluence.stanford.edu/display/OpenSim/How+IAA+Works>). Induced acceleration analysis allows us to infer how much the force exerted by a single muscle, or other forces, accelerates a generalized coordinate (i.e., a joint angle). The joint angles and muscles that presented statistically different patterns between the two groups were selected for the IAA.

2.2. Statistical analysis

With the aim to avoid *post hoc* regional focus biases or inter-component covariance bias [23], and to identify the movement phases with significant variations between the PFPG and CG, a vector-field statistical analysis [24], for both kinematics and muscle induced accelerations, was performed. For the statistical parametric mapping (SPM) figures that are shown in the Results section, the continuous black line below each comparison graphic shows *post hoc* scalar field *t*-tests. As this line crosses the dashed line either above or below, a statistical difference was found between groups. First, we compared joint kinematics between the CG and the PFPG. When a difference was observed, muscle induced accelerations were investigated; this approach reduces the amount of data to be analyzed. An alpha of 10% was considered for kinematics and 5% for muscle induced acceleration statistical tests. In this work, kinematics was used for selecting generalized coordinates candidates for the induced acceleration analysis. Thus, we decided using an unusually large alpha value for increasing the number of coordinates to be further investigated.

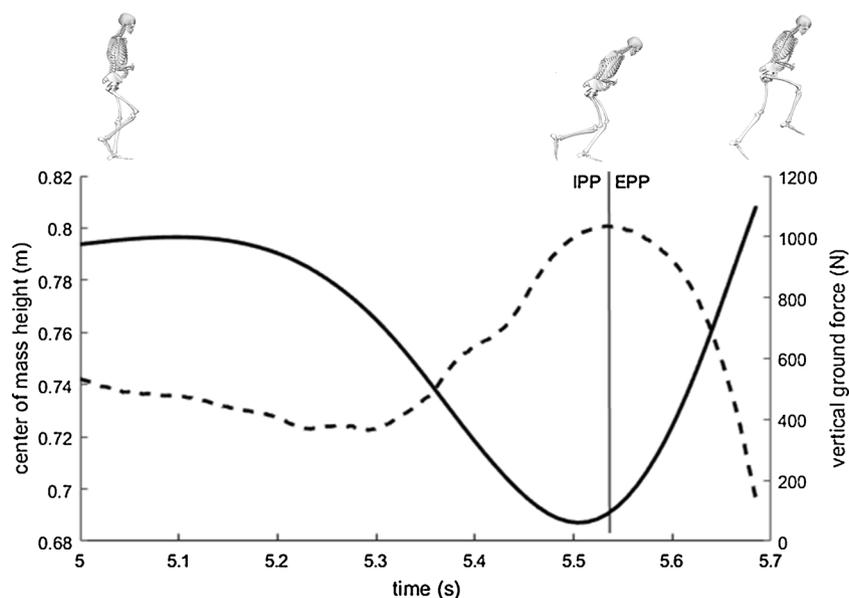


Fig. 1. Time course of vertical displacement of center of mass (solid) and vertical ground reaction force during preparation phase of single leg triple hop test. Vertical line represents the transition from initial propulsion phase to end of propulsion phase.

3. Results

3.1. Kinematics

Kinematics and statistical comparison vectors between the PFPG and the CG are shown in Fig. 2. Statistical differences between normal and patellofemoral pain subjects were found for lumbar extension (Fig. 2A), pelvic list (Fig. 2D), and pelvic tilt (Fig. 2E). This analysis highlights the changes in the proximal joint kinematics in the PFPG compared to the CG. The lumbar extension angle remained negative (spine flexed) and stable throughout the task in the PFPG. The pelvic tilt showed a difference between the two groups around 55 to 70% of the task total time when the volunteers are in the lowest height of its center of mass. Patellofemoral pain group performed an anterior pelvic tilt, while the controls remained with their pelvis closer to a neutral or posterior tilt. The pelvic list showed a difference from 58% to 64% of the task time, with PFPG presenting a pelvic drop when lowering their center of mass.

3.2. Lumbar extension induced acceleration analysis

Fig. 3 shows the induced accelerations of selected muscles to lumbar extension, which presented with statistical differences between the PFPG and the CG. Excessive lumbar flexion during the entire task in the PFPG compared to the CG may be explained by reduced activity in the lumbar extensors (shown as positive induced accelerations) erector spinae, from both sides (Fig. 2A, 10%–15%, 75% and 85% of the task time, and Fig. 2B, 70%–90% of the task time). Additionally, there was an increased activity of lumbar flexors (negative induced accelerations), namely external oblique contralateral to the jumping side (Fig. 3C, from start to 25%, and from 75% to the end) and internal obliques (Fig. 2D and E, 10% and 20%). Other muscles not crossing the lumbar joint, such as the rectus femoris (Fig. 3I, from start to 95%) and the psoas (Fig. 3H, 5%–50%) accelerate the trunk toward flexion. Contrarily, lines one and two of the gluteus maximus (Fig. 3F and G, 5% and after 80%) accelerate toward extension, but with different temporal patterns between the two groups.

3.3. Pelvic tilt induced acceleration analysis

Fig. 4 shows the accelerations induced by selected muscles to pelvic

tilt, which demonstrated statistical differences between the PFPG and the CG. In 55%–65% of the task time, the pelvis tilted anteriorly in the PFPG (Fig. 1D). Such differences in the PFPG can be related to smaller contributions to the posterior tilt of some muscles (positive induced accelerations), such as the external oblique from both sides (Fig. 4B and C, 45%–55% of the task), line one of the adductor magnus (Fig. 4G, from 55% to 65%), and the quadratus femoris (Fig. 4H, 40%–60%).

Some muscles, such as the iliopsoas (Figs. 4G and 3 H, 55%), showed less substantial contributions to the posterior pelvic tilt in subjects with patellofemoral pain compared to the CG. In contrast, the semitendinosus (Fig. 4I, 60%–75%) and the long head of the biceps femoris (Fig. 4K, 15%–65%) contributed predominantly to the anterior pelvic tilt (negative differences).

3.4. Pelvic list induced acceleration analysis

Selected muscles which demonstrated statistical differences of pelvic list induced accelerations between the two groups are shown in Fig. 5. The negative pelvic list corresponds to a pelvic drop of the opposite side of the jumping limb at 55%–65% of the task time, as shown in Fig. 1E. Negative induced accelerations can be observed from the internal oblique contralateral to the jumping side (Fig. 5A, from 20% to 60%), psoas (Fig. 5D, 70% of the task), quadratus femoris (Fig. 5E, 60%), gracilis (Fig. 5F, 60%–80%), lines one and two of the adductor magnus (Fig. 5G and H, from 55% to 65%), and adductor longus (Fig. 5F, until 60%).

By contrast, in the PFPG, some muscles had a tendency to raise the contralateral pelvis. The most remarkable were lines one and three of the gluteus maximus (Fig. 5B, from the start up to 80% and 5C, up to 50%), the semitendinosus (Fig. 5J), and the semimembranosus (Fig. 5K), until 30% and 70% of the task, respectively.

4. Discussion

The objective of this study was to investigate the differences in kinematic patterns and muscle action during the preparation phase of the SLTHT between subjects with patellofemoral pain syndrome and healthy controls. The joint angles were calculated using inverse kinematics and muscle actions inferred from the muscle induced accelerations, as estimated by OpenSim. Decreased lumbar extension, anterior pelvic tilt, and contralateral pelvic drop were observed; these changes

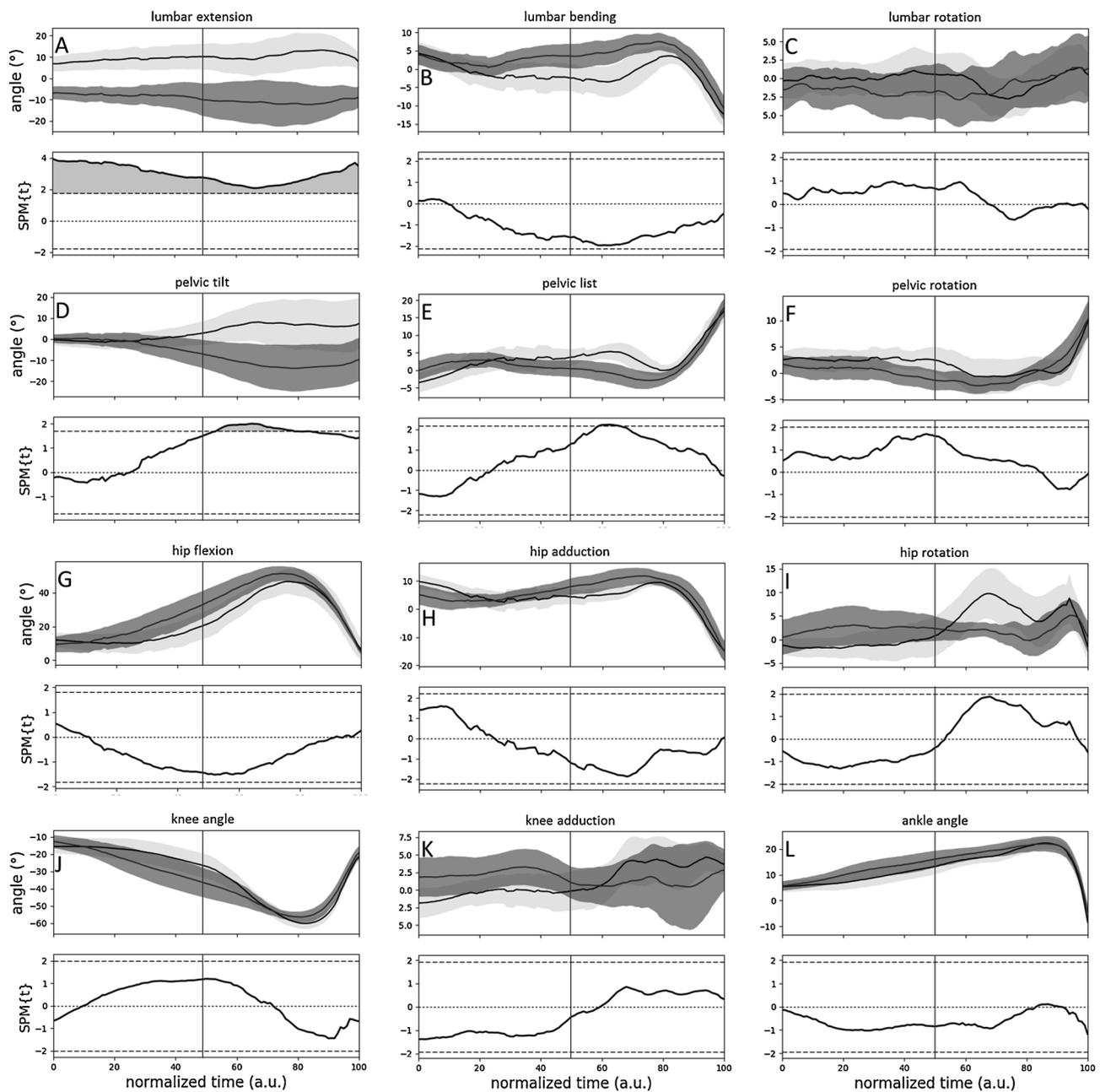


Fig. 2. Vector field kinematics analysis of joint angles from PFPG (black) and CG (gray) groups during the preparation phase of SLTHT. The subplots are: (A) lumbar extension (+ extension / - flexion), (B) lumbar bending (+ contralateral to jumping limb / - ipsilateral to jumping limb), (C) lumbar rotation (+ contralateral to jumping limb / - ipsilateral to jumping limb), (D) pelvic tilt (+ posterior / - anterior), (E) pelvic list (+ drop the pelvis in the side of the jumping limb / - drop the pelvis in the opposite side of the jumping limb), (F) pelvic rotation (+ anterior rotation of the pelvis in the opposite side of the jumping limb / - posterior rotation of the pelvis in the opposite side of the jumping limb), (G) hip flexion (+ flexion / - extension), (H) hip adduction (+ adduction / - abduction), (I) hip rotation (+ internal rotation / - external rotation), (J) knee extension (+ extension / - flexion), (K) knee adduction (+ adduction / - abduction) and (L) ankle dorsiflexion (+ dorsiflexion / - plantarflexion). The black vertical line represents the peak knee flexion angle. SPM (Statistical Parametric Mapping) continuous black line below each comparison graphic shows *post hoc* scalar field *t*-tests. As this line crosses the dashed line above or below, a statistical difference is found between groups.

confirm previous findings regarding motor control alterations in the proximal joints [9,10] associated with PFP. The group with patellofemoral pain showed adaptations toward lumbar flexion for the core muscles (lumbar extensors, lumbar benders, and lumbar rotators), gluteus maximus, psoas, rectus femoris, and the hamstrings. The external obliques, adductor magnus, and quadratus femoris showed lower contributions to posterior pelvic tilt in the PFPG compared to the CG. Some of these findings are in accordance with previous data reported in the literature [8–10,25] which relate patellofemoral pain syndrome to uncontrolled trunk and pelvic movement. We have also shown that groups of muscles produce opposite contributions to pelvic movement

in the frontal plane (pelvic list). Some muscles had a tendency to increase the contralateral pelvic drop observed in the PFPG, while others induce accelerations to raise it. This finding may be associated with changes in the hip muscles moment arms resulting from the postural adjustment and will be discussed later.

Modeling techniques used to estimate muscle force produce a large set of data, with many muscles working throughout the entire span of the movement. To avoid parameterization of data and possible loss of information, we performed a statistical test taking into account the whole kinematics and induced acceleration waveforms. Muscle induced accelerations showed statistical differences and highly variable

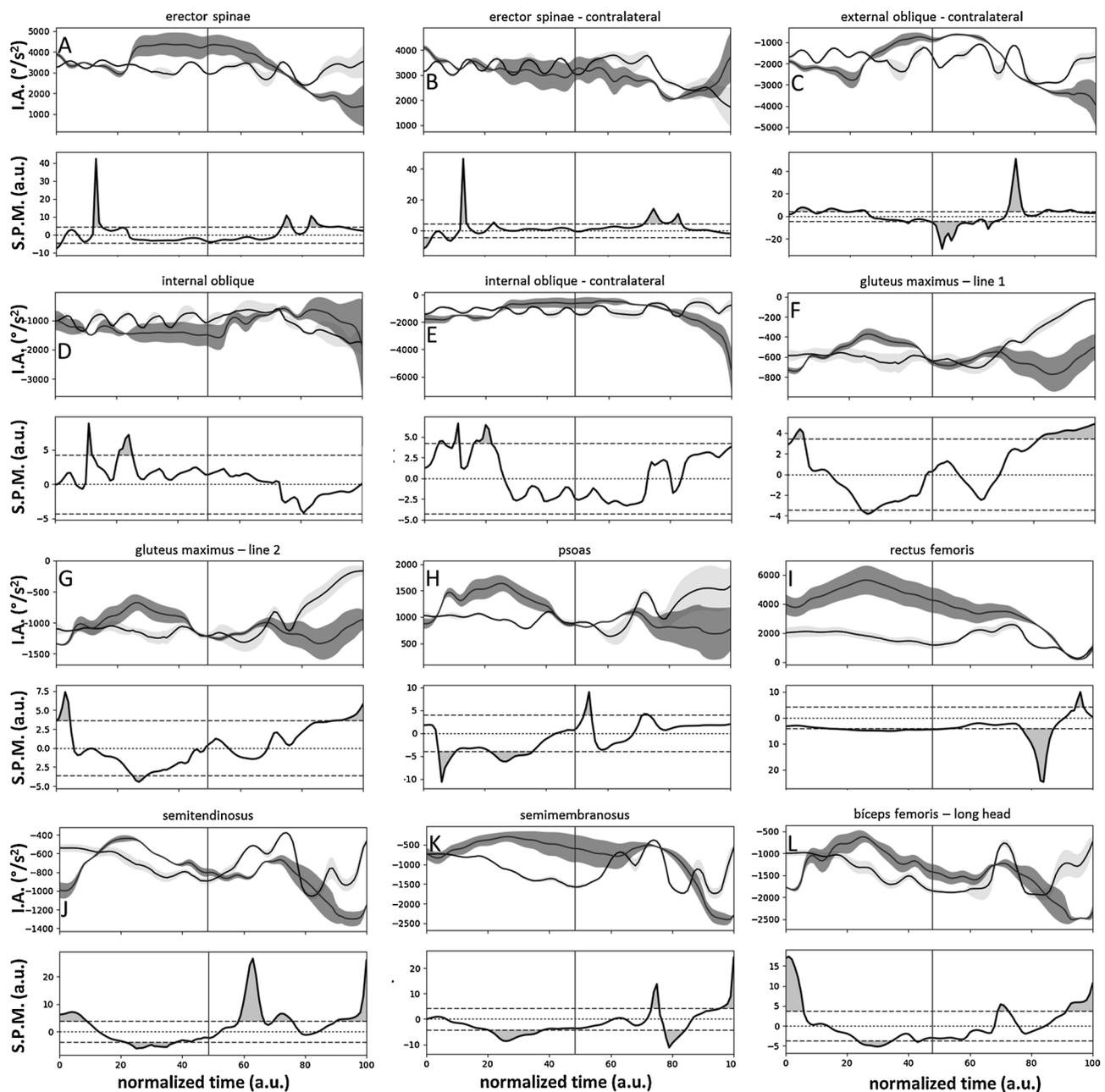


Fig. 3. Muscle induced accelerations contributions ($^{\circ}/s^2$) to lumbar extension from PFP (black) and CG (gray) during the preparation phase of SLTHT. The subplots are: (A) erector spinae, (B) erector spinae contralateral to the jumping side, (C) external oblique contralateral to the jumping side, (D) internal oblique, (E) internal oblique contralateral to the jumping side, (F) line one of gluteus maximus, (G) line two of gluteus maximus, (H) psoas, (I) rectus femoris, (J) semitendinosus, (K) semimembranosus and (L) long head of biceps femoris. Positive values indicate muscle contributions to lumbar extension and negative values shows muscle contribution to the opposite movement. The black vertical line represents the peak knee flexion angle.

patterns, even when no differences in kinematics were observed. Such analysis may highlight different patterns of muscle synergies to accomplish the same task, embedded in the kinematics and ground reaction forces data. Thus, we selected muscles with larger potentials to induce joint accelerations and focused the analysis on the periods with significant kinematic differences between the groups.

Subjects with patellofemoral pain performed lumbar flexion instead of lumbar extension during the task (Fig. 2A). The induced muscle accelerations in this coordinate pointed toward a reduced contribution to lumbar extension, or an increased contribution to lumbar flexion, of several muscles, including back extensors, lumbar benders and rotators, gluteus maximus, psoas, rectus femoris, and hamstrings (Fig. 4). Yelvar [8] found that PFP patients demonstrated a lower endurance of core muscles than age-matched pain-free controls. Furthermore,

Biabanimoghdam et al. [9] and Rojhani Shirazi et al. [10] showed a delay in core muscle actuation during sudden movement in PFP patients. In addition, Almeida et al. [25] related the lower peak torque of lateral core stabilizers, hip abductors, and extensors in women with patellofemoral pain. However, Thijs et al. [11] found no significant relationship between the core or lower limb isometric weakness and patellofemoral pain.

Core muscles stabilize the trunk and their deficient motor control may disrupt lumbar and pelvic stability, causing inappropriate force transmission to distal segments of the body [26]. Here, excessive anterior pelvic tilt positioning and a small drop were observed at approximately 60% of the task time. Simultaneously, some muscles showed lower sagittal pelvic and lumbar extension induced accelerations compared to controls; these included the external oblique

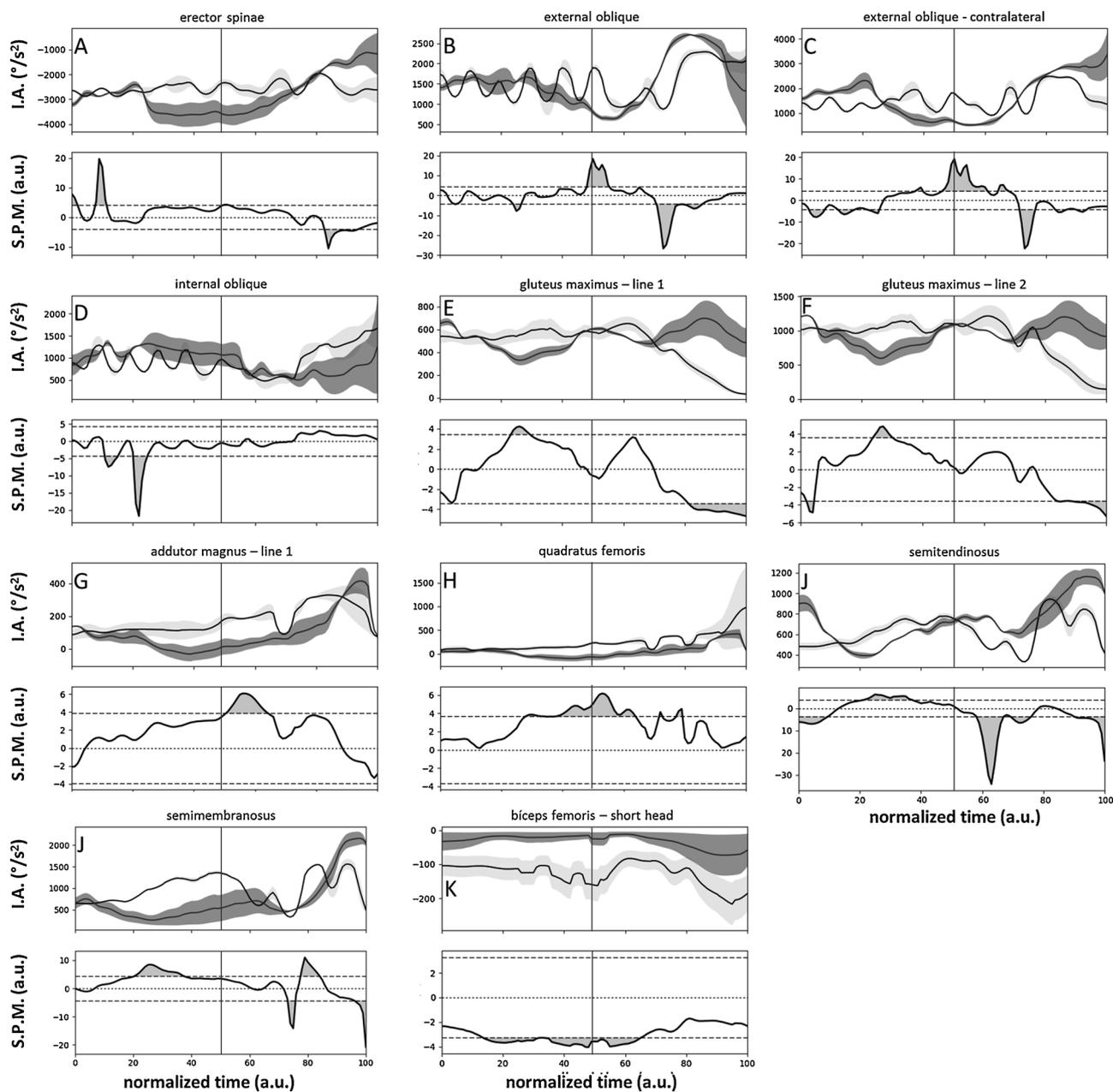


Fig. 4. Muscle contributions ($^{\circ}/s^2$) to pelvic tilt from PFPG (black) and CG (gray) during the preparation phase of SLTHT. The subplots are: (A) erector spinae, (B) external oblique, (C) external oblique contralateral to the jumping side, (D) internal oblique, (E) line one of gluteus maximus, (F) line two of gluteus maximus, (G) line one of adductor magnus, (H) quadratus femoris, (I) semitendinosus, (J) semimembranosus and (K) long head of biceps femoris. Positive values indicate muscle contributions to posterior pelvic tilt and negative values shows muscle contribution to the opposite movement. The black vertical line represents the peak knee flexion angle.

contralateral to the jumping side, the adductor magnus, and the quadratus femoris.

The kinematic changes observed in the PFPG may predispose to a lateral shift of the patella on the patellar groove. However, no significant single joint angle differences were observed between the two groups. As shown in Fig. 2(G–I), the SPM differences, in some instances, reach close to the significance threshold for hip flexion, adduction, and rotation. We hypothesized that, combined with the ankle and knee, such small differences could compensate for the contralateral pelvic drop. In this case, the relative position between the femur and the patella and the direction of the resulting force on the patella might be altered. Baldon et al. [27] found milder patellofemoral pain symptoms in subjects who underwent functional stabilization training involving the lumbar spine, pelvis, and lower limb, compared to subjects who

performed only lower limb strengthening.

Excessive trunk flexion and anterior pelvic tilt may represent an attempt to displace the body center of mass anteriorly and reduce the knee extension demands. This adaptation may explain some of the differences in the accelerations induced by the muscles that control pelvic list, i.e., hip adductors and abductors. Such differences may be associated with the pelvis and trunk positions. As the patellofemoral pain subjects tilt the pelvis anteriorly, the hip adduction moment arms of some muscles, such as the adductor longus and psoas, increase significantly, which would favor pelvic drop in the frontal plane. Furthermore, the gluteus maximus hip abduction moment arm also increases. Such changes associated with muscle control impairments, may go some way to explain the increased pelvic drop observed in the PFPG.

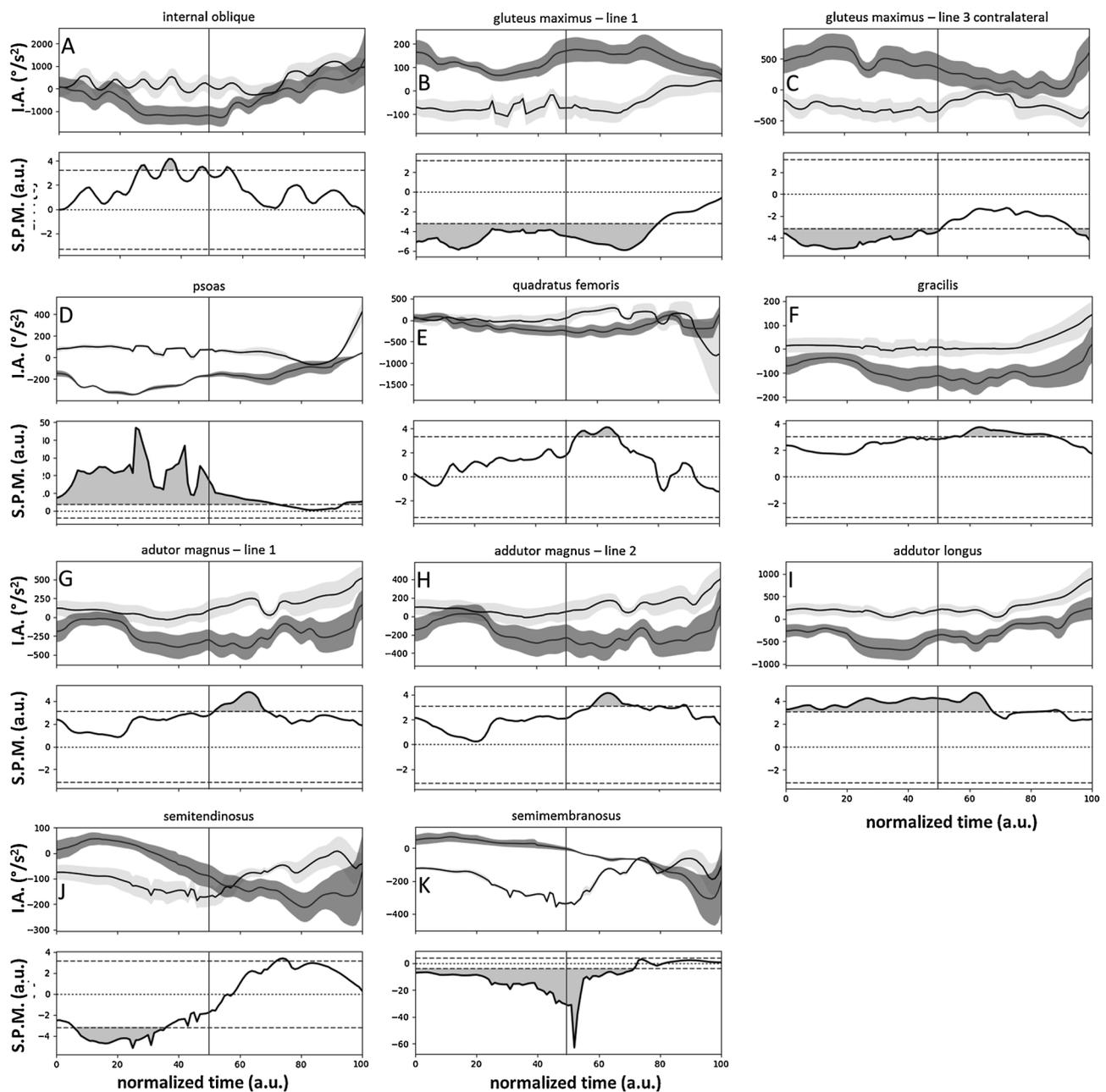


Fig. 5. Muscle contributions ($^{\circ}/s^2$) to pelvic list from PFPG (black) and CG (gray) during the preparation phase of SLTHT. The subplots are: (A) internal oblique, (B) line one of gluteus maximus, (C) line three of gluteus maximus, (D) psoas, (E) quadratus femoris, (F) gracilis, (G) line one of adductor magnus, (H) line two of adductor magnus, (I) adductor longus, (J) semitendinosus and (K) semimembranosus. Positive values indicate muscle contributions to opposite pelvis rising, and negative to drop. The black vertical line represents the peak knee flexion angle.

This study has some methodological limitations related to both data acquisition and the modeling process. For instance, the Plug in Gait model considers the foot as a single rigid body, hindering the kinematic analysis of the subtalar joint. In our model, this joint was locked in a neutral position, which may have impaired the muscle force estimations of the invertor and evertor muscles. Tiberio [28] reported that excessive foot pronation might lead to compensatory gait changes resulting in PFP syndrome. Here, the lack of ankle inversion/eversion angle precludes such observation. The lower limb model described by Hamner et al. [20] is formulated upon both anthropometric and muscle data from several sources. The maximum isometric muscle force (F_{om}) is a sensitive parameter in Hill-type muscle models [29]. Correa and Pandy [30] suggested a mass scaling factor for adjusting F_{om} to children. However, we found no specific technique in the literature to scale F_{om}

for young women, and as a result, the standard OpenSim scaling tool was used. The Hamner et al. model has been previously used for running; here, the observed range of motion for the lumbar joint exceeds that presented in [28] by approximately 15° . Furthermore, only the erector spinae, internal oblique, and external oblique torso muscles are present in the model. Although these are the most relevant, other back and abdominal muscles have not been considered with regards to which IAAs would yield significant differences between the groups. The whole lumbar motion occurs by the lumbosacral joint modeled as a ball-and-socket; thus, small movements between the individual vertebrae are not taken into account. Another limitation of this study is the relative inflexibility of the objective function choice in the OpenSim static optimization formulation, which was selected as the sum of the squared activations. Finally, the highly variable patterns from IAA and

statistical parametric mappings may be caused by an eventually high sensitivity of this variable to model parameters, by either modeling errors or by an improper application of IAA to a non-repetitive movement.

5. Conclusions

The kinematic analysis showed altered trunk and pelvic control in the PFG, characterized by increased lumbar extension, anterior pelvic tilt, and a small pelvic drop. Women with patellofemoral pain syndrome showed lumbar control alterations in the sagittal plane, as evidenced by lower muscle induced accelerations from several core and hip muscles. It is likely that this is a compensatory strategy to minimize knee extensor moment and reduce patellofemoral compressive force. The computational biomechanical techniques used in this study could be extended to include a broader set of functional tasks in order to better understand knee pain pathomechanics, and to help rehabilitation professionals to improve their treatment plan and prognosis.

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References

- [1] H. Collado, M. Fredericson, Patellofemoral pain syndrome, *Clin. Sports Med.* 29 (2010) 379–398, <https://doi.org/10.1016/j.csm.2010.03.012>.
- [2] P. Baquie, P. Brukner, Injuries presenting to an Australian sports medicine centre: a 12-month study, *Clin. J. Sport Med.* 7 (1997) 28–31, <https://doi.org/10.1097/00042752-199701000-00006>.
- [3] M.L. Ireland, J.D. Willson, B.T. Ballantyne, I.M. Davis, Hip strength in females with and without patellofemoral pain, *J. Orthop. Sport. Phys. Ther.* 33 (2003) 671–676, <https://doi.org/10.2519/jospt.2003.33.11.671>.
- [4] R. Ferber, I.M.C. Davis, D.S. Williams, Gender differences in lower extremity mechanics during running, *Clin. Biomech.* 18 (2003) 350–357, [https://doi.org/10.1016/S0268-0033\(03\)00025-1](https://doi.org/10.1016/S0268-0033(03)00025-1).
- [5] C.D. Pollard, I.M.C. Davis, J. Hamill, Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver, *Clin. Biomech.* 19 (2004) 1022–1031, <https://doi.org/10.1016/j.clinbiomech.2004.07.007>.
- [6] T.W. Kernozek, M.R. Torry, H. Van Hoof, H. Cowley, S. Tanner, Gender differences in frontal and sagittal plane biomechanics during drop landings, *Med. Sci. Sports Exerc.* 37 (2005) 1003–1012, <https://doi.org/10.1249/01.mss.0000171616.14640.2b>.
- [7] T.C. Liao, N. Yang, K.Y. Ho, S. Farrokhi, C.M. Powers, Femur rotation increases Patella cartilage stress in females with patellofemoral pain, *Med. Sci. Sports Exerc.* 47 (2015) 1775–1780, <https://doi.org/10.1249/MSS.0000000000000617>.
- [8] G.D.Y. Yelvar, Y. Çirak, M. Dalkılıç, Y.P. Demir, G. Baltacı, M. Kömürçü, Impairments of postural stability, core endurance, fall index and functional mobility skills in patients with patello femoral pain syndrome, *J. Back Musculoskelet. Rehabil.* 30 (2017) 163–170, <https://doi.org/10.3233/BMR-160729>.
- [9] M. Biabanimoghdam, A. Motealleh, S.M. Cowan, Core muscle recruitment pattern during voluntary heel raises is different between patients with patellofemoral pain and healthy individuals, *Knee* 23 (2016) 382–386, <https://doi.org/10.1016/j.knee.2016.01.008>.
- [10] Z. Rojhani Shirazi, M. Biabani Moghaddam, A. Motealleh, Comparative evaluation of core muscle recruitment pattern in response to sudden external perturbations in patients with patellofemoral pain syndrome and healthy subjects, *Arch. Phys. Med. Rehabil.* 95 (2014) 1383–1389, <https://doi.org/10.1016/j.apmr.2014.01.025>.
- [11] Y. Thijs, E. Pattyn, D. Van Tiggelen, L. Rombaut, E. Witvrouw, Is hip muscle weakness a predisposing factor for patellofemoral pain in female novice runners? A prospective study, *Am. J. Sports Med.* 39 (2011) 1877–1882, <https://doi.org/10.1177/0363546511407617>.
- [12] A.C. dos Reis, J.C.F. Correa, A.S. Bley, N.D. dos, A. Rabelo, T.Y. Fukuda, P.R.G. Lucareli, Kinematic and kinetic analysis of the single-leg triple hop test in women with and without patellofemoral pain, *J. Orthop. Sport. Phys. Ther.* 45 (2015) 799–807, <https://doi.org/10.2519/jospt.2015.5011>.
- [13] S.L. Delp, F.C. Anderson, A.S. Arnold, P. Loan, A. Habib, C.T. John, E. Guendelman, D.G. Thelen, OpenSim: open-source software to create and analyze dynamic simulations of movement, *IEEE Trans. Biomed. Eng.* 54 (2007) 1940–1950, <https://doi.org/10.1109/TBME.2007.901024>.
- [14] R.R. Neptune, S.A. Kautz, F.E. Zajac, Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking, *J. Biomech.* 34 (2001) 1387–1398, [https://doi.org/10.1016/S0021-9290\(01\)00105-1](https://doi.org/10.1016/S0021-9290(01)00105-1).
- [15] A.S. Bley, J.C.F. Correa, A.C. Dos Reis, N.D.D.A. Rabelo, P.H. Marchetti, P.R.G. Lucareli, Propulsion Phase of the single leg triple hop test in women with patellofemoral pain syndrome: a biomechanical study, *PLoS One* 9 (2014) e97606, <https://doi.org/10.1371/journal.pone.0097606>.
- [16] F.C. Alvim, P.R.G. Lucareli, L.L. Mengaldo, Predicting muscle forces during the propulsion phase of single leg triple hop test, *Gait Posture* 59 (2018) 298–303, <https://doi.org/10.1016/j.gaitpost.2017.07.038>.
- [17] K. Mackala, J. Stodółka, A. Siemiński, M. Coh, Biomechanical analysis of squat jump and countermovement jump from varying starting positions, *J. Strength Cond. Res.* 27 (2013) 2650–2661, <https://doi.org/10.1519/JSC.0b013e31828909ec>.
- [18] A. Struzik, J. Zawadzki, Leg stiffness during phases of countermovement and take-off in vertical jump, *Acta Bioeng. Biomech.* 15 (2013) 113–118 (Accessed July 9, 2019), <http://www.ncbi.nlm.nih.gov/pubmed/23952196>.
- [19] R.B. Davis, S. Öunpuu, D. Tyburski, J.R. Gage, A gait analysis data collection and reduction technique, *Hum. Mov. Sci.* 10 (1991) 575–587, [https://doi.org/10.1016/0167-9457\(91\)90046-Z](https://doi.org/10.1016/0167-9457(91)90046-Z).
- [20] S.R. Hamner, A. Seth, S.L. Delp, Muscle contributions to propulsion and support during running, *J. Biomech.* 43 (2010) 2709–2716, <https://doi.org/10.1016/j.jbiomech.2010.06.025>.
- [21] D.G. Thelen, F.C. Anderson, Using computed muscle control to generate forward dynamic simulations of human walking from experimental data, *J. Biomech.* 39 (2006) 1107–1115, <https://doi.org/10.1016/j.jbiomech.2005.02.010>.
- [22] F.E. Zajac, M.E. Gordon, Anonymous, Determining muscle's force and action in multi-articular movement, *Exerc. Sport Sci. Rev.* 17 (1989) 187–230, <https://doi.org/10.1556/AAlim.2015.0002>.
- [23] D. Knudson, Significant and meaningful effects in sports biomechanics research, *Sport. Biomech.* 8 (2009) 96–104, <https://doi.org/10.1080/14763140802629966>.
- [24] T.C. Pataky, M.A. Robinson, J. Vanrenterghem, Vector field statistical analysis of kinematic and force trajectories, *J. Biomech.* 46 (2013) 2394–2401, <https://doi.org/10.1016/j.jbiomech.2013.07.031>.
- [25] G.P.L. Almeida, A.P. de, M.C. Carvalho e Silva, F.J.R. França, M.O. Magalhães, T.N. Burke, A.P. Marques, Does anterior knee pain severity and function relate to the frontal plane projection angle and trunk and hip strength in women with patellofemoral pain? *J. Bodyw. Mov. Ther.* 19 (2015) 558–564, <https://doi.org/10.1016/j.jbmt.2015.01.004>.
- [26] C.E. Rivera, Core and lumbopelvic stabilization in runners, *Phys. Med. Rehabil. Clin. N. Am.* 27 (2016) 319–337, <https://doi.org/10.1016/j.pmr.2015.09.003>.
- [27] R. de, M. Baldon, F.V. Serrão, R. Scattono Silva, S.R. Piva, Effects of functional stabilization training on pain, function, and lower extremity biomechanics in women with patellofemoral pain: a randomized clinical trial, *J. Orthop. Sport. Phys. Ther.* 44 (2014) 240–A8, <https://doi.org/10.2519/jospt.2014.4940>.
- [28] D. Tiberio, Pathomechanics of structural foot deformities, *Phys. Ther.* 68 (1988) 1840–1849, <https://doi.org/10.1093/ptj/68.12.1840>.
- [29] C.Y. Scovil, J.L. Ronsky, Sensitivity of a Hill-based muscle model to perturbations in model parameters, *J. Biomech.* 39 (2006) 2055–2063, <https://doi.org/10.1016/j.jbiomech.2005.06.005>.
- [30] T.A. Correa, M.G. Pandey, A mass-length scaling law for modeling muscle strength in the lower limb, *J. Biomech.* 44 (2011) 2782–2789, <https://doi.org/10.1016/j.jbiomech.2011.08.024>.