Title: Altered Walking and Muscle Patterns Reduce Hip Contact Forces in Individuals with Symptomatic Cam FAI

Running title: Hip Muscle and Contact Forces in Cam FAI

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ABSTRACT

Background: Cam-type femoroacetabular impingement (FAI) is a causative factor for hip pain and early hip osteoarthritis. Although cam FAI can alter hip joint biomechanics, it is unclear what role muscle forces play and how they affect the hip joint loading.

Purpose/Hypothesis: The purpose was to examine the muscle contributions and hip contact forces in individuals with symptomatic cam FAI during level walking. Symptomatic cam FAI demonstrate different muscle and hip contact forces during gait.

Study Design: Controlled laboratory study.

Methods: Eighteen patients with symptomatic cam FAI (n_FAI = 18) were age- and BMI-matched with eighteen control participants (n_CON = 18). Each participant’s walking kinematics and kinetics were recorded throughout a gait cycle (ipsilateral foot-strike to ipsilateral foot-off), using a motion capture system and force plates. Muscle and hip contact forces were subsequently computed using a musculoskeletal modelling program and static optimization methods.

Results: The FAI group walked slower and with shorter steps; demonstrating reduced joint motions and moments during contralateral foot-strike, compared to the CON group. The FAI group showed reduced psoas major (median = 1.1, interquartile range (IQR) = 1.0–1.5 Newton/bodyweight (N/BW)) and iliacus forces (median = 1.2, IQR = 1.0–1.6 N/BW), during contralateral foot-strike, compared to the CON group (median = 1.6, IQR = 1.3–1.6 N/BW, p = 0.004; and median = 1.5, IQR = 1.3–1.6 N/BW, p = 0.03, respectively); which resulted in lower hip contact forces in the anterior (p = 0.026), superior (p = 0.02), and medial directions (p = 0.038). The three vectors produced a resultant peak force at the anterosuperior aspect of the acetabulum for both groups; with the FAI group demonstrating a substantially lower magnitude.
Conclusions: FAI participants altered their walking kinematics and kinetics, especially during contralateral foot-strike as a protective mechanism, and resulted in reduced psoas major and iliacus muscle force and anterosuperior hip contact force estimations.

Clinical Relevance: Limited hip mobility is not only attributed to bone-on-bone impingement, caused by the cam morphology, but could be attributed to musculature as well. Not only would the psoas major and iliacus be able to protect the hip joint during flexion-extension, athletic conditioning could further strengthen core muscles for improved hip mobility and pelvic balance.

Key Terms: femoroacetabular, impingement, muscle forces, motion, joint contact forces
What is known about the subject (only for reviewers, not in word count):

Cam FAI is a common cause for athletic hip injuries and results in early chondrolabral damage.

The bone-on-bone impingement is less likely to occur during level walking, given that the task involves a less demanding range of motion. This suggests that there may be other parameters, in addition to the bony deformity, that could alter walking biomechanics of FAI participants.

What this study adds to existing knowledge (only for reviewers, not in word count):

The reduced psoas major and iliacus muscle forces altered walking biomechanics in the FAI population; which, in turn, decreased the hip contact forces. The understanding of muscle contributions towards joint loading may provide a better strategy for non-surgical management in FAI individual specially.

Keywords: femoroacetabular, impingement, muscle, motion, forces
INTRODUCTION

Cam-type femoroacetabular impingement (FAI) is widely recognized as a causative factor for groin pain, athletic hip injuries, and early adult hip osteoarthritis. The morphology itself appears as an aspherical bony extension at the anterolateral and anterosuperior femoral head on clinical imaging (i.e., elevated alpha angles), leading to a reduced anterior head-neck offset. In addition to the alpha angles that quantify the cam morphology, clinical symptoms and reduced functional range of motion (ROM) have also been attributed to secondary anatomical characteristics.

Individuals with symptomatic cam FAI previously demonstrated altered kinematics and kinetics, compared to control participants, during walking, squatting, and climbing. Specifically during level walking, symptomatic individuals walked with reduced sagittal hip ROM, frontal hip and pelvic ROM, internal hip rotation; as well as with reduced flexion and external rotation moments. Recently, individuals with symptomatic cam FAI also demonstrated weaker muscle strength during maximum isometric voluntary contractions in abduction, adduction, flexion, and external rotation, suggesting that differences during activities of daily living were influenced by altered neuromuscular activation. Although dynamic muscle activity in individuals with FAI has only been investigated during squatting, no significant differences were observed with respect to the control population.

Previous biomechanical studies focused on the analyses of joint kinematics and kinetics associated with FAI, however, there was limited literature on muscle strength or, more importantly, the contributions of muscle forces towards hip joint loading, during an activity of daily living. Therefore, the purpose of this study was to examine the muscle contributions and hip contact forces in individuals with symptomatic cam FAI during level walking. We
hypothesized that symptomatic cam FAI patients demonstrate different muscle and hip contact forces during level walking.

METHODS

Study Design

This controlled laboratory study involved a case-control comparative analysis which compared a patient cohort (case – individuals with symptomatic cam FAI) to a healthy cohort (control – asymptomatic individuals with no cam morphology). All participants provided informed consent and the study was approved by the hospital and institution research ethics boards (OHSN-REB Protocol #2009537-01H).

Participants

A total of thirty-six male participants (n = 36) were included in this study. Eighteen patients (FAI; nFAI = 18) initially presented themselves to the senior orthopaedic surgeon’s clinical practice (PEB) with persisting unilateral clinical signs of hip pain and positive impingement tests; while another eighteen control participants (CON; nCON = 18) were recruited as volunteers from the general population. Each participant underwent pelvic CT imaging (Acquilion, Toshiba Medical Systems Corporation, Otawara, Japan; or Discover CT750, GE Healthcare, Mississauga, ON, Canada), to confirm that the affected, symptomatic hip of each FAI patient had a cam morphology, as defined by an axial 3:00 or radial 1:30 alpha angle larger than 50.5° and 60°, respectively; and to confirm that each CON participant did not indicate any morphology. Alpha angles in the axial 3:00 and radial 1:30 planes were measured, by a senior musculoskeletal radiologist (KR). Participants were excluded if they indicated any other hip morphology (e.g., slipped capital femoral epiphysis, Legg-Calvé-Perthes, dysplasia, overcoverage), body mass


index (BMI) greater than 30 kg/m², or had a history of previous lower limb surgeries, severe traumas, or musculoskeletal disorders. The two groups (FAI and CON) were matched for sex and BMI (Table 1) and the analysis compared the symptomatic hip for each FAI patient and the hip with the lowest (combined axial and radial) α angles for each CON participant (Figure 1A and 1B).

Table 1. Participant demographics and cam morphology parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>FAI</th>
<th>CON</th>
<th>p value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Participants (n)</td>
<td>18</td>
<td>18</td>
<td></td>
</tr>
<tr>
<td>Age (years)</td>
<td>38 ± 9</td>
<td>32 ± 6</td>
<td>0.050</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>176 ± 6</td>
<td>176 ± 6</td>
<td>0.401</td>
</tr>
<tr>
<td>BMI (kg/m²)</td>
<td>27 ± 4</td>
<td>26 ± 3</td>
<td>0.428</td>
</tr>
<tr>
<td>Axial 3:00 α angle (°)</td>
<td>56 ± 9</td>
<td>42 ± 5</td>
<td>0.001</td>
</tr>
<tr>
<td>Radial 1:30 α angle (°)</td>
<td>64 ± 6</td>
<td>51 ± 3</td>
<td>0.001</td>
</tr>
</tbody>
</table>

* Independent samples t-test, significant difference (p < 0.01) compared with CON

Figure 1. A) Patients with symptomatic cam femoroacetabular impingement (FAI), as depicted with an anterosuperior cam deformity, were compared to B) healthy control participants (CON). C) Each participant performed walking trials in the motion capture laboratory, capturing the ipsilateral foot-strike to foot-off kinematics and kinetics; then D) using a musculoskeletal model (consisting of the lower limb and torso segments), the resultant muscle and hip contact forces were determined.
Motion Analysis

Prior to CT imaging, radiopaque surface markers were placed on each participant’s left and right anterior superior iliac spines, left and right posterior superior iliac spines; in order to register the location of the surface markers with respect to the bony landmarks. After imaging, the radiopaque markers were replaced with retro-reflective markers, according to a customized marker set. At the motion capture laboratory, each participant was asked to repeat five level walking trials at a self-selected pace, where kinematics were captured using a ten-infrared camera system (Vicon MX-13, VICON, Oxford, UK; frequency = 200 Hz) and kinetics were captured using two fixed force plates (models FP4060-08, Bertec Corporation, Columbus, OH, USA; frequency = 1000 Hz). The gait analysis was limited to the affected side’s ipsilateral foot-strike to foot-off, as forces at the hip are relevant during the stance phase (Figure 1C). Walking speed, step length, and cadence were determined using motion analysis software (Nexus 1.8.5, VICON, Oxford, UK). The three-dimensional marker trajectories and the ground reaction forces were filtered (zero-lag, fourth order Butterworth filter, cut-off frequency = 6 Hz) and all gait variables were time-normalized to the stance cycle. Hip joint moments were normalized by bodyweight (BW) and walking speed and step length were further normalized to the participant’s leg length (LL).

Musculoskeletal Modelling

Muscle and hip contact forces were calculated in a musculoskeletal simulation program (OpenSim 3.1, SimTK, USA), using a well-established musculoskeletal model (‘Gait2392’). The model was selected as it was a conventionally used model for lower-limb studies to estimate joint contact forces and muscle activation patterns. The model consisted of the lower limb and torso segments, which modelled the hip and lumbar as ball-and-socket joints,
knee as a custom joint, and ankle as a hinge joint (Figure 1D). The pelvis reference system of the model presents a $13^\circ$ offset in the sagittal plane, with respect to the recommended ISB reference system, which should be taken into consideration when comparing pelvic and sagittal hip flexion-extension angles with previous FAI studies in the literature (without adjusting the estimations obtained from OpenSim).

From CT imaging, each participant’s pelvic width, depth, and height were measured and used to linearly scale the dimensions of the musculoskeletal model, accounting for the distance between virtual and experimental markers. The virtual marker positions were adjusted by registering their locations to their experimental coordinates, after estimating static joint angles through inverse kinematics. The maximum isometric force of the muscles in the musculoskeletal model were customized, using the participants’ anthropometric measurements to calculate the following scaling factor:

$$\frac{M_{\text{exp}} \cdot H_{\text{exp}}}{M_{\text{mod}} \cdot H_{\text{mod}}}$$

where $M_{\text{exp}}$ and $H_{\text{exp}}$ denoted the experimental mass and height, respectively. The $M_{\text{mod}}$ and $H_{\text{mod}}$ measurements referred to original values of the musculoskeletal model’s mass and height parameters ($M_{\text{mod}} = 74.2$ kg, $H_{\text{mod}} = 1.67$ m).

After scaling each model, a series of inverse kinematics, inverse dynamics, static optimization, and joint reaction analyses were performed. Static optimization implemented a quadratic cost function to determine the muscle forces (92 muscles), which included the muscles of interest: rectus femoris, sartorius, tensor fasciae latae, gluteus medius, gluteus minimus, gluteus maximus, semitendinosus, biceps femoris, psoas major, iliacus, and adductors. Muscle contraction dynamics (i.e., force-length-velocity relationship) were neglected, as they were shown to not influence muscle predictions during walking. Static optimization was
effective for muscle force estimations as it did not require modelling muscle dynamics and
previously demonstrated similar results with other rigorous and computationally demanding
algorithms during walking (i.e., dynamic optimization algorithms). Hip contact forces were
calculated as three-dimensional vectors acting on the acetabulum and expressed in the pelvic
coordinate system. Both hip contact and muscle forces were normalized by BW.

Statistical Analysis

Between-group differences for demographics, anatomical variables, and gait parameters were
examined using independent samples t-tests. Hip contact and muscle forces were compared using
a Mann-Whitney U test for non-parametric distributions, given that some of the variables failed
the Lilliefors normality test (CI = 95%). Effect size was calculated with Pearson’s correlation
coefficient \( r \), the non-parametric equivalent of Cohen’s \( d \), and was considered as either small (\( r = 0.1 \)), medium (\( r = 0.3 \)) and large (\( r = 0.5 \)). A sample size calculator (G*Power 3.1.9.3; Heinrich-Heine-Universität Düsseldorf, Germany) determined that the acceptable sample size
per group was 18 (\( n = 36 \)), to seek 80% of statistical power and detect a moderate-to-large effect
size.

RESULTS

Gait Parameters

The FAI patients walked with a slower walking speed (1.44 ± 0.16 m/s/LL) and smaller step
length (0.77 ± 0.07 m/LL), in comparison with the CON participants (1.63 ± 0.17 m/s/LL, \( p = 0.001 \) and 0.85 ± 0.07 m/LL, \( p = 0.004 \), respectively; Table 2). The FAI group demonstrated
slightly reduced functional ROM and joint moments in all planes (Figure 2), showing
significantly lower frontal adduction angle (–4.8 ± 3.8°) and sagittal hip extension moment (0.08
± 0.02 Newton-meter/Bodyweight (Nm/BW)), during contralateral foot-strike (terminal stance phase of gait), compared to the CON group (−7.5 ± 2.9°, p = 0.03, and 0.1 ± 0.01 Nm/BW, p = 0.002 respectively).

Table 2. Summary of cadence, walking speed, and step length parameters of each group

<table>
<thead>
<tr>
<th>Parameter</th>
<th>FAI</th>
<th>CON</th>
<th>p value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cadence (steps/min)</td>
<td>111.9 ± 7.9</td>
<td>116.4 ± 6.6</td>
<td>0.074</td>
</tr>
<tr>
<td>Walking Speed (m/s/LL) †</td>
<td>1.44 ± 0.16*</td>
<td>1.63 ± 0.17</td>
<td>0.001</td>
</tr>
<tr>
<td>Step Length (m/LL) †</td>
<td>0.77 ± 0.07*</td>
<td>0.85 ± 0.07</td>
<td>0.004</td>
</tr>
</tbody>
</table>

† Independent samples t-test, significant difference (p < 0.01) compared with CON
† normalized by leg length (LL)

Figure 2. Hip joint angles (top row) and moments (bottom row) in each of the sagittal (left), frontal (centre), and transverse planes (right), during the gait cycle, for the FAI (red/dashed) and CON groups (blue/solid). Plotted lines and bands denote the mean and standard deviation, respectively. Hip joint moments were normalized by bodyweight (BW) and the full stance phase was represented at ipsilateral foot-strike (IFS), contralateral foot-off (CFO), contralateral foot-strike (CFS), ipsilateral foot-off (IFO), while the gray arrow denotes the statistical difference in adduction and extension moment.

Muscle and Hip Contact Forces

Both groups had similar extensor, abductor, and adductor muscle forces (Figure 3), while the FAI group demonstrated significantly reduced psoas major (median = 1.1, interquartile range (IQR) = 1.0–1.5 Newton/Bodyweight (N/BW)) and iliacus forces (median = 1.2, IQR = 1.0–1.6
N/BW), during contralateral foot-strike (terminal stance of gait), compared to the CON group (median = 1.6, IQR = 1.3–1.6 N/BW, p = 0.004, r = 0.48; and median = 1.5, IQR = 1.3–1.6 N/BW, p = 0.03, r = 0.35, respectively).

Figure 3. Muscle forces during the gait cycle, for the FAI (red/dashed) and CON groups (blue/solid). Muscle forces were normalized by bodyweight (BW) and determined from static optimization. The full stance phase was represented at ipsilateral foot-strike (IFS), contralateral foot-off (CFO), contralateral foot-strike (CFS), ipsilateral foot-off (IFO). Plotted lines and bands denote the mean and standard deviation, respectively. The FAI group demonstrated significantly lower psoas major (top right) and iliacus muscle forces (bottom right), during the contralateral foot-strike, denoted by the asterisk (*).

The FAI group walked with lower hip contact forces in the anterior (median = 2.7, IQR = 2.2–3.9 N/BW), superior (median = 3.7, IQR = 3.5–4.2 N/BW), and medial directions (median = 0.5, IQR = 0.3–0.6 N/BW); compared to the CON group, during contralateral foot-strike, in each of the anterior (median = 4.0, IQR = 2.8–4.4 N/BW, p = 0.026, r = 0.37), superior (median = 4.2, IQR = 3.9–4.9 N/BW, p = 0.02, r = 0.39), and medial directions (median = 0.6, IQR = 0.5–0.7 N/BW, p = 0.038, r = 0.36; Figure 4A). The combination of the three vectors produced a resultant peak force at the anterosuperior aspect of the acetabulum for both groups (1:30 clock-
face orientation), during the contralateral foot-strike; with the FAI group demonstrating a substantially lower resultant magnitude (Figure 4B).

**Figure 4.** A) Hip contact forces during the gait cycle, for the FAI (red/dashed) and CON groups (blue/solid), in all three planes. The full stance phase was represented at ipsilateral foot-strike (IFS), contralateral foot-off (CFO), contralateral foot-strike (CFS), ipsilateral foot-off (IFO). Plotted lines and bands denote the mean and standard deviation, respectively. The FAI group demonstrated significantly lower forces during the contralateral foot-strike, denoted by the asterisk (*). B) The resultant average hip contact forces during the gait cycle, depicted in the sagittal plane of a right hip. Hip contact forces were normalized by bodyweight (BW) and reported with a ‘butterfly’ graph, showing magnitude (radar graph) and direction (acetabular clock-face orientation) with respect to the pelvic coordinate system.

**DISCUSSION**

In this study, we investigated if symptomatic patients with cam FAI demonstrated different muscle and hip contact forces during level walking. We found that the FAI group altered their
walking kinematics and kinetics, resulting in reduced psoas major and iliacus muscle forces and
anterosuperior hip contact force estimations. Knowing that the cam morphology does not likely
induce impingement during activities of daily living involving lower amplitudes of motion (i.e.,
level walking), muscle and hip contact forces estimated using musculoskeletal modelling are
crucial to help elucidate alterations in kinematics and kinetics. Furthermore, the differences in
joint loading may provide a better strategy for athletic conditioning, training program, or non-
surgical management. As Mayne and associates (2017) recently pointed out that knowledge
pertaining to hip muscle strength associated with FAI patients is very limited. Although
previous studies reported joint kinematics and kinetics associated with cam FAI, none of the
studies determined hip contact forces from musculoskeletal modelling, to compare joint loading
between a cohort of FAI and CON participants.

From this study, FAI individuals walked with lower peak hip extension, abduction, and
internal rotation angles, aligning well with previous findings. Many earlier studies of FAI
assumed that limited ROM was attributed to bone-on-bone impingement, caused by the cam
morphology. Although the squatting motion may be more prone to impingement and reduced
hip flexion, bone-on-bone impingement is less likely to occur during level walking. This
further suggests that there may be other parameters, in addition to the bony deformity, that alter
walking biomechanics of FAI participants. Moreover, our FAI group’s reduced step length,
walking speed, and hip extension could be associated with a protective mechanism. All
participants were instructed to walk at their self-selected pace; however, any apprehension to
pain would have reduced muscle forces and, consequently, joint contact forces. Lewis and
associates (2007) previously also noted that reducing hip extension may be beneficial for
individuals with anterior hip pain, instability, or labral tears. Participants completed their
walking trials, at their own self-selected pace, and confirmed that they did not experience any
difficulty or pain. Interestingly, four FAI participants in this study indicated cartilage damage at
the time of their surgery; which could have been further associated with their reduced walking
speed and ROM.\textsuperscript{24, 66}

Although the iliacus and psoas major contributed substantially to the hip contact forces,
with respect to other muscles, the FAI group showed lower peak forces during contralateral foot-
strike, resulting in lower hip contact forces. Both muscles (iliacus and psoas major) function
primarily as hip flexors and structurally combine at the iliopsoas tendon, which tensions directly
over the anterior capsulolabral complex,\textsuperscript{1, 45} from the pelvis’ pectineal eminence to the femur’s
lesser trochanter. During hip extension, the iliopsoas tendon and the adjacent medial branch of
the capsular iliofemoral ligament prevents anterior subluxation, generating a flexion moment.
The tension at the hip capsule interface may be further aggravated by the cam morphology,
which is exposed more anteriorly with hip extension (Figure 5). Domb and associates (2011)
reported varus hips with iliopsoas impingement resulted in labral damage,\textsuperscript{23} where a tighter psoas
tendon induced adverse stresses to the anterior capsulolabral complex during hip extension.\textsuperscript{1, 37, 44}
This was more evident as our FAI group not only walked with a reduced hip extension, but also
showed a lower extension moment. The resultant hip contact forces were directed towards the
anterosuperior acetabulum, from the femur onto the pelvis (in the pelvic reference system),
which would be consistent with locations of chondrolabral damage in FAI patients, noted during
intraoperative and imaging observations.\textsuperscript{8, 9, 25} This certainly supports the concept that not only
extreme ROMs contribute to hip damage in patients with cam FAI but also day-to-day activities.
Figure 5. Sagittal view of the right hip, during walking at: A) heel strike, with the iliopsoas (red) slackened and further away from the anterior femoral head; and at B) contralateral foot-strike, with the tighter iliopsoas closer to the anterior femoral head. Peak hip contact force vectors (orange arrows) occurred during the terminal stance phase of gait, with the hip extended. The FAI participants walked with smaller extension angles and flexion moments, resulting in reduced hip contact forces.

Resultant subject-specific hip contact forces and kinematics are also necessary inputs for finite element simulations, to examine hip joint stresses. In a recent finite element study that examined hip joint stresses due to cam FAI, subject-specific hip contact forces were applied on segmented 3-D models and showed that elevated hip joint stresses were not only dependent on the cam morphology, but were also attributed to secondary subject-specific characteristics (e.g., joint loading direction, anatomical geometry, bone material properties). Moreover, the finite element models that simulated symptomatic cam FAI had characteristic smaller femoral neck-shaft angles, compared to asymptomatic and control individuals. It was also recently
observed that symptomatic individuals indicated higher spinopelvic incidences; secondary to the cam morphology and varus neck parameters, which contributed to lower sagittal hip motions. Anatomically, the combination of the varus neck and large pelvic incidence could result in different force vector direction of the iliopsoas tendon, psoas major, and capsular ligaments, where even a more superiorly positioned lesser trochanter would shorten the iliopsoas, resulting in anterior hip pain. This could suggest why FAI individuals reduced their step lengths and walking speeds. In efforts to alleviate pain, non-surgical treatment of FAI should further emphasize on core stability and muscle strengthening (iliopsoas, flexors, gluteal muscles) to alter sagittal plane moment arms and examine if symptoms would improve. Similarly, asymptomatic cam individuals and postoperative cam patients may also benefit from early targeted muscle strengthening. Not only would the psoas major and iliacus function to protect the hip joint during flexion-extension, proper athletic conditioning could further strengthen core muscles for improved hip mobility and pelvic balance. In the early stages of symptomatic FAI, it would be crucial to strategize effective non-surgical management, in efforts to optimize muscle strengthening regimens and delay the progression of clinical signs.

Limitations

First, our cohort consisted of only males. During the recruitment stage, there were very few female patients who presented themselves with clinical symptoms. As such, it was increasingly difficult to balance the participant groups with female participants without introducing additional sex-, age-, and BMI-related variances. Symptomatic cam FAI is more prevalent in younger, athletic males; however, statistical associations between sex, activity level, and symptoms is warranted in the future. Also, our symptomatic group was slightly older. Although age can influence walking and hip motions, several of older FAI participants not only demonstrated
higher amplitudes of hip motions and hip contact forces, but also faster walking speeds than control participants. Second, magnetic resonance imaging (MRI) data of the entire lower limbs were not available, hence the musculoskeletal model used in this study was based on cadaveric measurements and did not reflect detailed subject-specific anatomy. Nonetheless, the current simulations were comparable with instrumented prostheses’ data where hip contact forces also reached up to 5 BW during similar walking speeds, but further reflected the importance to account for muscle contributions in joint loading. Third, the model was used within a consistent framework to compare a pathological (FAI) to a healthy (CON) group. Muscle forces were calculated using static optimization, which may not be able to model the muscle co-contraction mechanisms altered by a joint pathology. However, static optimization provided muscle activations consistent with EMG measurements during various walking speeds, and did not require the additional measurements of muscle activities that typical EMG-driven models would. Moreover, quadratic cost functions in static optimization have been previously implemented to estimate muscle forces, while also demonstrating similar results to computationally demanding algorithms. Fourth, in addition to the comparative framework of our study design, including an asymptomatic group of FAI participants (with cam deformities but no clinical signs or pain) through a randomized control trial would further provide further insights into differences in muscles contributions and hip joint loading.

CONCLUSION

This study provided insights into the link between cam FAI and hip contact forces. The FAI group’s altered gait mechanics (especially in hip extension) demonstrated a protective
mechanism, to reduce psoas major and iliacus muscle forces, and resulted in lower hip contact
forces.

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REFERENCES


